

## Mechanics and energetics of level walking with powered ankle exoskeletons

Gregory S. Sawicki<sup>1,2,3\*</sup> and Daniel P. Ferris<sup>1,2,4,5</sup>

<sup>1</sup>Human Neuromechanics Laboratory, Departments of <sup>2</sup>Movement Science, <sup>3</sup>Mechanical Engineering, <sup>4</sup>Biomedical Engineering and <sup>5</sup>Physical Medicine and Rehabilitation, University of Michigan-Ann Arbor, Ann Arbor, MI 48109, USA

\*Author for correspondence (e-mail: gsawicki@umich.edu)

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### SUMMARY

**Robotic lower limb exoskeletons that can alter joint mechanical power output are novel tools for studying the relationship between the mechanics and energetics of human locomotion. We built pneumatically powered ankle exoskeletons controlled by the user's own soleus electromyography (i.e. proportional myoelectric control) to determine whether mechanical assistance at the ankle joint could reduce the metabolic cost of level, steady-speed human walking. We hypothesized that subjects would reduce their net metabolic power in proportion to the average positive mechanical power delivered by the bilateral ankle exoskeletons. Nine healthy individuals completed three 30 min sessions walking at 1.25 m s<sup>-1</sup> while wearing the exoskeletons. Over the three sessions, subjects' net metabolic energy expenditure during powered walking progressed from +7% to -10% of that during unpowered walking. With practice, subjects significantly reduced soleus muscle activity (by ~28% root mean square EMG,  $P < 0.0001$ ) and negative exoskeleton mechanical power (-0.09 W kg<sup>-1</sup> at the beginning of session 1 and -0.03 W kg<sup>-1</sup> at the end of session 3;  $P = 0.005$ ). Ankle joint kinematics returned to similar patterns to those observed during unpowered walking. At the end of the third session, the powered exoskeletons delivered ~63% of the average ankle joint positive mechanical power and ~22% of the total positive mechanical power generated by all of the joints summed (ankle, knee and hip) during unpowered walking. Decreases in total joint positive mechanical power due to powered ankle assistance (~22%) were not proportional to reductions in net metabolic power (~10%). The 'apparent efficiency' of the ankle joint muscle-tendon system during human walking (~0.61) was much greater than reported values of the 'muscular efficiency' of positive mechanical work for human muscle (~0.10–0.34). High ankle joint 'apparent efficiency' suggests that recoiling Achilles' tendon contributes a significant amount of ankle joint positive power during the push-off phase of walking in humans.**

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### INTRODUCTION

Humans are adept at harnessing the passive dynamics of their lower limbs to save energy during each walking step (Alexander, 1991; Cavagna et al., 2002; Kuo et al., 2005). During the single support phase, the center of mass trajectory approximates that of an energy conservative inverted pendulum (Alexander, 1995; Cavagna and Margaria, 1966; Kuo et al., 2005). As the center of mass moves upward then downward along a curved arc, gravitational potential energy and kinetic energy cycle nearly out of phase so that, in theory, zero mechanical work is required to sustain motion (Cavagna et al., 1976). The swing leg also behaves like a pendulum and will oscillate freely near its natural frequency with very little energy input (Mochon and McMahon, 1980). If swing and stance leg dynamics are matched so that they share a common cycle period, very little mechanical energy is required over a step. However, despite the available energy-saving pendular mechanisms, walking still requires a significant amount of metabolic energy (Kuo et al., 2005).

Walking like an inverted pendulum has energetic consequences. First, pendular exchange during single support is not purely passive. Computer simulations of walking indicate that considerable muscular work is required to drive the 'exchange' of kinetic and potential energy, particularly during the first half of single support (Neptune et al., 2004). Second, pendular exchange can only occur within a single step. At the end of each step, the leading leg collides into the ground, negative work is performed on the center of mass

and energy is lost. In order to maintain steady walking (i.e. zero net work on the center of mass over a stride), the lost energy must be exactly replaced. Positive mechanical work is required to redirect the velocity of the center of mass from the downward portion of one inverted pendulum to the upward portion of the next. Donelan et al. examined the mechanics of the step-to-step transition focusing on the individual limbs during double support. They found that while the leading limb performs negative work during the collision, the trailing limb performs simultaneous positive work to restore most of the energy of the center of mass (Donelan et al., 2002b). For walking at 1.25 m s<sup>-1</sup>, ~70% of the positive work performed on the center of mass occurs during double support (15.4 J of 21.7 J total). Furthermore, the mechanical work performed during double support increases with increasing step length and exacts a proportional metabolic cost (Donelan et al., 2002a). The combined results of these studies and others indicate that mechanical work performed during step-to-step transitions is a major determinant of the metabolic cost during level walking (Donelan et al., 2002a; Donelan et al., 2002b; Gottschall and Kram, 2003; Grabowski et al., 2005; Kuo et al., 2005).

One considerable drawback to studies employing center of mass level mechanical analyses is that they cannot directly address the relative roles of the lower limb joints in generating mechanical power during walking. Although it is clear that a substantial amount of work is done by the trailing limb during double support, our

understanding of how the ankle, knee and hip joints generate that energy is limited. Inverse dynamics can be used to get at possible sources of the push-off power burst by partitioning joint work into contributions from the hip, knee and ankle (Winter, 1990). Few studies have quantified joint work directly, but estimates from single limb joint power curves over a full walking stride at  $1.6 \text{ m s}^{-1}$  suggest that the ankle (~38%) and hip (~50%) combine to generate the majority of the positive work summed over the lower limb joints (Eng and Winter, 1995). However, when viewing only the push-off period of double support, it is evident that the ankle joint contributes more power than either the hip or knee (Kuo et al., 2005; Winter, 1991). Without direct *in vivo* measurements of triceps surae–Achilles' tendon dynamics and other ankle joint plantar flexors (e.g. tibialis posterior, peroneus longus), it is difficult to assess whether the majority of ankle joint push-off power is generated by positive work performed by actively shortening muscle or by passively recoiling tendon.

The Achilles' tendon may recycle a significant amount of elastic energy to help power the push-off phase of human walking. Recent advances in ultrasonography have facilitated examination of muscle–tendon interaction dynamics during walking (Fukunaga et al., 2001; Ishikawa et al., 2005; Lichtwark and Wilson, 2006). Results between studies are consistent and indicate that both soleus and gastrocnemius muscles perform some but not all of the ankle joint positive work during push-off. Furthermore, the Achilles' tendon undergoes a substantial amount of strain and recoils in a 'catapult action', allowing muscles to remain nearly isometric, at an operating point favoring economical force production.

Powered lower limb exoskeletons offer a novel means to alter the mechanics of walking at the level of the joints (rather than the center of mass) and study the human physiological response. Recently, Gordon and Ferris used a unilateral powered lower limb orthosis to study motor adaptation during walking (Gordon and Ferris, 2007). The results showed that humans can rapidly learn to walk with ankle joint mechanical assistance controlled by their own soleus muscle (i.e. under proportional myoelectric control). Over two 30 min powered walking practice sessions, individuals altered their soleus muscle activation to command distinct bursts of exoskeleton power focused at the push-off phase of walking. Although these results suggest that the human nervous system can selectively alter muscle activation patterns to produce efficient exoskeleton mechanics, measurements of users' metabolic energy expenditure were not taken to assess changes in metabolic cost.

The purpose of the present study was to quantify the metabolic cost of ankle joint work during level walking. We used bilateral powered exoskeletons to alter joint level mechanics in order to answer two questions. (1) How much can powered plantar flexion assistance during push-off reduce the metabolic cost of walking? (2) What is the 'apparent efficiency' of ankle joint work? Classical work from steep uphill walking indicates that human muscles perform positive mechanical work with a 'muscular efficiency' ( $\eta_{\text{muscle}}^+$ ) that asymptotically approaches ~0.25 (i.e. 1 J positive mechanical energy consumes ~4 J metabolic energy) (Margaria, 1968; Margaria, 1976). We assumed that positive mechanical work delivered by powered exoskeleton artificial muscles would directly replace biological ankle extensor positive muscle work. Thus, we hypothesized that for every 1 J of positive work the exoskeletons delivered, the user would save 4 J of metabolic energy. Stated differently, we hypothesized that ankle joint work is performed with an 'apparent efficiency' ( $\eta_{\text{ankle}}^+$ ) equal to ~0.25, reflecting underlying positive work performed by ankle extensor muscles. Further, we expected that subjects' net metabolic power would be

reduced in proportion to the relative contribution of exoskeleton positive work to the summed positive joint work (ankle + knee + hip) over a stride. We also expected reduced muscle activation amplitudes in the triceps surae group during powered walking. To test these ideas we compared subjects' net metabolic power and electromyography (EMG) amplitudes with exoskeletons powered *versus* unpowered during level, steady-speed walking. In addition, for powered walking we used measurements of artificial muscle forces and moment arm lengths to compute the average mechanical power delivered by the exoskeletons over a stride. With simultaneous measurements of the mechanics and energetics of powered walking we computed the 'apparent efficiency' of ankle joint positive work to gain insight into the underlying ankle extensor muscle–tendon function. Studying the relationship between mechanics and energetics at the level of the joints is an important step in integrating results from isolated muscle experiments with whole-body locomotion.

## MATERIALS AND METHODS

### Subjects

We recruited nine (4 males, 5 females) healthy subjects (body mass,  $77.8 \pm 12.4 \text{ kg}$ ; height,  $179 \pm 9 \text{ cm}$ ; leg length,  $93 \pm 5 \text{ cm}$ ; means  $\pm$  s.d.) who exhibited no gait abnormalities and had not previously walked with powered exoskeletons. Each participant read and signed a consent form prepared according to the Declaration of Helsinki and approved by the University of Michigan Institutional Review Board for human subject research.

### Exoskeletons

We constructed bilateral, custom-fitted ankle–foot orthoses (i.e. exoskeletons) for each subject (Fig. 1). Details of the design and performance of the exoskeletons are documented elsewhere (Ferris et al., 2005; Ferris et al., 2006; Gordon et al., 2006). Briefly, the lightweight exoskeletons (mass,  $1.21 \pm 0.12 \text{ kg}$  each) consisted of a polypropylene foot section attached to a carbon fiber shank with a hinge joint that allowed free motion about the ankle flexion–extension axis of rotation. We attached artificial pneumatic muscles (length,  $46.0 \pm 1.7 \text{ cm}$ ) along the posterior shank between two stainless steel brackets (moment arm,  $10.4 \pm 1.2 \text{ cm}$ ) to provide plantar flexor torque. A physiologically inspired controller incorporated the user's own soleus EMG to dictate the timing and amplitude of mechanical assistance (i.e. proportional myoelectric control; see supplementary material movie 1) (Gordon and Ferris, 2007).

### Protocol

Subjects practiced walking with bilateral ankle exoskeletons on a motorized treadmill set to  $1.25 \text{ m s}^{-1}$  during three separate practice sessions (session 1, session 2, session 3; Fig. 1). Our previous work with a unilateral exoskeleton showed that changes in kinematics and EMG reached steady state after two 30 min practice sessions. We chose three sessions based on pilot studies that indicated no further reduction in net metabolic power during powered walking with an additional practice during a fourth session ( $N=3$ ; ANOVA,  $P=0.97$ ). Thus, we considered data from the end of the third practice session to be representative of adapted powered walking. The practice sessions were separated by 3–5 days to allow for motor consolidation (Gordon and Ferris, 2007; Shadmehr and Holcomb, 1997). Each session followed the same walking time frame (Fig. 1).

At the start of the session subjects walked for 10 min wearing bilateral ankle exoskeletons unpowered (unpowered). Subjects then completed 30 min of walking with the exoskeletons powered

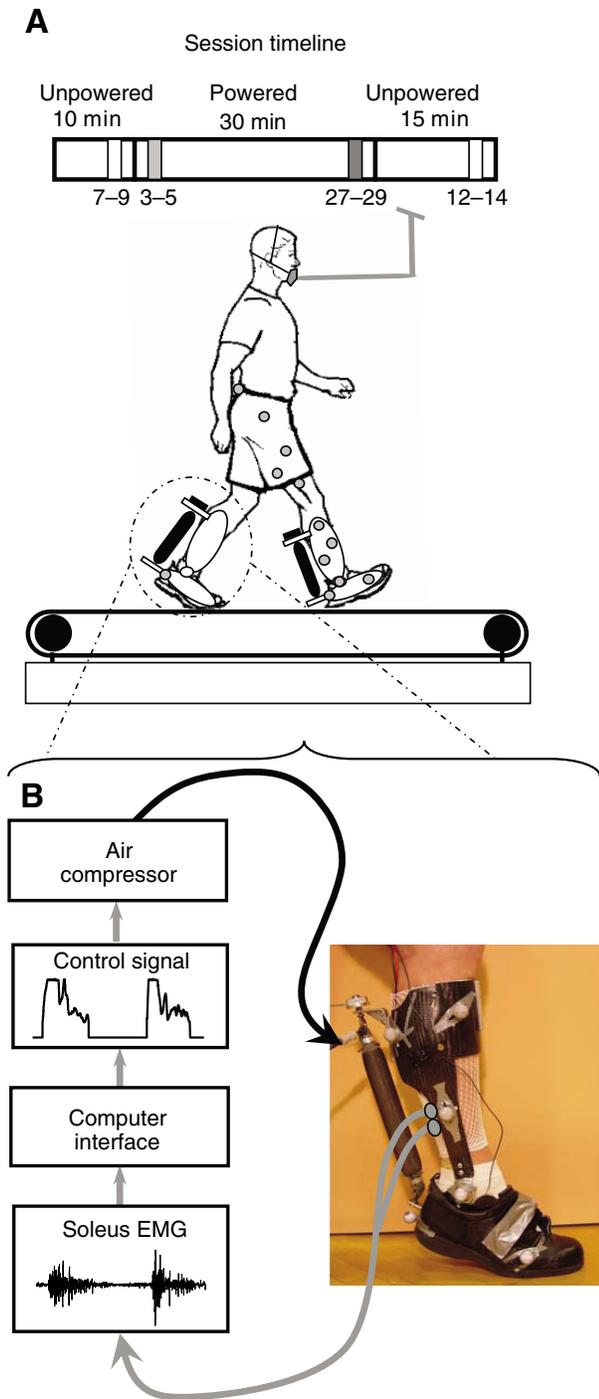


Fig. 1. Experimental set-up. (A) Subjects completed three practice sessions over a 7 day period. In each session, subjects walked on a motorized treadmill for 10 min with exoskeletons unpowered, 30 min with exoskeletons powered, and 15 min with exoskeletons unpowered. Outlined boxes indicate periods during which data were analyzed: unpowered beginning (minutes 7–9), powered beginning (minutes 3–5), powered end (minutes 27–29) and unpowered end (minutes 12–14). (B) During powered walking, bilateral ankle–foot orthoses (i.e. exoskeletons) drove ankle extension with artificial pneumatic muscles controlled using the subject’s own soleus surface electromyography (EMG; i.e. under proportional myoelectric control). We collected joint kinematics using reflective markers and motion capture,  $O_2$  and  $CO_2$  flow rates using open-circuit spirometry, and artificial muscle forces using compression force transducers.

(powered). Finally, subjects walked for 15 min with exoskeletons unpowered (unpowered). Subjects chose their preferred step length, step width and step frequency throughout. We tuned the gain and threshold of the proportional myoelectric controller during the initial unpowered walking bout so that background noise was eliminated and the control signal saturated for at least five consecutive steps. We then doubled the gain to encourage a reduction in soleus muscle recruitment (Gordon and Ferris, 2007).

We collected 10 s trials ( $\sim 7$  full strides) of kinematic, EMG and artificial muscle force data at the beginning of each minute during each practice session. Metabolic data were collected continuously. For analysis, we averaged data from minutes 7–9 of the first unpowered bout (unpowered beginning), minutes 3–5 (powered beginning) and 27–29 (powered end) of the powered bout, and minutes 12–14 of the second unpowered bout (unpowered end).

#### Metabolic cost

We used an open-circuit spirometry system (Physiodyne Instruments, Quogue, NY, USA) to record  $O_2$  and  $CO_2$  flow rates (Blaxter, 1989; Brooks et al., 1996). We converted averaged flow rates for each of the 2 min analysis intervals to units of metabolic power (watts) using the standard equations documented by Brockway (Brockway, 1987). To obtain the net metabolic power we averaged data from minutes 4–6 of a 7 min quiet standing trial and subtracted this from the gross metabolic power (Griffin et al., 2003). Net metabolic power values were then divided by subject mass. Throughout each session, care was taken to monitor the respiratory exchange ratio (RER) and ensure that subjects stayed in their aerobic range ( $RER < 1$ ) (Brooks et al., 1996). We used the net metabolic power from the unpowered beginning interval to compute percentage differences between unpowered and powered walking during each session.

#### Kinematics

We used an 8-camera video system (frame rate 120 Hz; Motion Analysis Corporation, Santa Rosa, CA, USA) and placed 29 reflective markers on the subject’s pelvis and lower limbs and recorded their positions during treadmill walking. We used custom-made software (Visual 3D, C-Motion, Rockville, MD, USA) to apply a 4th order Butterworth low-pass filter (cutoff frequency 6 Hz) and smooth raw marker data. Using the smoothed marker data, we calculated joint angles (relative to neutral standing posture) and angular velocities (ankle, knee, hip) for both legs. We marked heel-strike and toe-off events using footswitches (1200 Hz; B & L Engineering, Tustin, CA, USA) and calculated the step period (time from heel-strike of one leg to heel-strike of the other leg) and double support period (time from heel-strike of one leg to toe-off of the other). To calculate step length and step width we computed the fore–aft and lateral distances between calcaneus markers at heel-strike. Joint angles for the right and left legs were averaged from heel-strike (0%) to heel-strike (100%) to get the stride cycle average joint kinematics profiles.

#### Joint mechanics

To establish baseline joint mechanical power output we collected seven overground trials at  $1.25 \text{ m s}^{-1}$  for each subject walking with unpowered exoskeletons. To ensure that trials were within  $\pm 0.05 \text{ m s}^{-1}$  of the target speed, we used infrared timers triggered at the beginning and end of the walkway. We used two force platforms (sampling rate 1200 Hz; Advanced Mechanical Technology Inc., Watertown, MA, USA) to record ground reaction forces under each foot (left then right). Combining force platform

and marker data, we used inverse dynamics to calculate ankle, knee and hip mechanical power over the stride for each leg (Visual3D software). We used standard regression equations to estimate subjects' anthropometry (Zatsiorsky and Seluyanov, 1983) and adjusted foot and shank parameters to account for added exoskeleton mass and inertia. We averaged joint powers for the right and left legs (from heel-strike to heel-strike for each leg) and divided by subject mass to get the stride cycle average exoskeleton mechanical power.

We quantified the average rate of joint positive and negative mechanical work over a step. For each joint, we integrated only the positive (or negative) portions of both the left and right mechanical power curves (from right heel-strike to left heel-strike to capture simultaneous trailing and leading limb joint powers), summed them, and divided the total by the average step period.

### Exoskeleton mechanics

We used single-axis compression load transducers (1200 Hz; Omega Engineering, Stamford, CT, USA) to record the forces produced by the artificial pneumatic muscles during powered walking. We measured the artificial muscle moment arm with the ankle joint in the neutral position during upright standing posture (moment arm,  $10.4 \pm 1.2$  cm). We multiplied measured moment arm lengths and smoothed artificial muscle force data (low-pass filtered, 4th order Butterworth, cutoff frequency 6 Hz) to compute the exoskeleton torque for each leg. To determine the mechanical power delivered by the exoskeletons we multiplied the torque and ankle joint angular velocity (from motion capture). We averaged the exoskeleton power for the right and left legs (from heel-strike to heel-strike for each leg) and divided by subject mass to get the stride cycle average exoskeleton mechanical power.

We quantified the average rate of exoskeleton positive and negative mechanical work over a stride for comparison with net metabolic power and baseline joint mechanics. We integrated only the positive (or negative) portions of both the left and right exoskeleton mechanical power curves (from left heel-strike to left heel-strike), summed them, and divided the total by the average stride period.

### Electromyography

We recorded bilateral lower limb surface EMG (1200 Hz; Konigsberg Instruments, Inc., Pasadena, CA, USA) from soleus, tibialis anterior, medial gastrocnemius and lateral gastrocnemius using bipolar electrodes (inter-electrode distance, 3.5 cm) centered over the belly of the muscle along its long axis. EMG amplifier bandwidth filter was 12.5–920 Hz. We placed electrodes to minimize cross-talk and taped them down to minimize movement artifact. We marked the locations of the electrodes on the skin so we could place them in the same position from session to session. We high-pass filtered (4th order Butterworth, cutoff frequency 20 Hz), rectified and low-pass filtered (4th order Butterworth, cutoff frequency 10 Hz) each of the EMG signals (i.e. linear envelope). We averaged the linear enveloped EMG for the right and left legs (from heel-strike to heel-strike for each leg) to get stride cycle averages. We normalized the curves using the peak value (average of left and right) for each muscle during the first unpowered walking bout (unpowered beginning) during each session.

To quantify changes in EMG amplitude, we computed stance phase root mean square (r.m.s.) average EMG amplitudes from the high-pass filtered, rectified EMG data of each leg. We averaged r.m.s. EMG values from each leg and normalized them using the average r.m.s. value from the unpowered beginning interval.

### Ankle joint 'apparent efficiency' via exoskeleton performance index

We combined mechanical and metabolic analyses to determine the exoskeleton performance index and ankle joint 'apparent efficiency' ( $\eta_{\text{ankle}}^+$ ). First, we calculated metabolic power savings due to the exoskeletons by subtracting the net metabolic power during the first unpowered walking interval in each session from the net metabolic power during each of the powered walking intervals in that session. Cycle ergometry studies have demonstrated that computing this difference provides a valid method of testing metabolic efficiency of the leg musculature (Gaesser and Brooks, 1975; Poole et al., 1992). It accounts for the fact that some metabolic cost during locomotion can be attributed to sources other than limb muscle energetics (e.g. breathing, circulation, digestion, etc.), resulting in whole-body metabolic power calculations that parallel direct lower limb metabolic power across different workloads [also discussed for loaded walking in the appendix of Griffin et al. (Griffin et al., 2003)]. Next, we assumed that changes in metabolic energy consumption would reflect the cost of the biological muscle positive work replaced by the powered exoskeletons. Classical work from steep uphill walking indicates that human muscles perform positive mechanical work with a 'muscular efficiency' ( $\eta_{\text{muscle}}^+$ ) that asymptotically approaches  $\sim 0.25$  (Margaria, 1968; Margaria, 1976). Thus, we multiplied changes in net metabolic power by  $\eta_{\text{muscle}}^+ = 0.25$  to yield the expected amount of positive mechanical power delivered by the exoskeletons. Then we divided the measured average positive mechanical power by the expected average positive mechanical power delivered by the exoskeletons to yield the exoskeleton performance index (Eqn 1):

$$\text{Exoskeleton performance index} = \frac{\Delta \text{ Net metabolic power} \times \eta_{\text{muscle}}^+}{\text{Average exoskeleton positive mechanical power}} \quad (1)$$

To compute the ankle joint 'apparent efficiency'  $\eta_{\text{ankle}}^+$  (Asmussen and Bonde-Petersen, 1974) we inverted the performance index and scaled it by  $\eta_{\text{muscle}}^+$  (Eqn 2). Therefore with  $\eta_{\text{muscle}}^+ = 0.25$ , performance index = 1 yields 'apparent efficiency' = 0.25:

$$\text{Ankle joint 'apparent efficiency'} = \frac{\text{Average exoskeleton positive mechanical power}}{\Delta \text{ Net metabolic power}} = \frac{\eta_{\text{muscle}}^+}{\text{Exoskeleton performance index}} \quad (2)$$

It is important to note that while the performance index depends directly on the assumed value for  $\eta_{\text{muscle}}^+$  the 'apparent efficiency' ( $\eta_{\text{ankle}}^+$ ) does not (see Eqns 1 and 2). We chose to calculate the performance index as an intermediate step in the 'apparent efficiency' computation for three reasons: (1) as change in net metabolic power approaches zero, 'apparent efficiency' asymptotically approaches infinity non-linearly (biasing means and complicating statistical analyses) whereas performance index approaches zero nearly linearly; (2) performance index may be more intuitive than the 'apparent efficiency' because it increases as reductions in metabolic cost increase (i.e. as performance improves); and (3) the performance index can give insight into the underlying mechanical function of the ankle muscle-tendon system. The performance index represents an upper bound on the

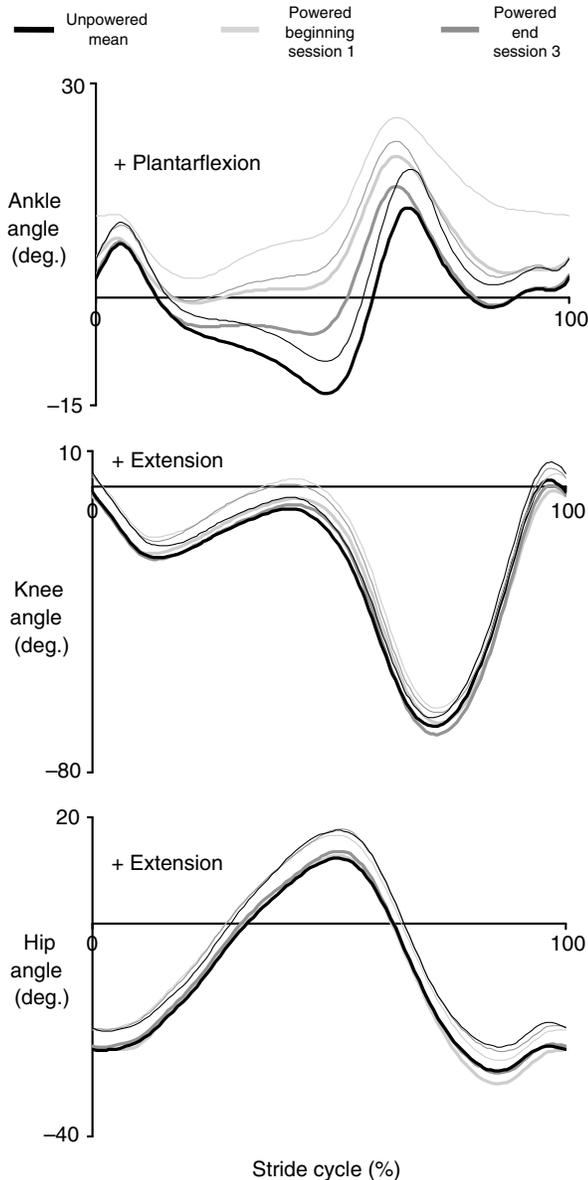


Fig. 2. Joint kinematics. Thick curves are nine subject mean ankle, knee and hip joint kinematics for unpowered walking from the beginning of practice session 3 (black) and powered walking at the beginning of practice session 1 (light gray) and end of practice session 3 (dark gray). Thin curves are +1 s.d. and match colors for means. Curves are the stride average of left and right legs and are plotted from heel-strike (0%) to heel-strike (100%). Stance is ~0% to 60% of the stride, swing 60% to 100%. For all joints, zero degrees is upright standing posture. Ankle plantarflexion, knee extension and hip extension are positive.

fraction of ankle joint positive mechanical work performed by active muscle shortening (*versus* elastic energy delivered by passive tendon recoil). For example, a performance index of 1.0 (i.e.  $\eta^+_{\text{ankle}} = \eta^+_{\text{muscle}}$ ) would indicate that all of the exoskeleton pneumatic muscle work replaced underlying biological muscle performing positive mechanical work with  $\eta^+_{\text{muscle}}$ . On the other hand, a performance index of 0.5 (i.e.  $\eta^+_{\text{ankle}} = 2\eta^+_{\text{muscle}}$ ) would indicate that the positive muscle work replaced by exoskeleton assistance comprised only 50% of the ankle joint positive mechanical work, while the remainder of the work could be

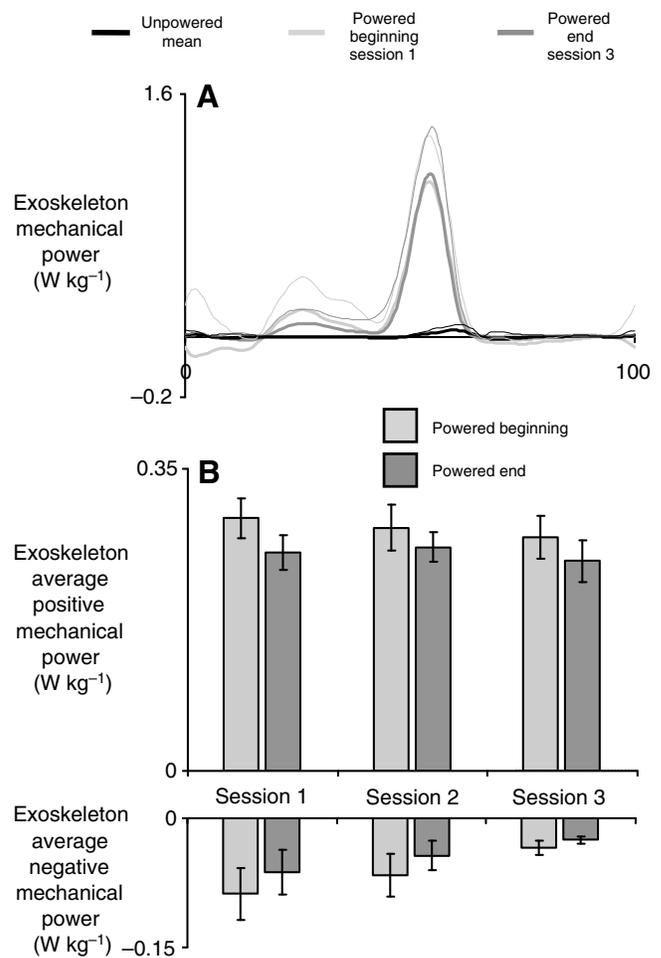


Fig. 3. Ankle exoskeleton mechanical power. (A) Nine subject mean (thick curves) + 1 s.d. (thin curves) of exoskeleton mechanical power delivered over the stride from heel-strike (0%) to heel-strike (100%; left and right exoskeletons are averaged for each subject). Curves are three session average for unpowered walking (black) and powered walking at the beginning of practice session 1 (light gray) and end of practice session 3 (dark gray). Mechanical power is computed as the product of exoskeleton torque and ankle joint angular velocity and is normalized by subject mass. Positive power indicates energy transferred to the user and negative power indicates energy absorbed from the user. (B) Bars show the nine subject mean exoskeleton average positive and negative mechanical power over a stride for powered walking. Error bars are  $\pm 1$  s.e.m. Practice sessions (1–3) are shown left to right with beginning period (minutes 3–5) in light gray and end period (minutes 27–29) in dark gray. All mechanical power values are normalized by subject mass.

attributed to elastic energy recoil (see Discussion). Note that the performance index, or estimated fraction of mechanical work performed by active muscle, can also be expressed as the ratio  $\eta^+_{\text{muscle}}/\eta^+_{\text{ankle}}$ .

**Statistical analyses**

We used JMP statistical software (SAS Institute Inc., Cary, NC, USA) to perform repeated measures ANOVA. When we found a significant effect ( $P < 0.05$ ) we used *post-hoc* Tukey honestly significant difference (THSD) tests to determine specific differences between means. Statistical power analyses were done for tests yielding significance ( $P < 0.05$ ).

In the first two analyses (one for powered walking data, one for unpowered walking data) we assessed the effects of practice session (session 1, session 2, session 3) and period (beginning, end) on net metabolic power, exoskeleton mechanics, stance phase r.m.s. EMG and gait kinematics metrics (two-way ANOVA: session and period). In the other three ANOVA analyses (one for session 1, session 2 and session 3) we assessed the effects of exoskeleton condition (unpowered, powered) and period (beginning, end) on net metabolic power, stance phase r.m.s. EMG and gait kinematics metrics (two-way ANOVA: condition and period).

## RESULTS

### Joint kinematics

During powered walking, subjects initially (powered beginning session 1) walked with increased ankle plantar flexion throughout the stride. By the end of the third practice session (powered end session 3), stance phase ankle joint kinematics returned closer to the unpowered condition, but push-off started earlier in stance and peak plantar flexion angle was larger (Fig. 2). With practice, powered swing phase ankle kinematics became similar to the pattern observed during unpowered walking.

Knee and hip joint kinematics were not altered by exoskeleton powering and there were no changes in unpowered ankle, knee or hip joint kinematics over the practice sessions (Fig. 2).

### Exoskeleton mechanics

Ankle exoskeletons produced passive torques near zero during unpowered walking. During the beginning of the first powered interval (powered beginning session 1), the exoskeletons produced plantar flexor torque over most of the stance phase. Exoskeletons also produced some extensor torque during the swing phase. With practice, exoskeleton torque became narrowly focused near the push-off phase of stance and was absent during swing. At the end of the third powered session (powered end session 3), peak exoskeleton torque reached  $\sim 0.47 \text{ Nm kg}^{-1}$  or  $\sim 37\%$  of the peak ankle joint moment from overground trials during unpowered walking.

Changes in exoskeleton torque were reflected in the mechanical power they delivered to the user's ankle joints. Because exoskeleton torque was initially spread over the stride there were periods of negative mechanical work done (i.e. energy absorption) by the mechanical assistance during early stance and in swing (Fig. 3). The exoskeletons absorbed  $-0.09 \pm 0.03 \text{ W kg}^{-1}$  (mean  $\pm$  s.e.m.) average negative mechanical power and delivered  $0.29 \pm 0.02 \text{ W kg}^{-1}$  average positive mechanical power over the stride at the beginning of the first powered session (powered beginning session 1). As torque became more focused near push-off, negative mechanical work done during both swing and early stance was reduced. By the end of the third powered session (powered end session 3), exoskeleton average negative mechanical power was  $\sim 70\%$  lower than during the initial powered interval (powered beginning session 1; ANOVA,  $P=0.005$ ;

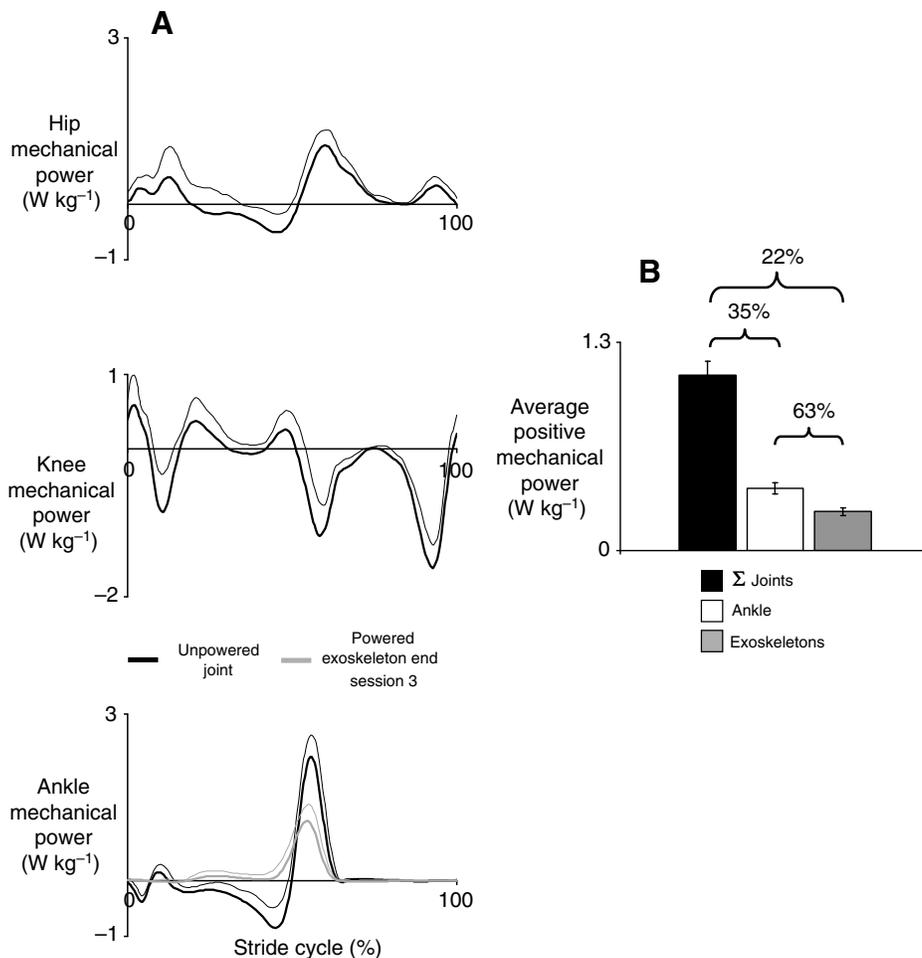


Fig. 4 Lower limb joint kinetics. (A) Nine subject mean (thick black curve) + 1 s.d. (thin black curve) mechanical power delivered by each of the lower limb joints over the stride from heel-strike (0%) to heel-strike (100%). Left and right legs are averaged for each subject. Curves are for unpowered walking overground at  $1.25 \text{ m s}^{-1}$ . The mean exoskeleton mechanical power from the end of practice session 3 (thick dark gray curve) + 1 s.d. (thin dark gray curve) is overlaid on the bottom subplot for the ankle joint mechanical power. Mechanical power is computed as the product of exoskeleton torque and ankle joint angular velocity and is normalized by subject mass. Positive power indicates energy transferred to the user and negative power indicates energy absorbed from the user. (B) Bars showing the nine subject mean positive mechanical power delivered by the sum of the ankle, knee and hip joints (black) and ankle joint (white) during unpowered walking and the exoskeletons (gray) during powered walking. Error bars are  $\pm 1$  s.e.m. All mechanical power values are normalized by subject mass. Braces indicate the percentage contribution of bars from right to left. For example, the exoskeleton average positive mechanical power was 63% of the ankle joint average positive mechanical power over the stride.

THSD, session 3 < session 1). Exoskeleton average positive mechanical power was not different across practice sessions (ANOVA,  $P=0.29$ ), but was significantly lower at the end of each session when compared with the beginning of each session (ANOVA,  $P=0.001$ ; THSD, end < beginning). At the end of the third practice session (powered end session 3), the exoskeletons delivered  $0.24 \pm 0.02 \text{ W kg}^{-1}$  average positive mechanical power over a stride. This was 63% of unpowered ankle joint average positive mechanical power and 22% of unpowered average positive joint mechanical power summed across the joints (ankle + knee + hip; Fig. 4).

**Metabolic cost**

As the exoskeletons absorbed less mechanical energy from the user, the net metabolic power during powered walking decreased to levels below that of unpowered walking. Initially, powered assistance increased net metabolic power by  $0.26 \pm 0.28 \text{ W kg}^{-1}$  (powered beginning session 1; Fig. 5A). This was ~7% higher than the net metabolic power during unpowered walking (unpowered beginning session 1). With practice, subjects reduced net metabolic power significantly both across (ANOVA,  $P=0.0001$ ; THSD, session 3 < session 2, session 3 < session 1) and within sessions (ANOVA,  $P=0.006$ ; THSD, end < beginning; Table 1). The net metabolic power at the beginning of the first powered session (powered beginning session 1) was  $3.84 \pm 0.30 \text{ W kg}^{-1}$  but was reduced by 22% (to  $2.99 \pm 0.17 \text{ W kg}^{-1}$ ) by the end of the third powered session (powered end session 3). Further, the net metabolic power was significantly lower (-10%) with exoskeletons powered ( $2.99 \pm 0.17 \text{ W kg}^{-1}$ ; powered end session 3) versus unpowered ( $3.31 \pm 0.11 \text{ W kg}^{-1}$ ; unpowered beginning session 3), by the end of the third practice session (ANOVA,  $P=0.03$ ; THSD, powered < unpowered; Fig. 5A, Table 1).

The metabolic cost of unpowered walking decreased across sessions (ANOVA,  $P=0.001$ ; THSD, session 2 < session 1, session 3 < session 1) but was not different within sessions (ANOVA,  $P=0.34$ ; Table 1). Unpowered net metabolic power was ~8% lower during session 3 when compared with session 1.

**Exoskeleton performance index and ankle joint ‘apparent efficiency’**

The metabolic benefit of powered ankle assistance increased with practice. Exoskeleton performance was significantly higher within practice sessions (ANOVA,  $P=0.004$ ; THSD, end > beginning) and followed an increasing trend across practice sessions (ANOVA,  $P=0.05$ ; Fig. 5B). Initially, powered assistance perturbed gait, net metabolic cost was elevated, and exoskeleton performance index was negative ( $-0.14 \pm 0.19$  during powered beginning session 1). By the end of session 3 (powered end session 3), exoskeleton average positive mechanical power ( $0.24 \pm 0.02 \text{ W kg}^{-1}$ ) reduced the net metabolic power by  $0.32 \pm 0.12 \text{ W kg}^{-1}$  and performance index was

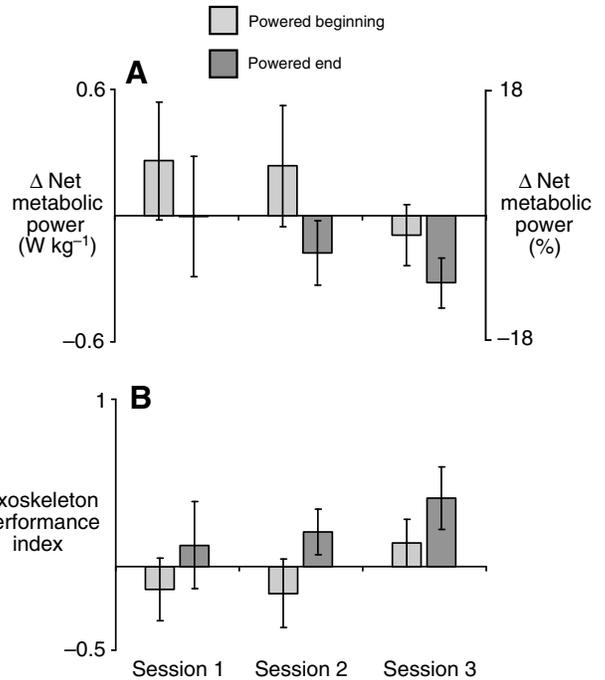


Fig. 5 Exoskeleton performance. (A) Bars showing the nine subject mean change in net metabolic power (powered – unpowered) due to powered assistance from bilateral exoskeletons. Error bars are ±1 s.e.m. All metabolic power values are normalized by subject mass. Right axis indicates the change in net metabolic power as a percentage difference from unpowered walking during each session. (B) Bars indicating nine subject mean ± 1 s.e.m. exoskeleton performance index (unitless). Performance index indicates the fraction of average exoskeleton positive mechanical power that results in a reduction in net metabolic power, assuming that artificial muscle work directly replaces biological muscle work. A performance index of 1.0 would indicate that all of the exoskeleton pneumatic muscle work replaced underlying biological muscle performing positive mechanical work with  $\eta^+_{\text{muscle}}$ . For both panels, practice sessions (1–3) are shown left to right with beginning period (minutes 3–5) in light gray and end period (minutes 27–29) in dark gray.

positive ( $0.41 \pm 0.19$ ; Fig. 5B). The ankle joint ‘apparent efficiency’ was 0.61 at the end of session 3.

**Electromyography**

Subjects immediately reduced their soleus muscle activation during powered walking and continued to do so with practice (Fig. 6, Table 2). By the end of the third practice session, stance phase soleus r.m.s. EMG amplitude was 28% lower in the powered (powered end session 3) versus unpowered (unpowered beginning session 3) condition (ANOVA,  $P<0.0001$ ; THSD, powered < unpowered).

Table 1. Metabolic cost of exoskeleton walking

Metric	Unpowered		Powered		Condition $P$ value; THSD
	Beginning	End	Beginning	End	
Session 1 Net metabolic power ( $\text{W kg}^{-1}$ )	$3.58 \pm 0.10$	$3.52 \pm 0.10$	$3.84 \pm 0.30$	$3.57 \pm 0.31$	$P=0.33$
Session 2 Net metabolic power ( $\text{W kg}^{-1}$ )	$3.40 \pm 0.10$	$3.39 \pm 0.08$	$3.64 \pm 0.33$	$3.23 \pm 0.18$	$P=0.81$
Session 3 Net metabolic power ( $\text{W kg}^{-1}$ )	$3.31 \pm 0.11$	$3.25 \pm 0.10$	$3.22 \pm 0.20$	$2.99 \pm 0.17$	$P=0.03^*$ Powered < Unpowered

Net metabolic power data ( $\text{W kg}^{-1}$ ) are means ± s.e.m.,  $N=9$ . See Materials and methods for calculations.

Statistics: THSD, Tukey honestly significant difference test results; Condition, Powered/Unpowered;  $P<0.05$  indicates statistical significance; \*statistical power >0.65.

Soleus r.m.s. was lower at the end when compared with the beginning of the powered interval during each practice session (ANOVA,  $P=0.01$  in session 1,  $P=0.007$  in session 2,  $P=0.004$  in session 3; THSD, all end < beginning).

Initially subjects increased activity in their tibialis anterior muscle throughout the stride providing a reaction torque in response to powered assistance. With practice, activity patterns returned to normal (Fig. 7, Table 2). During the beginning of powered walking (powered beginning session 1), tibialis anterior stance phase r.m.s. EMG was 52% higher than in the unpowered condition (unpowered beginning session 1; ANOVA,  $P=0.001$ ; THSD, powered > unpowered). During powered walking, tibialis anterior activity decreased both across (ANOVA,  $P=0.001$ ; THSD, session

3 < session 1) and within (ANOVA,  $P=0.001$ ; THSD, end < beginning) practice sessions (Fig. 7, Table 2). By the third session, there was no significant difference between unpowered and powered walking tibialis anterior r.m.s. amplitude (ANOVA,  $P=0.05$ ).

At the end of the third powered session (powered end session 3), lateral gastrocnemius r.m.s. EMG amplitude was ~10% lower than in unpowered walking (unpowered beginning session 3). Medial gastrocnemius amplitude was reduced as well, but only by ~4%. However, none of the observed reductions in stance phase r.m.s. EMG amplitude for medial or lateral gastrocnemius during powered walking were statistically significant (ANOVA,  $P=0.52$  for medial gastrocnemius and  $P=0.09$  for lateral gastrocnemius; Table 2).

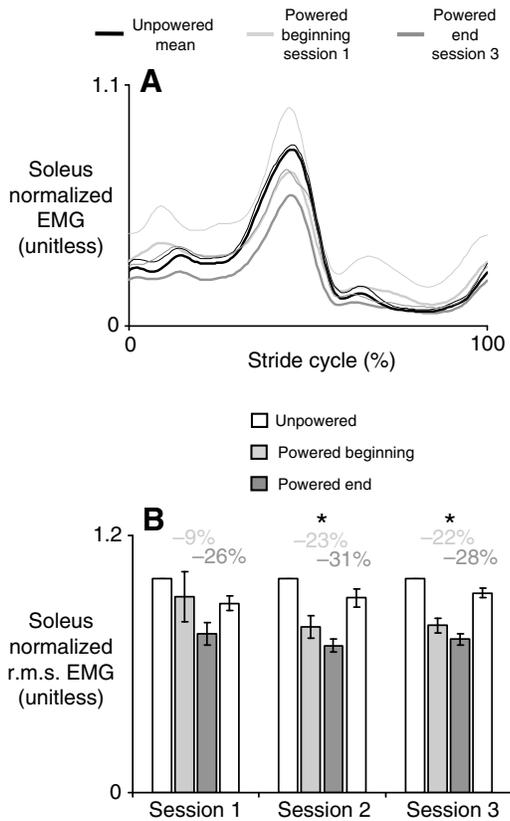


Fig. 6 Soleus EMG. (A) Nine subject mean (thick curves) + 1 s.d. (thin curves) of soleus normalized linear enveloped (high-pass cutoff frequency 20 Hz and low-pass cutoff frequency 10 Hz) muscle activity over the stride from heel-strike (0%) to heel-strike (100%). Left and right legs are averaged for each subject. Stance phase is ~0% to 60% and swing ~60% to 100% of the stride. Thick curves are three session average for unpowered walking (black) and powered walking at the beginning of practice session 1 (light gray) and the end of practice session 3 (dark gray). Thin curves are +1 s.d. and follow the same color scheme as means. Curves are normalized to the peak value during unpowered walking at the beginning of each session (unitless). (B) Bars showing the nine subject mean of stance phase r.m.s. average soleus muscle activation (unitless). Error bars are  $\pm 1$  s.e.m. Practice sessions (1–3) are shown left to right with unpowered walking periods (minutes 7–9 and minutes 12–14 at beginning and end, respectively) in white and powered beginning periods (minutes 3–5) and powered end periods (minutes 27–29) in dark gray. Percentages listed above bars for powered walking indicate the difference from unpowered beginning in each session. Asterisks indicate a statistically significant difference between powered and unpowered walking (ANOVA,  $P<0.05$ ).

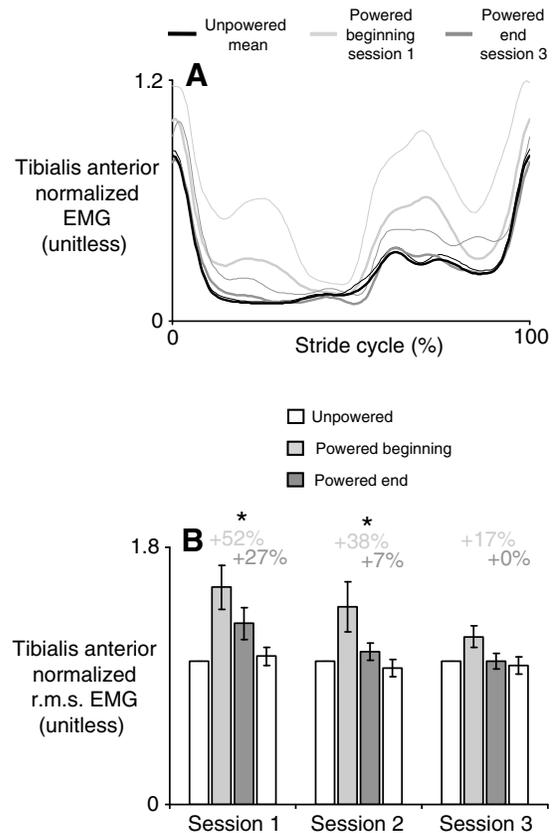


Fig. 7 Tibialis anterior EMG. (A) Nine subject mean (thick curves) + 1 s.d. (thin curves) of tibialis anterior normalized linear enveloped (high-pass cutoff frequency 20 Hz and low-pass cutoff frequency 10 Hz) muscle activity over the stride from heel-strike (0%) to heel-strike (100%). Left and right legs are averaged for each subject. Stance phase is ~0% to 60% and swing ~60% to 100% of the stride. Thick curves are three session average for unpowered walking (black) and powered walking at the beginning of practice session 1 (light gray) and the end of practice session 3 (dark gray). Thin curves are +1 s.d. and follow the same color scheme as means. Curves are normalized to the peak value during unpowered walking at the beginning of each session (unitless). (B) Bars showing the nine subject mean of stance phase r.m.s. average tibialis anterior muscle activation (unitless). Error bars are  $\pm 1$  s.e.m. Practice sessions (1–3) are shown left to right with unpowered walking periods (minutes 7–9 and minutes 12–14 at the beginning and end, respectively) in white and powered beginning periods (minutes 3–5) in light gray and powered end periods (minutes 27–29) in dark gray. Percentages listed above bars for powered walking indicate the difference from unpowered beginning in each session. Asterisks indicate a statistically significant difference between powered and unpowered walking (ANOVA,  $P<0.05$ ).

Table 2. Ankle joint muscle root mean square electromyography

	Metric	Unpowered		Powered		Condition <i>P</i> value; THSD
		Beginning	End	Beginning	End	
Session 1	Sol. r.m.s.	1.00±0.00	0.88±0.03	0.91±0.12	0.74±0.05	<i>P</i> =0.05
	TA r.m.s.	1.00±0.00	1.04±0.06	1.52±0.15	1.27±0.11	<i>P</i> =0.001** Powered > Unpowered
	MG r.m.s.	1.00±0.00	0.96±0.04	1.02±0.06	0.92±0.05	<i>P</i> =0.85
	LG r.m.s.	1.00±0.00	0.97±0.02	1.06±0.13	0.90±0.06	<i>P</i> =0.93
Session 2	Sol. r.m.s.	1.00±0.00	0.91±0.04	0.77±0.05	0.69±0.03	<i>P</i> <0.0001** Powered < Unpowered
	TA r.m.s.	1.00±0.00	0.96±0.06	1.38±0.18	1.07±0.06	<i>P</i> =0.02* Powered > Unpowered
	MG r.m.s.	1.00±0.00	0.93±0.06	0.93±0.04	0.94±0.06	<i>P</i> =0.45
	LG r.m.s.	1.00±0.00	0.95±0.03	1.00±0.10	0.82±0.06	<i>P</i> =0.21
Session 3	Sol. r.m.s.	1.00±0.00	0.93±0.03	0.78±0.04	0.72±0.03	<i>P</i> <0.0001** Powered < Unpowered
	TA r.m.s.	1.00±0.00	0.97±0.06	1.17±0.08	1.00±0.06	<i>P</i> =0.05
	MG r.m.s.	1.00±0.00	0.98±0.03	0.98±0.05	0.96±0.06	<i>P</i> =0.52
	LG r.m.s.	1.00±0.00	1.04±0.03	0.97±0.09	0.90±0.06	<i>P</i> =0.09

Values (means ± s.e.m.) are root mean square (r.m.s.) average from stance phase only normalized to unpowered beginning condition and are therefore unitless; *N*=9. See Materials and methods for calculations.

Statistics: THSD, Tukey honestly significant difference test results; Condition, Powered/Unpowered; *P*<0.05 indicates statistical significance; \*statistical power >0.65; \*\*statistical power >0.80.

Sol., soleus; TA, tibialis anterior; MG, medial gastrocnemius; LG, lateral gastrocnemius.

### Gait kinematics

Initially subjects took shorter and wider steps during powered *versus* unpowered walking. Step length was 724±9 mm during unpowered walking (unpowered beginning session 1) and 713±10 mm during powered walking (powered beginning session 1; ANOVA, *P*=0.006; THSD, powered < unpowered). At the beginning of session one, step width was 105±10 mm during unpowered walking and 127±8 mm during powered walking (ANOVA, *P*<0.0001; THSD, powered > unpowered). By the end of the third session, subjects' step width during powered walking (120±12 mm; powered end session 3) was not different from that during unpowered walking (123±11 mm; unpowered beginning session 3; ANOVA, *P*=0.05). In the third session, step length remained slightly shorter in powered (717±14 mm) *versus* unpowered (732±14 mm) walking (ANOVA, *P*=0.01; THSD, powered < unpowered). There were no significant changes in step period or double support period due to powered assistance.

### DISCUSSION

In this study we quantified the metabolic cost of ankle joint work during level, steady-speed walking. We used bilateral powered exoskeletons to alter joint level mechanics and answer two questions. (1) Does powered plantar flexion assistance during push-off reduce the metabolic cost of walking? (2) What is the 'apparent efficiency' of ankle joint work? Our results indicate that when powered ankle exoskeletons deliver 22% of the positive work generated by the joints (ankle + knee + hip), users reduce net metabolic power by ~10%. We determined that the 'apparent efficiency' of ankle joint work is 0.61; that is, for every 1 J of positive mechanical work delivered by ankle exoskeletons, users save ~1.6 J of metabolic energy.

We are aware of only one other study reporting oxygen consumption during walking with powered lower limb exoskeletons. Norris et al. built bilateral powered ankle foot orthoses with hardware based on our previous designs (Ferris et al., 2005; Ferris et al., 2006) but using an alternative control scheme based on ankle joint kinematics rather than soleus EMG (Norris et al., 2007). They examined the effects of augmented plantar flexion power on the economy and preferred walking speed in younger and older adults (Norris et al., 2007). They found that when young adults walked with powered assistance, gross metabolic energy per stride was ~8%

lower and preferred walking speed ~7% higher when compared with unpowered walking. Because they used a different type of exoskeleton controller (kinematic-based timing rather than proportional myoelectric control), did not keep speed constant in their comparisons, had subjects complete only a very short period of training (less than 20 min), and did not measure inverse dynamics of their subjects, it is difficult to make comparisons between their findings and ours.

Our results are consistent with previous studies from our own laboratory using a unilateral powered ankle exoskeleton under soleus proportional myoelectric control. Gordon and Ferris (Gordon and Ferris, 2007) found that within two 30 min practice sessions (~45 min cumulative powered walking), humans reduced soleus activation by ~35%, returned to near normal ankle joint kinematics, eliminated exoskeleton negative mechanical power generation, and delivered positive exoskeleton mechanical power focused at push-off (~0.13–0.15 W kg<sup>-1</sup> for 12–14 J). As expected, with practice, our bilateral exoskeletons delivered nearly twice the average positive mechanical power (0.24±0.02 W kg<sup>-1</sup>) when compared with the single unilateral exoskeleton in the study of Gordon and Ferris. We also observed similar changes in ankle joint kinematics, soleus EMG (~28% reduction) and exoskeleton average negative mechanical power (~70% reduction) over three training sessions. Gordon and Ferris quantified the time for key metrics (e.g. soleus r.m.s. amplitude, exoskeleton positive and negative work, and ankle joint angle correlation common variance) to reach steady values. For the metrics they studied, they observed no further changes after ~45 min of cumulative powered walking. In the current study, we did not assess the rate of motor adaptation during powered walking, but data on three subjects showed no further reductions in net metabolic power during a fourth day of practice. Both tibialis anterior r.m.s. activation and step width remained elevated and did not return to baseline values observed in unpowered walking until the end of the third session. These results indicate that motor adaptation to bilateral powered assistance is not complete until ~90 min of practice. Thus, learning to walk with bilateral exoskeletons appears to be a more challenging task than learning to walk with a unilateral exoskeleton. Our results also suggest that changes in net metabolic power may occur more slowly than changes in joint kinematics and muscle activation patterns during adaptation to powered walking.

One limitation of our study was that the exoskeletons added mass to the lower limbs of the subjects, increasing the metabolic cost of walking compared with walking without the exoskeletons. Added distal mass (applied at the feet) increases the net metabolic cost of walking by ~8% per added bilateral kilogram (Browning et al., 2007). We compared the net metabolic power for powered *versus* unpowered exoskeleton walking, rather than for powered *versus* without exoskeleton walking, to prevent any increases in metabolic cost due to added distal mass from affecting our results. The inverse dynamics analysis we carried out to assess lower limb joint powers also accounted for added exoskeleton mass and inertia, and should therefore reflect the additional mechanical work required to swing the legs.

Although net metabolic power was reduced by powered assistance, the reduction was not as large as expected. Contrary to our hypothesis, net metabolic power did not decrease in proportion to the contribution of the average positive mechanical power delivered by the exoskeletons to the total positive mechanical power generated by the ankle, knee and hip. Powered ankle exoskeletons delivered 22% of the total (ankle + knee + hip) positive mechanical power across the lower limb joints, but the net metabolic cost of walking decreased by only 10%.

It was possible that differences in net metabolic power between powered and unpowered conditions could have been confounded by differences in gait kinematics. Studies have demonstrated that the metabolic cost of walking increases with increasing step length (Donelan et al., 2002a), step width (Donelan et al., 2001) and step frequency (Bertram and Ruina, 2001). We compared step length, step width, double support period and step period between powered and unpowered walking in all three sessions. Initially subjects took wider and shorter steps during powered walking. By the end of the third session there were no differences in step width between powered and unpowered walking. Subjects took shorter steps (~2%) during powered compared with unpowered walking but these changes are too small to appreciably affect net metabolic cost (Donelan et al., 2002a).

The metabolic cost of walking with the powered exoskeletons could also have been affected by a number of other factors. Co-activation about a joint can be very costly metabolically. A compensatory dorsiflexor reaction torque during stance could have resulted in smaller than expected reductions in net metabolic power when using the exoskeletons. We reject this possibility because by the end of three practice sessions, tibialis anterior activation was not significantly different during powered *versus* unpowered walking. Although we did not measure muscle activity for more proximal muscles (e.g. quadriceps, hamstrings), previous results indicate that changes in those muscles due to powered ankle assistance are not significant (Gordon and Ferris, 2007).

Another possibility was that adaptation to the powered ankle exoskeletons involved compensation at other joints that incurred a significant metabolic cost. We found no substantial changes in knee or hip joint kinematics due to powered assistance at the ankle at any point during practice. As mentioned previously, EMG analyses of walking with a unilateral powered exoskeleton indicated no differences in quadriceps or hamstrings muscle activation after two 30 min practice sessions (Gordon and Ferris, 2007). Furthermore, in a study of overground walking with unilateral knee-ankle-foot orthoses, we used an inverse dynamics analysis and found no difference in total ankle joint moment during powered *versus* unpowered walking (Sawicki and Ferris, 2006). Thus, we are confident that during powered walking ankle exoskeletons replaced rather than augmented ankle joint torque and muscles at other joints did not substantially change their dynamics.

Another potential confounding factor is that our analyses assumed that positive mechanical work performed by muscles explains the total net metabolic cost of walking. We acknowledge that there are also metabolic costs associated with muscle activation, isometric muscle force production and negative muscle work (Beltman et al., 2004; Hogan et al., 1998; Ryschon et al., 1997), but evidence suggests that metabolic energy expenditure during walking is dominated by the cost of performing positive muscle work. Recent studies indicate that muscles perform substantially more positive work than negative work during level walking (DeVita et al., 2007; Umberger and Martin, 2007). Umberger and Martin used inverse dynamics to compute joint power over the whole gait cycle and reported  $0.72 \text{ W kg}^{-1}$  positive and only  $-0.37 \text{ W kg}^{-1}$  negative average mechanical power summed over the ankle, knee and hip during preferred cadence walking at  $1.3 \text{ m s}^{-1}$  (Umberger and Martin, 2007). Using Umberger and Martin's data, and assuming that work is performed with a 'muscular efficiency' of 0.25 for positive work and  $-1.20$  for negative work (Margaria, 1976), then 91% ( $0.72/0.25=2.88 \text{ W kg}^{-1}$ ) of the net metabolic cost of walking can be attributed to positive work and only 9% ( $-0.37/-1.20=0.30 \text{ W kg}^{-1}$ ) to negative work. Umberger and Martin's data also support the idea that positive and negative work alone (without consideration of the metabolic cost of muscle force), when scaled by the appropriate efficiencies, predicts well the net metabolic power of preferred frequency walking ( $2.88+0.30=3.18 \text{ W kg}^{-1}$ ). Considering these factors, and the fact that our exoskeletons only altered positive joint work during walking, the net metabolic cost of walking should only decrease by 91% of the contribution of average positive mechanical power delivered by the exoskeletons to the summed average lower limb positive joint mechanical power over a stride. Thus, when exoskeletons delivered 22% of the summed joint positive mechanical power over a stride, the net metabolic power should have decreased by only ~20%. Our value of 10% is still substantially lower than the 20% calculated by assuming that only muscle performs the mechanical work during walking (a doubtful assumption given the possibility for passive tissues, e.g. Achilles' tendon, to store and return elastic energy).

We originally hypothesized that ankle joint work is performed with an 'apparent efficiency' of 0.25, but our results indicated a value of 0.61. Predicting the 'apparent efficiency' to be 0.25 relied on two key assumptions: (1) that the 'muscular efficiency' of positive work is ~0.25 and (2) that all of the positive work done at the ankle joint is performed by active muscle shortening.

Our assumption that the 'muscular efficiency' of positive work is ~0.25 was based on a classic study of uphill walking (Margaria, 1968; Margaria, 1976). Margaria argued that on steep uphill slopes, walking becomes like climbing a ladder and muscles produce force mostly while shortening. On steep slopes, whole-body efficiency approaches ~0.25 and should represent fairly well the efficiency of underlying positive muscle work. In addition, studies using pedal ergometry have reported estimates for 'muscular efficiency' that are on average ~0.25 and range from 0.15 to 0.34 (Gaesser and Brooks, 1975; Poole et al., 1992; Whipp and Wasserman, 1969). Direct measurements of the efficiency over full contraction cycles in isolated mammalian muscle range between 0.10 and 0.19 (rat and mouse soleus and EDL) (Smith et al., 2005). In humans, 'muscular efficiency' of positive mechanical work has been reported for the dorsiflexors (~0.15) during *in vivo* concentric contractions (Ryschon et al., 1997). Taking all of these data together, we feel that 0.25 is a reasonable estimate for the efficiency of human muscle performing positive mechanical work. We note that our estimates of 'apparent efficiency' are converted from the performance index

by a factor equivalent to the assumed  $\eta^+_{\text{muscle}}$  (see Eqns 1 and 2). Thus, although the calculated performance index does depend on the assumed value of  $\eta^+_{\text{muscle}}$ , the ‘apparent efficiency’,  $\eta^+_{\text{ankle}}$ , does not. In summary, we are confident that our estimate of the ‘apparent efficiency’ of ankle joint positive work is reliable.

The fact that compliant tendons in series with muscles can deliver positive work by recycling stored strain energy (Biewener and Roberts, 2000; Roberts, 2002) directly challenges our second assumption that all of the positive work done at the ankle joint is performed by active muscle shortening. Elastic energy storage and return (e.g. in the Achilles’ tendon) could lead to calculations of ‘apparent efficiency’ for ankle joint positive mechanical work that are much higher than  $\eta^+_{\text{muscle}}$ . Recent evidence from ultrasound experiments in humans supports this idea. Both soleus and medial gastrocnemius muscles remain nearly isometric during the push-off phase of stance during walking. The available published data indicate that at least 50% of the positive mechanical work delivered by the triceps surae–Achilles’ complex originates from elastic recoil of the Achilles’ tendon (Ishikawa et al., 2005; Lichtwark and Wilson, 2006). Our performance index can be interpreted as an indicator of the upper bound on the fraction of ankle joint mechanical work performed by all active plantar flexor muscles (see Materials and methods). Using  $\eta^+_{\text{muscle}}=0.25$ ,  $\eta^+_{\text{ankle}}=0.61$  is equivalent to a performance index ( $\eta^+_{\text{muscle}}/\eta^+_{\text{ankle}}$ ) of 0.41, indicating that, at most, active muscle shortening contributes 41% of the total ankle joint positive work during walking. If we relax our assumption that  $\eta^+_{\text{muscle}}=0.25$ , and acknowledge that it could range from 0.10 to 0.34, we estimate that active muscles might perform as little as 16% ( $0.10/0.61=0.16$ ) and as much as 56% ( $0.34/0.61$ ) of the total ankle joint mechanical work. Therefore, Achilles’ tendon recoil must deliver between 44% and 84% of the positive work generated at the ankle joint during the push-off phase of level walking. We conclude that if active muscles perform 16% to 56% of the ankle joint positive work, then a 22% reduction in positive mechanical power output of the lower limb joints due to powered ankle exoskeletons would yield a 3% to 12% decrease in net metabolic power. Our observed 10% reduction in net metabolic power falls within this range.

#### IMPLICATIONS AND FUTURE WORK

Biomechanists and physiologists have been debating the metabolic costs of human walking for more than 75 years (Alexander, 1991; Cavanagh and Kram, 1985; Elftman, 1939; Kuo et al., 2005; Ruina et al., 2005; Saunders et al., 1953; Taylor, 1994; Williams, 1985). A clear relationship between the mechanics and energetics of locomotion remains elusive principally because of the challenge in integrating results from isolated muscle experiments to explain whole-body energy consumption. Using robotic exoskeletons to perturb joint level dynamics can help integrate measurements from isolated muscle with whole-body experiments. Exoskeletons for more proximal joints (hip and knee) could allow calculations of their ‘apparent efficiency’ and provide some insight into the relative contribution of muscle work *versus* tendon storage and return at each of the lower limb joints. Similar techniques could also be used to study joint muscle–tendon function during locomotion under various workloads (e.g. changing walking speeds or surface inclines) to investigate how muscle–tendon systems meet increasing demand for power.

Our analysis was based on a work/efficiency description of the relationship between locomotor mechanics and energetics. Others have argued that the cost of producing muscle force (not work) is a more reliable predictor of energy consumption during locomotion (Griffin et al., 2003; Kram, 2000; Pontzer, 2005; Pontzer, 2007).

Future work could use exoskeletons and a force/economy approach to shed light on the issue of cost of muscle force *versus* cost of muscle work as the primary determinant of metabolic energy consumption during walking.

From an applied science standpoint, our findings have implications for the design of the state-of-the-art lower limb assistive devices of the future (i.e. exoskeletons and prostheses). A primary goal of robotic exoskeletons is to reduce metabolic energy expenditure during human locomotion by replacing biological muscle work with artificial muscle work (Guizzo and Goldstein, 2005). Our results suggest that metabolic energy savings are likely to be much more modest than expected when using an exoskeleton to supplant joint work, especially at joints with considerable elastic compliance. Powering joints that rely more on power production due to positive muscle work rather than positive work performed by recoiling tendon may lead to larger reductions in metabolic cost (Ferris et al., 2007).

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