Title: Increasing step width reduces the requirements for subtalar joint moments and powers.

Type: Original Article

Authors: Jayishni N. Maharaj, Lauren E. Murry, Andrew G. Cresswell, Glen A. Lichtwark

School of Human Movement and Nutrition Sciences,

The University of Queensland,

Brisbane, 4072, Queensland, Australia

Corresponding Author. Jayishni Maharaj,

School of Human Movement and Nutrition Sciences,

The University of Queensland,

Brisbane, 4072, Queensland, Australia

Tel: +61 7 3365 6482,

Fax: +61 7 3365 6877.

Email address: jayishni.m@uq.edu.au

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ABSTRACT

The subtalar joint (STJ) contributes to the absorption and generation of mechanical energy (and power) during walking to maintain frontal plane stability. Previous observational studies have suggested that there may be a relationship between step width and STJ supination moment. This study directly tests the hypothesis that walking with a step width greater than preferred would reduce STJ moments, energy absorption, and power generation requirements, while increasing energy absorption at the hip during initial contact. Participants (n = 12, 7 females) were asked to walk on an instrumented treadmill at a constant velocity and cadence at a range of fixed step widths ranging from 0.1 - 0.4 times leg length (L). Walking at step widths greater than preferred (0.149 ± 0.04 L) reduced peak STJ moments at initial contact and propulsion which subsequently reduced the negative and positive work performed at the STJ. There was a 43% reduction in energy absorption (negative work) and approximately 30% decrease in positive work at the STJ as step width increased from 0.1L to 0.4L. An increase in energy absorption at the knee and hip was evident with an increase in step width during initial contact, although minimal mechanical changes were observed at the proximal joints during propulsion. These results suggest an increase in step width reduces the forces generated by muscles at the STJ across stance and is therefore likely to be beneficial in the prevention and treatment of their injuries. In terms of rehabilitation, the increase in mechanical costs occurring due to an increase in energy absorption by the hip and knee is of minimal concern.

INTRODUCTION

Muscles and tendons play a role in both absorbing and generating mechanical energy (and power). The musculoskeletal structures within the foot, such as those at the subtalar joint (STJ), contribute to the mechanical requirements of the lower limb during walking (Arndt et al., 2004; Piazza, 2005). The STJ generally functions to maintain frontal plane stability as it absorbs energy during pronation following heel strike and generates power for STJ supination at push-off (Piazza, 2005), with the tibialis posterior (TP) muscle and tendon considered a key muscle in recycling energy across the stance phase (Maharaj et al., 2016). Numerous factors influence the dynamics of the STJ during walking, including the properties of the TP tendinous tissue (Maharaj et al., 2017a), footwear (Maharaj et al., 2018), foot structure (Maharaj et al., 2017b) and gait parameters (Maharaj et al., 2017b; 2016).

Deviations in step width may alter frontal plane mechanics and hence the function of its constituting structures such as the STJ. We recently undertook an observational study which demonstrated a moderate linear relationship between step width and STJ moments, with wider step widths being associated with lower STJ moments (Maharaj et al., 2017b). Coupled with the finding that a narrow step width is linked to an increase in TP muscle activation (Maharaj et al., 2018), it is likely that step width may reduce the required TP muscle force generation and subsequent stresses and strains imposed on its tendon. TP tendon strain amplitude has also been directly related to energy absorption at the STJ during initial contact (Maharaj et al., 2017a) and thus a reduction in the STJ moment could also modify the power absorbed/generated at the STJ (joint power = joint moment x joint angular velocity). Therefore, step width could be a modifiable gait parameter that may serve as conservative treatment for lower limb pathologies, such as posterior tibialis tendon dysfunction, by reducing the STJ moments or energy absorption demands during early stance. With evolving interest in changing gait mechanics to minimise injury, a better understanding of how gait characteristics influence joint function during walking is needed.

Increases in step width are known to raise the metabolic costs of walking (Donelan et al., 2001). The increase in metabolic cost has been directly related to an increase in mechanical demands of lower limb muscles to redirect the CoM, particularly during double support (Donelan et al., 2001). In particular, during the initial contact phase there is a greater requirement to absorb energy as step width deviates from the preferred step width (Donelan et al., 2001). Given that we expect energy absorption at the STJ to decrease during this period, then it follows that structures at more proximal joints, are likely to absorb the additional energy necessary with a wider step width. For instance, the additional absorption of energy may occur at the hip (Rankin et al., 2014; Wang and Srinivasan, 2014), as it plays a significant role in generating medio-lateral forces during walking (Neptune and McGowan, 2016; Pandy et al., 2010). However, there have been no studies investigating the effects of step width on STJ mechanics and potential concomitant changes in hip, knee and ankle mechanics.

To our knowledge, the effect of step width on the mechanical demands of the STJ has not been experimentally tested and any mechanical consequences on proximal structures have also never been explored. The primary objective of this study was to investigate the effect of step width on lower limb mechanics and energetics during walking, with a particular focus on the effect on the STJ. We hypothesised that walking with a step width greater than preferred would reduce moments and hence the energy absorption and power generation requirements of the STJ but would increase the energy absorbed at the hip during initial contact.

METHODS

Twelve individuals (7 female; height: 1.73 ± 0.08 m; body mass: 67 ± 8 kg; age: 23 ± 3 yrs.) with no history of lower limb injury in the previous two years or known neurological impairment gave informed consent to participate in the study. The local university ethics committee approved the protocol, which was conducted according to the declaration of Helsinki.

Familiarisation

Familiarisation consisted of two sets of 10-min walking on a treadmill at a standardized walking velocity established as a function of leg length for each participant. The walking velocity was calculated using Froude's number .25, which is close to the velocity that minimizes cost of transport (Minetti, 2001). Participants walked at their preferred step width in the first trial and at the widest width in the second trial. Preferred cadence of each participant was calculated during the first trial by timing twenty steps over the last 2-min and was also maintained during the second trial (see below).

Experimental Protocol

Participants walked barefoot on a force instrumented tandem treadmill (DBCEEWI, AMTI, USA) at varying step widths whilst 3-dimensional (3D) motion capture and ground reaction forces were synchronously recorded. These step widths included each subject's preferred width and step widths equal to 0.10, 0.20, 0.30, and 0.40 times leg length (L, height from the ground to greater trochanter). Step widths were enforced by instructing participants to have their second toe roughly in contact with a straight-line laser beam, which was aligned along the treadmill belt, during the foot contact period. The step frequency was controlled using a metronome which was determined during the first familiarisation trial at preferred walking velocity (Froude's number .25). The order of step widths were randomised and participants were given a 2-min period to adapt to the width prior to 20-sec of continuous data collection.

An eight camera, 3D opto-electric motion capture system (Qualysis, Gothenburg, Sweden) was used to capture (200 Hz) the position of 40 reflective markers (9 mm) placed on the pelvis, thigh, shank and foot of both limbs. The marker set consisted of a four-marker cluster on the thigh and shank segments and anatomical markers placed over the right and left anterior-superior iliac spines; right and left posterior-superior iliac spines; medial and lateral femoral epicondyles; medial and lateral malleoli. For the feet, a multi-segment marker set was used to track the motion of the rear-, mid- and fore-foot segments using 8 anatomical marker locations previously detailed (Leardini et al., 2007; Maharaj et al., 2017b).

Ground reaction forces and moments were sampled from plates embedded under front and rear split belts at 1000 Hz. Participants were instructed to position themselves on the treadmill such that heel strike occurred on the front plate and toe off on the rear. This protocol allowed separate left- and right- limb contributions and gait events to be established.

A relaxed standing trial was first captured and the positions of the anatomical markers were used to scale a 15-segment (pelvis; bilateral thigh, shank, talus, rear-foot, mid-foot, fore-foot, digits) musculoskeletal model. The calibration process scales the mass, dimensions and inertial parameters of each segment to match the anthropometrics of the individual, using anatomical reference markers. Models were scaled from a generic 3D musculoskeletal model available in OpenSim software (Delp et al., 2007), which was modified so that each foot consisted of five segments: talus, calcaneus, mid-foot, fore-foot and toes. The hip, knee and ankle had six degrees of freedom (DOF, three translational and three rotational) while the multiple segments in the foot were ascribed an oblique axis to rotate about the three orthogonal coordinate planes simultaneously, see Maharaj et al. (2017) for more details. The STJ was modelled as a revolute joint with a joint axis of 37° inclination and −21° deviation to the midline of the body (Lewis, 2007).

Joint kinematics and kinetics

Marker positions and analogue force platform signals were filtered at a common frequency using a second-order low-pass Butterworth filter with a cut-off frequency of 25 Hz (van den Bogert and De Koning, 1996). Inverse kinematics and inverse dynamic analyses were conducted in OpenSim V3.3 to output joint angles and moments, respectively. Briefly, the inverse kinematics analysis performs a weighted least-squares fit of the model markers to the experimental marker positions to get the pose of the model at each point in time and from which joint angles are calculated. The inverse dynamics analysis combined the model kinematics and the measured ground reaction force data of each foot to compute net joint moments at the subtalar, ankle, knee and hip joints using OpenSim (Delp et al., 2007). These moments were multiplied by joint angular velocities (the first derivative of joint angles) to obtain instantaneous joint powers for the subtalar, ankle, knee and hip. Joint angles, moments and powers were computed in all three dimensions.

Individual limb power

We calculated the instantaneous limb power generated by the right leg using the individual limbs method as detailed by Donelan et al. (2001). Briefly, individual limb power was calculated as the dot product of the right limb's GRF vector and the velocity of the centre of mass (CoM). CoM velocity was obtained by integrating the CoM acceleration (net forces divided by body mass) and subsequently adding treadmill belt velocity to the fore-aft component.

Joint and individual limb powers were integrated with respect to time over the negative power period in initial contact and over the positive power period in propulsion to compute negative and positive work respectively. Net mechanical work at each joint was calculated by integrating power over the entire stride cycle (defined from heel-strike to subsequent heel-strike) and summed to compute summed subtalar-ankle–knee–hip work. Net CoM mechanical work was calculated by integrating individual limb power over the entire stride cycle. Step length was calculated using centre of pressure measures and was defined as the anterior-posterior distance between the right and left limb in the direction of progression. Walking velocity was divided by step length and multiplied by 60 s/min to compute step frequency (steps/min).

Statistical analysis

Each individual's kinematic, kinetic, and individual limb power data were time normalized to 101 points and averaged over at least three strides. Discrete variables were calculated during *initial contact* and *propulsion*, defined when individual limb powers either dropped below or raised above zero. The main outcome variables analysed were: step length, cadence, summed subtalar-ankle–knee–hip work, net CoM mechanical work, peak STJ displacements; peak STJ moments; subtalar, ankle, knee, hip negative and positive work. Prior to further analysis, a D'Agostino-Pearson omnibus test was used to check the normality of data. A repeated measures analysis of variance (ANOVA) was used to describe the effects of step width. Statistical differences were established at $P \leq 0.05$. Results are presented as mean and ± standard deviation unless otherwise stated.

RESULTS

During the preferred condition, participants walked with a mean step width of 13 ± 3 cm, which was 0.149 ± 0.04 L. The mean step length and step frequency during the preferred condition was 71 ± 3 cm and 123 ± 5 steps/min respectively. The step length and frequency during all walking conditions were no different to the preferred condition, see Figure 2A & B. Net mechanical work, calculated as the sum of hip, knee, ankle and STJ work or the individual limb work, was close to zero in the preferred condition and remained invariant across all conditions (Figure 2C). Summed subtalar-ankle-knee-hip work (0.2 ± 0.07 $J.kg^{-1}$) was slightly higher than CoM mechanical work (0.02 ±0.05 $J.kg^{-1}$).

During the initial contact period, structures at the STJ absorbed energy by generating a supination moment to resist STJ pronation. As step width increased, a less pronated STJ in initial contact was evident ($p<0.01$, Figure 3A), while kinetic analysis revealed lower supination moments during initial contact and push off ($p<0.01$, Figure 3B). STJ negative work (or negative area under the power curve, Figure 3C) significantly decreased ($p<0.01$), with 43% less energy being absorbed at the STJ when step width was increased from 0.1L to 0.4L. Approximately 29% less positive work was generated by the structures at the STJ in propulsion when step width increased from 0.1L to 0.4L.

Figure 4 illustrates the mean CoM, hip, knee, ankle and subtalar joint power trajectories. In the preferred condition, the structures at the knee had the greatest contribution to negative work performed during initial contact (23%), followed by the ankle (18%), hip (14%), and STJ (4%). During propulsion, the structures at the ankle (49%) had the greatest influence on positive work, followed by the hip (28%), STJ (18%), and knee (5%). Increasing step width progressively increased the negative work performed by the limb during initial contact ($p < 0.01$) but had minimal effect on positive work done during propulsion (Figure 5). As step width increased, there were no differences in the work performed at the ankle during initial contact, although significant increases in negative work were evident at the knee and hip ($p < 0.01$). The power generated by the ankle, knee and hip during propulsion did not change significantly across conditions.

DISCUSSION

Consistent with our hypothesis, an increase in step width reduced STJ moments across stance and hence the energy absorption requirements at initial contact and positive work during propulsion. As predicted, an increase in step width increased the overall negative work requirements of the leg, which was achieved through an increase in work absorbed at the knee and hip during initial contact. Minimal mechanical changes were observed at the proximal joints during propulsion. Our findings suggest that changes in step width may reduce the energy absorption and power generation requirements of the soft tissues at the STJ with nominal mechanical impact on proximal joint structures. These assumptions need to be confirmed using muscle level analysis such as electromyography as reductions in net STJ moments may be a result of an increase in antagonist muscle activation or a decrease in TP force generation. The implications of modifying step width to help prevent or treat overuse injuries are discussed below.

The reduced mechanical work performed at the STJ was driven by a decrease in joint moments, which is due to a shorter STJ moment arm relative to the ground reaction force vector. We suspect that an increased step width results in reduced forces of the muscles that contribute to STJ supination moments (primarily the TP and tibialis anterior) as the ground reaction force vector moved closer to the STJ with increased step width. We computed the mean distance between the STJ centre and the ground reaction force vector in one subject across 10 strides and found a difference of 15 ± 4 mm between the 0.4L and 0.1L conditions. These results are also consistent with recent findings which show that walking at a faster velocity reduced step width and increased the activation of the TP muscle (Maharaj et al., 2018). However, it would be interesting to determine if the change in force requirement with an increase in step width also reduces the activation requirements of the TP muscle. A potential reduction in TP muscle force generation with increasing step width should have the effect of reducing TP tendon stresses and strain during initial contact (Maharaj et al., 2016). However, the subsequent reduction in STJ positive work in propulsion may require greater work to be performed by structures at the ankle, which may be masked in our results due to their larger more significant role in the sagittal plane.

Modifying an individual's step width may have metabolic consequences, particularly as preferred step width is closely aligned with minimising energetic costs of walking (Donelan et al., 2001). Donelan et al. (2001), have previously demonstrated substantial metabolic costs for walking at step widths greater than preferred, 0.13 times leg length, which is similar to the preferred condition in this study, 0.149 L. The increase in energy absorption at the knee and hip observed in this study may explain the increased metabolic cost of walking with wider steps. Proximal muscle-tendon units at the knee and hip are known to be less efficient than distal limb muscles like the triceps surae or TP. This is partly because of their reduced capacity to recycle elastic energy and an increased requirement to activate larger volumes of muscle as these muscles typically have longer fibre lengths (Biewener and Roberts, 2000; Farris and Sawicki, 2012; Roberts et al., 1997). Metabolic cost plays an essential role in the movement strategy employed by humans, however, it's contribution is not exclusive (Summerside et al., 2018). Gait movement is a weighted decision of several parameters based on the individual including stability, safety and/or perceived comfort, parameters (Daley, 2008; Daley and Usherwood, 2010; Grimmer et al., 2008). During rehabilitation, lowering the energy cost of locomotion may not be of high priority and a comprise to offload an injured tissue or avoid painful gait should be perceived of higher importance.

There are limitations in the generalisability of our results. Firstly, while participants were given substantial time to familiarise to the various step width conditions, it is uncertain whether the observed biomechanical changes persist beyond the short term. It is possible that after a period of habituation further mechanical changes may occur. Secondly, step width in this study was manipulated via laser lines, which were projected on to the treadmill as the participants walked. This method provided sufficient preparatory time for the participant to meet the step width constraint. However, enforcing the step width target may itself exact altered gait mechanics. These constrains were imposed at all step width conditions and is thus likely to have a similar effect across all trials. Finally, due to the constraints of the musculoskeletal model some of the limb energy due to translation of joints and soft tissue is not accounted for (Zelik and Kuo, 2010; Zelik et al., 2015).

In conclusion, our findings indicate that an increase in step width reduces STJ moments across stance and hence the work performed by its soft tissue structures. The reduction in STJ loading could reduce the stresses and strains of the TP tendon and thereby be beneficial in the prevention and management of its dysfunction. The increase in energy absorption at the hip and knee is likely to increase the mechanical costs of walking, although this may be of minimal concern for rehabilitation.

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CONFLICT OF INTEREST

The authors declare that they have no conflict of interest, financial or otherwise, related to the materials discussed in this manuscript.

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FIGURES

Figure 1: Images illustrating the four step width conditions assessed in the study including 0.1, 0.2, 0.3, and 0.4 times leg length.

Figure 2: The effect of step width on A. normalised step length (top), B. step frequency (middle), and net mechanical work in the lower limb (bottom). Step length and frequency remained consistent across all conditions. Net mechanical work, right limb (triangles) and summed subtalar-ankle-knee-hip work (squares), remained close to zero and similar across all conditions. Step length was normalised to leg length (m/m). Plots represent group mean and error bars SD.

Figure 3: Mean subtalar joint (STJ) displacement (top), normalised moments (middle) and normalised power (bottom) over a gait cycle across varying step width conditions. During initial contact, STJ displacement, moments and negative work decreased with an increase in step width. At propulsion, STJ positive work and moments dropped at wider steps with minimal effect on STJ displacements. Plots represent group mean ensembles for all subjects (n = 12).

Figure 4: Mean centre-of-mass (CoM), hip, knee, ankle and subtalar joint power tracjectories, from top to bottom. Calcuations of power were performed in all three dimenisions. CoM power typically followed a consistent pattern of negative and positive work oscillations and was used to define initial contact and propulsion. The area within the negative power plot illustrates negative

work or energy absorption and the area of the positive power plot indicates positive work or power generation.

Figure 5: Changes in negative and positive work at the subtalar joint, ankle, knee and hip from left to right during walking. Subtalar asterisks indicate that an increase in step width significantly decreased negative and positive work at the STJ, while knee and hip asterisks indicate an increase in negative work at the knee and hip with increased step width.

Figure 1

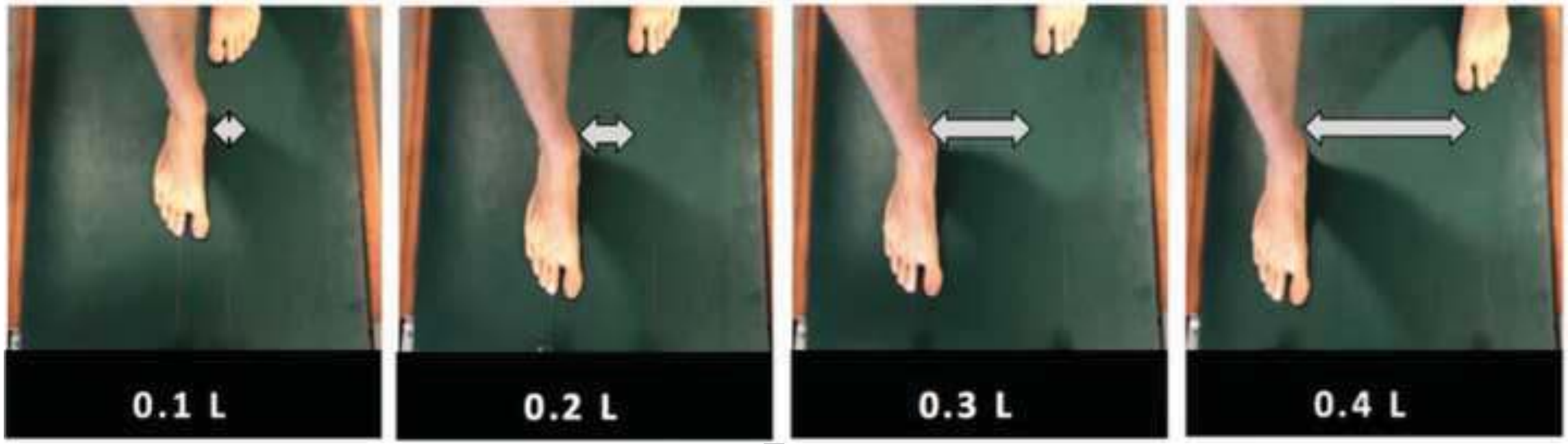
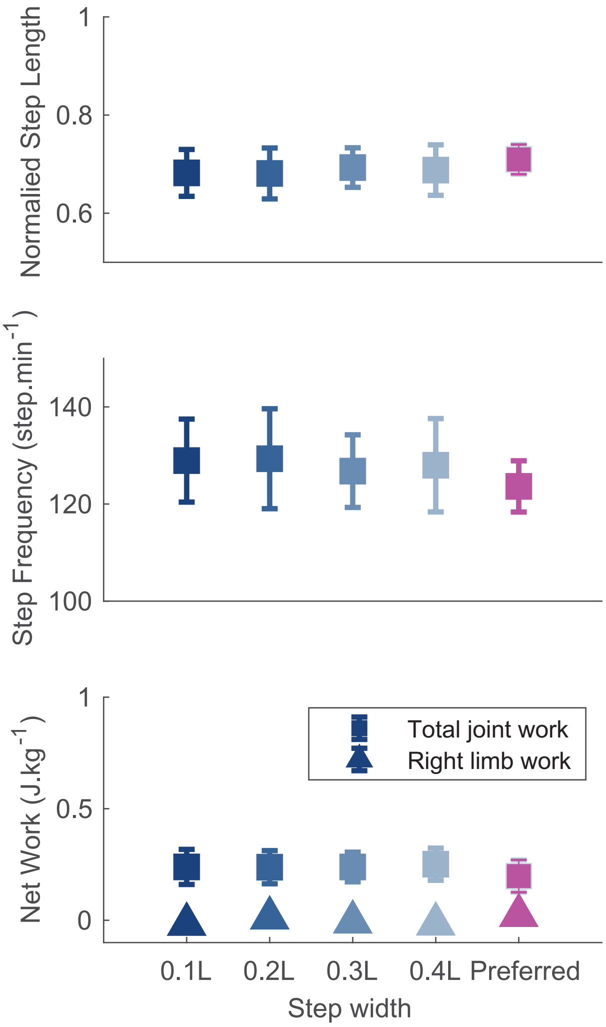
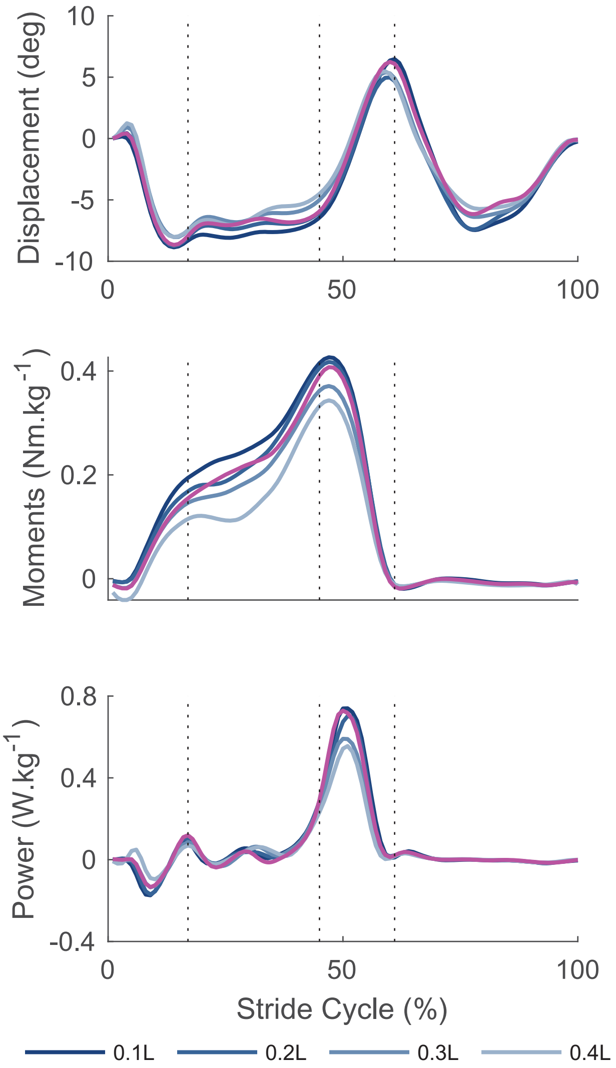


Figure 2



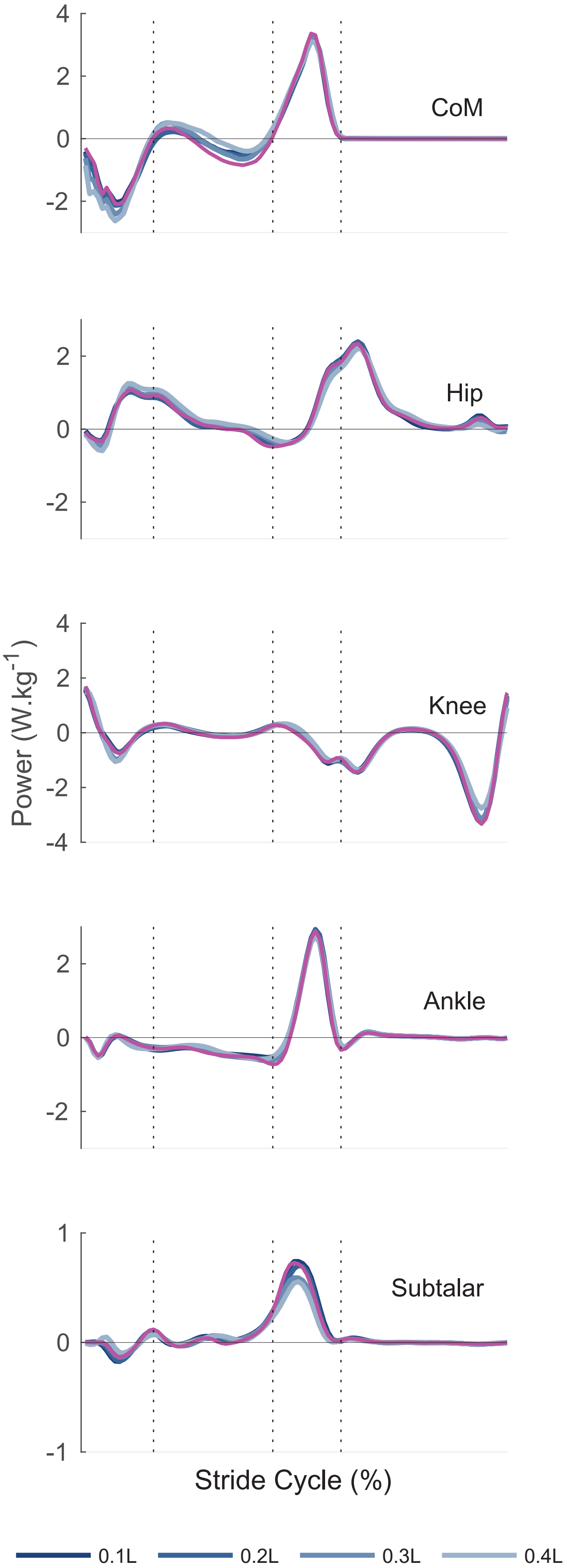
Normalied Step Length
1
0.8
0.6
Step Frequency (step.min^{-1})
140
120
100
Net Work (J.kg^{-1})
1
0.5
0
Total joint work
Right limb work
0.1L
0.2L
0.3L
0.4L
Preferred
Step width

Figure 3



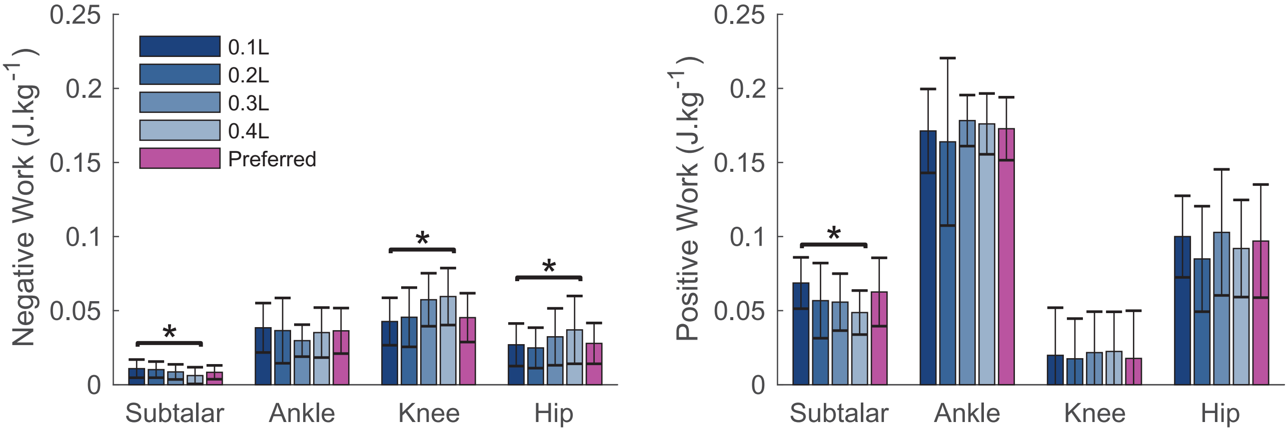
Displacement (deg)
10
5
0
-5
-10
0
50
100
Moments (Nm.kg^{-1})
0.4
0.2
0
Power (W.kg^{-1})
0.8
0.4
0
-0.4
0
50
100
Stride Cycle (%)
0.1L
0.2L
0.3L
0.4L

Figure 4



CoM
Hip
Knee
Ankle
Subtalar
Power (W.kg^{-1})
Stride Cycle (%)
4
2
0
-2
-4
1
-1
0.1L
0.2L
0.3L
0.4L

Figure 5



0.25
0.2
0.15
0.1
0.05
0
Negative Work (J.kg^{-1})
0.1L
0.2L
0.3L
0.4L
Preferred
*
*
*
Subtalar
Ankle
Knee
Hip
0.25
0.2
0.15
0.1
0.05
0
Positive Work (J.kg^{-1})
*
Subtalar
Ankle
Knee
Hip