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The influence of a unilateral fixed ankle on metabolic and mechanical demands during walking in unimpaired young adults

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ABSTRACT

The plantarflexors provide a major source of propulsion during walking. When mechanical power generation from the plantarflexor muscles is limited, other joints may compensate to maintain a consistent walking velocity, but likely at increased metabolic cost. The purpose of this study was to determine how a unilateral reduction in ankle plantarflexor power influences the redistribution of mechanical power generation within and across limbs and the associated change in the metabolic cost of walking. Twelve unimpaired young adults walked with an ankle brace on the dominant limb at 1.2 m/s on a dual-belt instrumented treadmill. Lower extremity kinematics and kinetics as well as gas exchange data were collected in two conditions: (1) with the brace unlocked (FREE) and (2) with the brace locked (FIXED). The brace significantly reduced ankle plantarflexion excursion by $12.96 \pm 3.60^{\circ}$ ($p < 0.001$) and peak ankle mechanical power by 1.03 ± 0.51 W/kg ($p < 0.001$) in the FIXED versus FREE condition. Consequently, metabolic power (W/kg) of walking in the FIXED condition increased by 7.4% compared to the FREE condition ($p = 0.03$). Increased bilateral hip mechanical power generation was observed in the FIXED condition ($p < 0.001$). These results suggest that walking with reduced ankle power increases metabolic demand due to the redistribution of mechanical power generation from highly efficient ankle muscle-tendons to less efficient hip muscle-tendons. A within and across limb redistribution of mechanical workload represents a potential mechanism for increased metabolic demand in pathological populations with plantarflexion deficits or those that walk with an ankle–foot orthosis that restricts range of motion.

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1. Introduction

Individuals with neurologic injuries (e.g. post-stroke) commonly exhibit diminished strength of ankle dorsiflexion and/or plantarflexor musculature (Bohannon and Andrews, 1998; Hsu et al., 2003; Kerrigan et al., 1998). Deficits in dorsiflexor and plantarflexor force production cause different, yet related problems with gait. Dorsiflexor weakness can result in dropfoot, such that the foot remains in a plantarflexed position during swing phase. As many as 20% of individuals post stroke demonstrate dropfoot (Wade et al., 1987) and commonly adopt compensatory hip strategies, such as hip hiking or circumduction, to avoid tripping on the toes during swing phase (Kerrigan et al., 1998; Kesar et al., 2010; Olney and Richards, 1996).

The most common treatment for dropfoot is an ankle–foot orthosis (AFO), which immobilizes the ankle in a neutral position. While an AFO assists weakened dorsiflexors, it can also reduce plantarflexion torque (Burdett et al., 1988). Because the plantarflexor muscle group provides the majority of power generation during terminal stance (i.e. push-off) to propel the body forward (McGowan et al., 2010, 2008; Sawicki et al., 2009), a reduction in plantarflexor torque represents an important contributor to reduced walking velocity in people with neurological injuries (Hall et al., 2011; Nadeau et al., 1999; Olney et al., 1994). Identifying compensatory mechanisms resulting from limited plantarflexion power generation could improve and help focus rehabilitation strategies.

To overcome deficits in joint mechanical power generation, compensations at other joints within and across limbs must occur to maintain a consistent walking velocity (Allen et al., 2011; Cruz et al., 2009; Lewis and Ferris, 2008). For example, reduced ankle torque generation on one side of the body can be compensated for by generating more mechanical power with the contralateral limb (Allen et al., 2011; Chen et al., 2003; Cruz et al., 2009; Jonkers

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et al., 2009). Increasing contralateral ankle power generation specifically, might limit increases in metabolic cost by making use of the ankle's compliant muscle-tendons (i.e., longer elastic tendons and shorter fascicles) which may be more efficient than knee or hip muscles (Sawicki et al., 2009). On the other hand, recruitment at more proximal joints (i.e. hip or knee) on the contralateral and/or ipsilateral limbs may be necessary to maintain sufficient mechanical power generation to maintain walking velocities (Lewis and Ferris, 2008; Nadeau et al., 1999). Increased reliance on mechanical power generation from these less efficient proximal joints (Nadeau et al., 1999) may increase metabolic energy expenditure and contribute to reduced walking endurance.

The purpose of this study was to determine the mechanical and metabolic effects of walking with unilaterally reduced ankle mechanical power output. We hypothesized that restricting ankle motion and power generation unilaterally would induce (1) an increase in mechanical power generation at the contralateral ankle and hip and (2) an increase in ipsilateral hip mechanical power generation. Importantly, we expected the redistribution of lower-limb joint power generation to increase the net metabolic cost of walking.

2. Materials and methods

2.1. Subjects

Twelve unimpaired individuals (6F/6M; 24.5 ± 4.6 years old; 171.9 ± 10.7 cm; 69.7 ± 9.5 kg) participated in this study. All subjects read, understood and signed a consent document approved by the IRB of the University of North Carolina at Chapel Hill prior to data collection.

2.2. Procedures

Subjects wore a lockable ankle brace (Ballert International, LLC, FL; mass= 1.35 kg, Fig. 1a) on the dominant limb while walking on a dual belt instrumented treadmill (Bertec Corp., Columbus OH, Fig. 1b) at 1.2 m/s (2.7 mph). Subjects wore their own running style shoe on the unbraced limb. We tested each subject during two counterbalanced conditions on the same day. When locked (FIXED) with the ankle in a neutral position, the brace restricted angular motion with the intention of reducing ankle power generation. In the unlocked condition (FREE), the subjects could freely move the ankle in the sagittal plane between 40° of dorsiflexion and 40° of plantarflexion.

A wireless metabolic cart (OxyCon, Yorba Linda, CA) recorded breath-by-breath flow rates of O_2 and CO_2 (milliliters per kilogram per minute) that were inspired/expired during trials. Prior to data collection, the metabolic cart was calibrated with known volumes and concentrations of gas. First, baseline gas flow rates of each subject were recorded while standing quietly for five minutes. Then, each subject walked for five minutes to ensure a metabolic 'steady state'. Beginning with the fifth minute, kinematic and kinetic data were collected for 30 s (described below) and this interval was used for all subsequent analyses. Between conditions, a break of at least four minutes was provided to return metabolic data to resting values before initiating the next condition.

An eight camera motion capture system (MX40+, VICON/PEAK, Denver, CO) collected three dimensional coordinates of the pelvis and both feet, shanks, and thighs (120 Hz) using 14 mm reflective markers. Clusters of four markers each were placed on rigid thermoplastic shells and secured to the posterior lateral thighs and shanks bilaterally with elastic wraps (Fig. 1b). The shank cluster attached to the dominant limb was placed under the brace to track the motion of the bony segment while wearing the brace. A thermoplastic shell with three markers was secured to the posterior pelvis and additional markers were placed at the head of the second metatarsal, mid-point of the posterior heel, and the head of the fifth metatarsal to track the motion of the foot. Ground reaction force data (960 Hz) were measured from the dual belt instrumented treadmill and synchronized with the motion capture system. While walking on the treadmill, all participants wore a safety harness that did not restrict lower extremity movement and provided no body weight support.

2.3. Data processing

Sampled breath-by-breath metabolic data (VO_2 and VCO_2) were averaged over the 30 s interval of interest (e.g., beginning of minute 5) to calculate metabolic power (Brockway, 1987). Physiologic steady state in each condition was confirmed by comparing average VO_2 during the first five and last five seconds of the 30 s interval using a paired sample *t*-test (p=0.991). Mass-specific net metabolic power (W/kg) was calculated by subtracting the baseline standing energy expenditure from the walking data and normalizing to body mass.

Marker trajectory and GRF data were filtered with 6 and 20 Hz low pass filters, respectively. A seven segment kinematic model tracked the motion of the lower limbs and pelvis segments. Lower extremity sagittal plane joint angles (hip, knee, and ankle) were calculated using Euler angles (deg.) and the instantaneous angular velocity (deg./s) was calculated as the first derivative of the joint angle using Visual 3D (C-Motion, Germantown, MD). Joint powers were calculated as the product of sagittal plane joint moments and angular velocities and normalized to body mass.

For each kinematic and kinetic variable, outcome measures were calculated for each step over the 30-s collection interval and then averaged for each condition using custom written LabVIEW software (National Instruments, Austin, TX). Ankle excursion was quantified as the absolute difference between the maximum dorsiflexion angle and maximum plantarflexion angle during the stride cycle. The peak positive power at the ankle and hip was determined during the last 30% of stance phase from each step and averaged over each 30-s trial (W/kg). In addition, the integral of the positive component of the average lower extremity joint power curves (i.e., ankle, knee, and hip) was calculated over the gait cycle for the braced and unbraced limbs in each condition, then divided by the average stride time to yield average positive power for each joint. These average positive

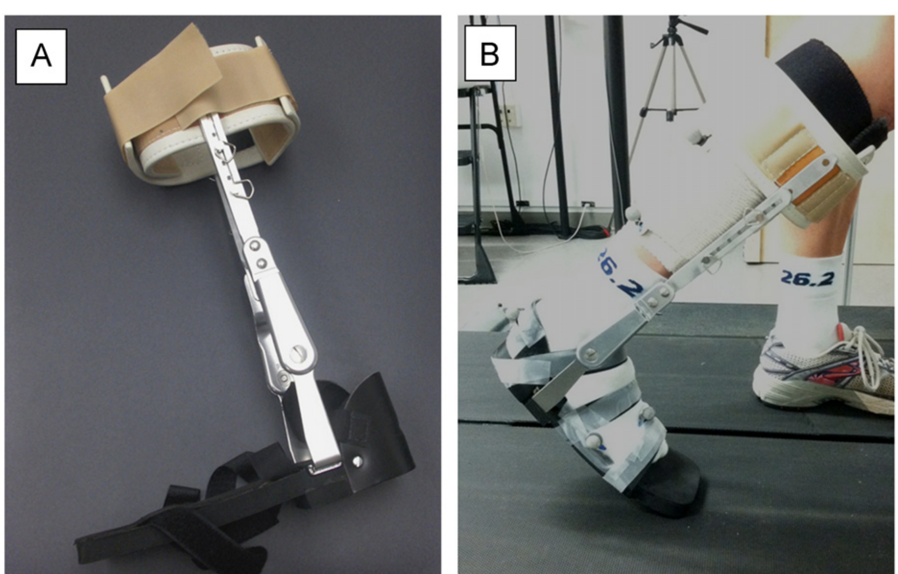

Fig. 1. (A) Lockable ankle brace (Ballert International, LLC, FL). (B) Lockable ankle brace restricting ankle motion during treadmill walking at 1.2 m/s.

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power contributions from each joint were summed to determine the total average positive power for each limb (W/kg) (Farris and Sawicki, 2012b):

$$\overline{P}_{total} = \overline{P}_{ankle} + \overline{P}_{knee} + \overline{P}_{hip}$$

where $\overline{P}_{ankle}$, $\overline{P}_{knee}$, and $\overline{P}_{hip}$ represents the average positive power generated by the ankle, knee, and hip muscle-tendons, respectively. For graphical depiction, the average percent contribution of positive power generation was calculated for the braced and unbraced limbs as

$$\overline{P}_{joint} \,/\, \overline{P}_{total} \times 100 = \%\text{contribution}$$

for the ankle, knee, and hip in both the FREE and FIXED conditions.

2.4. Data analysis

Statistical analyses were conducted using SPSS (SPSS 16.0, Chicago, IL). Mass specific net metabolic power (W/kg) and total average positive mechanical power generation (combined power generated by both limbs; W/kg) were compared between the FREE and FIXED conditions using paired t-tests ($\alpha = 0.05$). Seven separate two-way repeated measures ANOVAs (condition x limb) were performed for peak plantarflexion angle, ankle excursion, peak mechanical power generation at the ankle and hip and average positive power generation at the ankle, knee, and hip. Post hoc testing was performed with Bonferoni corrected paired t-tests.

Pearson product-moment correlation coefficients were computed to determine potential relationships between the change in mass-specific net metabolic cost and the change in both peak and average mechanical power generation at the ankles and hips between the FIXED and FREE conditions.

3. Results

3.1. Net metabolic power

Mass specific net metabolic power was 7.4% higher in the FIXED ankle condition (3.76 ± 0.55 W/kg) than during the FREE condition (3.50 ± 0.58 W/kg; $p = 0.030$).

3.2. Kinematics

Significant interaction effects were observed at the ankle (ankle excursion, peak plantarflexion angle, both $p < 0.001$), such that walking with the ankle FIXED (Fig. 2) yielded a significant reduction in the braced ankle's excursion (FREE: $27.12 \pm 3.63°$; FIXED: $14.16 \pm 2.80°$; t-test: $p < 0.001$) and peak plantarflexion angle (FREE: $-8.51 \pm 4.67°$; FIXED: $1.61 \pm 3.63°$; t-test: $p < 0.001$) compared to the FREE condition. Ankle excursion did not differ when the brace was unlocked in the FREE condition compared to the unbraced limb in the FREE condition (braced: $27.12 \pm 3.63°$ vs. unbraced: $28.8 \pm 5.58°$). Peak plantarflexion angle, however, was reduced in the braced limb in the FREE condition ($8.51 + 4.67°$) compared to the unbraced limb in the FREE condition ($18.28 + 5.34°$; $p < 0.001$).

3.3. Joint kinetics

Peak Powers: Walking in the FIXED condition altered peak mechanical power generation at both the ankle and the hip ipsilaterally (Fig. 3). A significant interaction effect (condition x limb) was observed for both peak ankle ($p < 0.001$) and hip ($p = 0.030$) power generation. Peak ankle power generation in late stance was reduced in the braced limb during the FIXED condition compared to the FREE condition ($p < 0.001$) (Fig. 4). In the FIXED condition, we found a significant reduction in the braced limb's peak ankle power generation compared to the unbraced limb ($p < 0.001$), reflecting a mechanical asymmetry between limbs. At the hip, peak power generation at terminal stance in the braced and unbraced limbs was greater during the FIXED condition compared to the FREE condition (braced limb: $p = 0.022$; unbraced limb: $p < 0.001$), however there was no difference between limbs in the FIXED condition (Figs. 3 and 4).

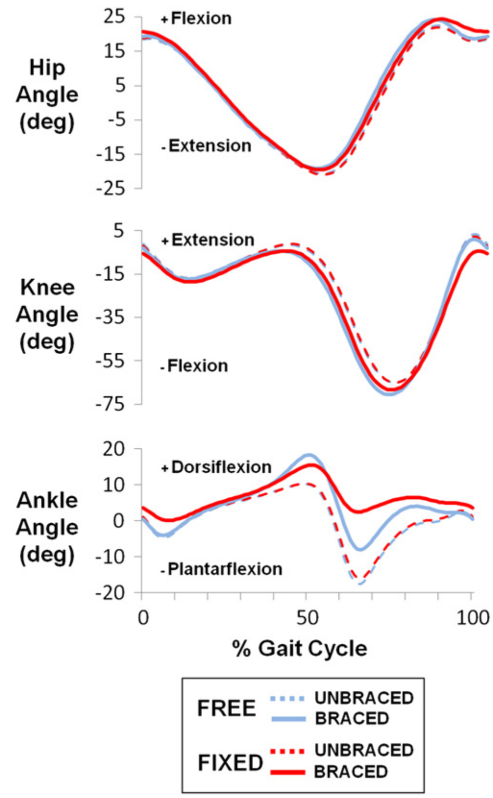

Fig. 2. Ensemble average of hip (top), knee (middle), and ankle (bottom) joint angles (degrees) from 0 (heel contact) to 100% (heel contact of same leg) of gait cycle of braced (solid) and unbraced (dashed) limbs during FREE (blue) and FIXED (red) conditions for all subjects. For all joints, zero degrees is neutral (standing) posture. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

No relationship was observed between the change in the braced limb's peak ankle or hip power generation during terminal stance between conditions and the change in the mass-specific net metabolic power between conditions (ankle: $r = -0.564$; $p = 0.056$; hip: $r = 0.152$; $p = 0.636$).

Average Power: The total average positive power for both limbs was $1.15 + 0.12$ W/kg during the FREE condition and 1.11 ± 0.16 W/kg during the FIXED condition (Table 1). Distribution of positive power at the ankle, knee, and hip for each limb while walking in the FREE and FIXED conditions can be found in Table 1 and Fig. 5.

A significant interaction effect (condition x limb) was found for the average positive power generation at the ankle ($p < 0.001$). Post hoc analysis demonstrated average positive power generated at the braced ankle in the FIXED condition was significantly less than during the FREE condition ($p < 0.001$) as well as less than the unbraced ankle during the FIXED condition ($p < 0.001$; Table 1). Interestingly, the unbraced ankle also significantly decreased the average mechanical power generation in the FIXED condition compared to the FREE condition ($p = 0.002$). At the knee joint, no significant main effects or interaction effect were observed.

A main effect for condition was found for average positive hip power generation ($p = 0.005$). Average positive power generated from the hip of the braced limb during the FIXED condition was significantly greater than during the FREE condition ($p = 0.007$), but not greater than the unbraced limb during the FIXED condition. The hip of the unbraced limb also generated significantly greater average positive mechanical power during the FIXED condition compared to the FREE condition ($p < 0.004$).

Change in average positive power generated by the ankles (braced + unbraced) was not significantly correlated with change

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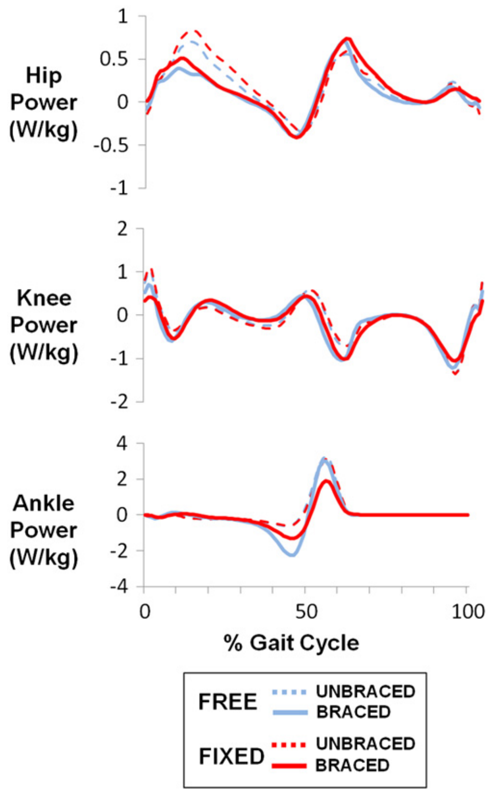

Fig. 3. Ensemble average of hip (top), knee (middle), and ankle (bottom) mechanical power (W/kg) from 0 (heel contact) to 100% (heel contact of same leg) of gait cycle of braced (solid lines) and unbraced (dashed lines) limbs during FREE (blue) and FIXED (red) conditions of all subjects. Positive values signify power generation, negative values signify power absorption. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

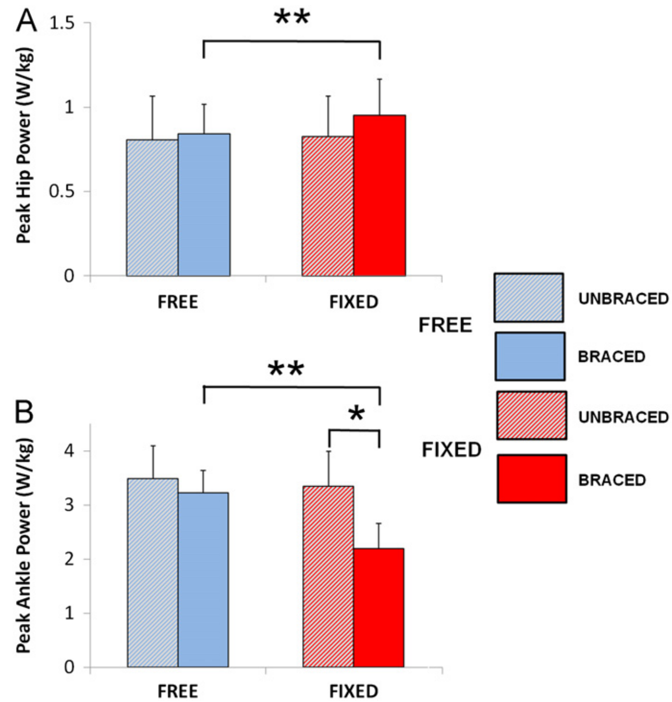

Fig. 4. Influence of fixed ankle walking on peak ankle (A) and peak hip (B) mechanical power generation (W/kg) during terminal stance phase for all subjects during FREE (blue) and FIXED (red) walking conditions. Error bars represent standard deviations. * indicates significant difference between braced and unbraced limb; ** indicates significant difference between FIXED and FREE conditions, $p<0.05$. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

Table 1
Average postive power generation (W/kg) at ankle, knee, and hip for braced and unbraced limbs in FREE and FIXED conditions.

		Hip (W/kg)	Knee (W/kg)	Ankle (W/kg)	Total (W/kg)
Braced	Free	0.19 ± 0.07^{a}	0.11 ± 0.04	0.24 ± 0.03^{a}	$0.53\pm0.13^{a,b}$
	Fixed	0.22 ± 0.09^{a}	0.10 ± 0.04	$0.15\pm0.04^{a,b}$	$0.47\pm0.16^{a,b}$
Unbraced	Free	0.23 ± 0.06^{a}	0.12 ± 0.03	0.27 ± 0.04^{a}	0.62 ± 0.14^{b}
	Fixed	0.26 ± 0.07^{a}	0.12 ± 0.04	$0.25\pm0.04^{a,b}$	0.63 ± 0.15^{b}
Total	Free	0.42 ± 0.10^{c}	0.22 ± 0.06	0.51 ± 0.06^{c}	1.15 ± 0.12
	Fixed	0.49 ± 0.14^{c}	0.22 ± 0.07	0.40 ± 0.07^{c}	1.11 ± 0.16

[a] Indicates significant difference ($p<0.05$) between FREE and FIXED conditions.
[b] Indicates significant difference ($p<0.05$) between braced and unbraced limbs.
[c] Indicates significant ($p<0.05$) difference between total positive power (braced+unbraced limbs) in FREE and FIXED conditions.

in mass specific metabolic power between conditions ($r=-0.325$, $p=0.302$). A significant relationship was found between the change in average positive power generation of the hips (unbraced+braced) and the change in mass specific metabolic power ($r=0.582$, $p=0.047$, Fig. 6).

4. Discussion

Our hypotheses that walking with unilaterally reduced ankle power generation (i.e., FIXED condition) would elicit a redistribution of mechanical power generation (to both the ipsilateral and contralateral hips) and require greater metabolic demand to walk at a constant velocity were supported. The hypothesis that mechanical power generation would increase at the contralateral ankle, however, was not supported. Results of this study have implications for understanding the role of focal ankle muscle weakness and/or orthosis use, and may provide insight into compensations that can develop following injury and/or disease.

Subjects responded to the reduction in the braced ankle's mechanical power output by increasing average mechanical power at both the ipsilateral and contralateral hips (Fig. 5). This redistribution of mechanical power is consistent with other reports documenting a power shift between hip and ankle joints during walking (Allen et al., 2011; Cruz et al., 2009; Lewis and Ferris, 2008). Unexpectedly, however, the unbraced ankle reduced its contribution by ~7% during FIXED ankle walking.

Previous attempts to understand the role of ankle push-off on energy expenditure tested people with unilateral arthrodesis compared to unimpaired subjects (Waters et al., 1988). Those authors noted a non-significant 3% increase in energy expenditure in the individuals with arthrodesis, however, walking velocity was not controlled and no mechanical analyses were performed. We have extended that previous work by controlling walking velocity (1.2 m/s), and unilaterally locking one ankle with an ankle brace. Locking the ankle in the FIXED condition served to model plantarflexor weakness and/or use of an AFO to restrict excessive plantarflexion. Modeling such constraints in a healthy population provides the opportunity to mechanistically study the consequences (e.g. mechanical, metabolic or stability/maneuverability) of reduced ankle power generation using a controlled, within-subject comparison between conditions. As anticipated, the imposed reduction of ankle power generation in our subjects increased the metabolic demand of walking (by ~7.4%) at the same velocity.

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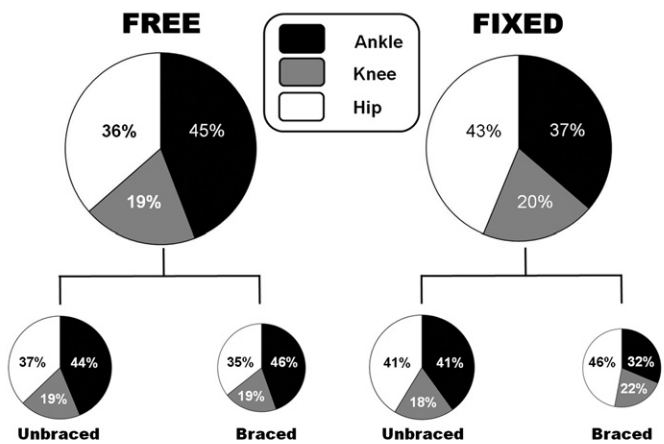

Fig. 5. Average percent contribution of positive power generation (W/kg) for all subjects during FREE (left) and FIXED (right) walking conditions at the ankle (black), knee (gray) and hip (white). The total positive power from both limbs (top pies) are further divided into unbraced (left) and braced (right) limbs for each condition. Size of pie is relative to unbraced limb during FREE condition.

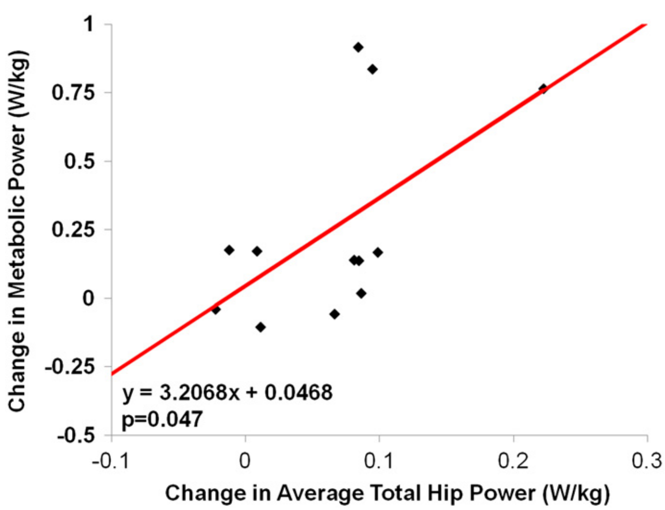

Fig. 6. Correlation between change in average total (unbraced + braced) hip power generation (W/kg; FIXED–FREE) and change in mass specific net metabolic power (W/kg; FIXED–FREE).

Interestingly, although the overall mechanical demand of the task in FREE (1.15 W/kg) vs. FIXED (1.11 W/kg) walking was not significantly different, the redistribution *within* and *across* the limbs required more metabolic energy to move at an identical speed. Importantly, the increase in average positive power generation at the hips from the FREE to the FIXED condition was directly related to the change in metabolic power. It is possible that reduced efficiency to generate mechanical power at the hip compared to the ankle (Farris and Sawicki, 2012b; Sawicki et al., 2009) contributed to the increase in metabolic cost during the FIXED condition.

Although the redistribution of mechanical power was correlated with the change in metabolic power, estimates of joint efficiency are not included in these predictions. In fact, our observed increase in net metabolic power of 7.4% in the FIXED condition compared to the FREE condition supports the contention that differences in elastic energy return between the more distal (ankle) and proximal (hip) muscles may contribute to increased metabolic demands (Kuo, 2002; Sawicki et al., 2009). Increased elastic energy return (i.e., efficiency) at the ankle are thought to be a result of the storage and return of energy by the Achilles tendon (Ishikawa et al., 2005; Sawicki et al., 2009). In particular, the catapulting action by tendons of the ankle muscles may be advantageous to return elastic energy that had been stored through mid-stance, adding to the power generated by plantarflexor muscles during terminal stance (Farris and Sawicki, 2012a,b; Ishikawa et al., 2005). While the elastic properties of the Achilles tendon allow some energy to be recycled with each step, the anatomy of the hip, with long muscle fascicles and limited tendon lengths, may reduce efficiency of power generation at the hip (Biewener and Roberts, 2000; Sawicki et al., 2009; Wickiewicz et al., 1983).

Using joint-level efficiency estimates ($\sim$0.61 for ankle versus $\sim$0.25 for hip and knee) we can convert the observed changes in the distribution of mechanical power generation to a prediction for percent change in metabolic cost (Sawicki et al., 2009). The observed $\sim$11% increase in hip contribution and $\sim$14% decrease in ankle contribution to total power output in the braced limb (see Fig. 5) would yield a $\sim$6.5% increase in net metabolic power (assuming, as observed, no difference in total power output between FREE and FIXED walking conditions and the distributions across joints documented in Fig. 5).

Others have examined the effect of restricted ankle motion bilaterally and noted no change in the energetic cost of walking (Vanderpool et al., 2008). A key notable difference, however, is the use of a curved rocker bottom on the braces used by Vanderpool and colleagues, which likely reduced collisional losses during step-to-step transitions (Adamczyk and Kuo, 2009). If the curved rocker bottom limited the need for ankle step-to-step transition work, then the presumed inefficiency of proximal muscles becomes relatively less critical as less positive power generation is necessary to maintain walking velocity. The flat bottom sole of our brace replicated the bottom of a traditional shoe, and therefore did not reduce collisional losses to provide the same metabolic benefit as the curved rocker bottom used by Vanderpool and colleagues. Instead, the loss of ankle power generation by our brace required a mechanical compensation by the hip joints, ultimately resulting in an increase in metabolic cost.

Mechanical lower-extremity joint power generation is developed through a combination of moment (force) and angular velocity at each joint. Therefore, impairments in force and/or angular velocity may contribute to reduced joint power generation, while increased force and/or angular velocity can increase joint power generation to serve as a compensation or recovery strategy. In our study, the redistribution of positive power generation from the ankle to the hip suggests that strengthening muscles (i.e., increasing muscle force) at proximal joints may be an effective strategy to increase hip mechanical power for individuals with plantarflexion power deficits who are incapable of increasing plantarflexion power (e.g., individual with paresis of ankle plantarflexors or those who wear a restrictive AFO). Strengthening compensatory muscles (e.g., hip flexors and extensors), however, may not fully resolve gait deficits created by a reduction in ankle plantarflexion power generation. Therefore strengthening ankle plantarflexors or providing passive or active assistance from a wearable ankle robot may be critical (Wiggin et al., 2011).

A potential limitation of our study is that the ankle brace had a fixed length footplate and heel height that could not be adjusted to match the subjects' foot dimensions. The height and length of the footplate appeared to alter the movement pattern from normal walking without the brace, which is evident in the between limb differences in kinematics and kinetics during the FREE condition (Figs. 2 and 3). Despite this issue, the within-subjects design of this study allows us to examine the compensations and metabolic demands of walking with reduced ankle power generation.

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In conclusion, by using a brace that limited ankle motion unilaterally we were able to reduce ankle power generation on the braced limb and observe a redistribution of lower-limb joint mechanical power to both hips in unimpaired young adults. This redistribution of power generation to more proximal joints coincided with an increase in the metabolic power required to maintain walking speed. Further study is necessary to determine if the efficiency of different joints (i.e. ankles vs. hip) is responsible for the observed relationship between joint power redistribution and metabolic demand.

Conflict of interest statement

There is nothing to declare with respect to conflict of interest.

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