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# The advantages of a rolling foot in human walking

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## Summary

**The plantigrade human foot rolls over the ground during each walking step, roughly analogous to a wheel. The center of pressure progresses on the ground like a wheel of radius 0.3 *L* (leg length). We examined the effect of varying foot curvature on the mechanics and energetics of walking. We controlled curvature by attaching rigid arc shapes of various radii to the bottoms of rigid boots restricting ankle motion. We measured mechanical work performed on the center of mass (COM), and net metabolic rate, in human subjects (*N*=10) walking with seven arc radii from 0.02–0.40 m. Simple models of dynamic walking predict that redirection of COM velocity requires step-to-step transition work, decreasing quadratically with arc radius. Metabolic cost would be expected to change in proportion to mechanical work. We measured the average rate of negative work performed on the COM, and found that it followed the trend well ($r^2$=0.95), with 2.37 times as much work for small radii as for large. Net metabolic rate (subtracting quiet standing) also decreased with increasing arc radius to a minimum at 0.3 *L*, with a slight increase thereafter. Maximum net metabolic rate was 6.25 W kg$^{-1}$ (for small-radius arc feet), about 59% greater than the minimum rate of 3.93 W kg$^{-1}$, which in turn was about 45% greater than the rate in normal walking. Metabolic rate was fit reasonably well ($r^2$=0.86) by a quadratic curve, but exceeded that expected from COM work for extreme arc sizes. Other factors appear to increase metabolic cost for walking on very small and very large arc feet. These factors may include effort expended to stabilize the joints (especially the knee) or to maintain balance. Rolling feet with curvature 0.3 *L* appear energetically advantageous for plantigrade walking, partially due to decreased work for step-to-step transitions.**

Key words: metabolic energy, locomotion, biomechanics, rocker bottom.

## Introduction

During each step of human walking, the center of pressure exerted against the ground progresses forward from heel to toe. This progression is similar to the rolling of a wheel, with the complex actions of the ankle, foot and shoe somehow resulting in an overall motion analogous to that of a rigid curved surface. Rolling contact of the entire foot with the ground is characteristic of plantigrade gaits, and is unique to humans among bipeds. Other bipeds such as birds employ a digitigrade gait that allows for long stride lengths because the foot can be extended during ground contact. The relatively flexed, plantigrade foot need not, however, be at a disadvantage. The effective shape or curvature emulated by the rolling foot may, for example, offer mechanical or energetic benefits. Here we examine the mechanical and metabolic consequences of different rolling foot curvatures during human walking.

Empirical evidence indicates that humans normally produce a particular effective foot curvature. The forward progression of the center of pressure is similar to that of a rolling wheel with radius equal to 30% of leg length (McGeer, 1990a). Hansen et al. proposed another method for evaluating the effective 'roll-over shape' of the knee-ankle-foot complex (Hansen et al., 2004), by transforming successive center-of-pressure locations during a step into a limb-fixed coordinate system and fitting a curve to these locations. They found that a simple circular shape matched empirical data well, with a radius of curvature agreeing closely with McGeer's 30% of leg length (McGeer, 1990a). They also found human effective roll-over shape to be remarkably invariant to factors such as walking speed, shoe height and carried load (Hansen and Childress, 2004; Hansen and Childress, 2005; Hansen et al., 2004).

Curvature of the foot bottom has long been exploited in rehabilitation applications. Therapeutic shoes are designed with curved, rocker-bottom surfaces for patients with peripheral neuropathy, diabetic ulcers or transmetatarsal amputation. These shoes reduce plantar pressure under the forefoot and improve walking performance (e.g. Schaff and Cavanagh, 1990). For persons wearing a rigid lower limb cast that immobilizes the ankle, cast shoes provide a rocker bottom shape, promoting a more natural gait (Dhalla et al., 2003; Wu et al., 2004). However, despite the clear benefits provided by these aids, there is little understanding of how rolling foot curvature affects the mechanics and energetics of walking.

The rolling foot may be studied with dynamic walking models. These models liken the stance leg to an inverted pendulum and the swing leg to a swinging pendulum (Mochon and McMahon, 1980). McGeer showed that the coupled pendulums, with a collisional ground contact for the stance foot, can produce a passive dynamic walking gait (McGeer, 1990a). Modeling the feet with circular arcs rigidly attached to the legs, McGeer found that the cost of transport decreased as the arc's radius of curvature increased. One might expect the curved foot's advantage to arise from a greater distance traveled during the stance phase. However, a radius of curvature of 30% of leg length (McGeer, 1990a) confers negligible distance advantage compared to a point foot. Nor is there an advantage in the pendulum motion of either leg, which is conservative of mechanical energy for either curved or point feet. This suggests little advantage to the rolling itself.

The advantage of curved feet may be from their effect on step-to-step transitions. Step-to-step transitions refer to the work performed to redirect the body's center of mass (COM) velocity from one step to the next (Donelan et al., 2002b). The leading leg performs negative work and the trailing leg positive work as the COM velocity is redirected from the pendular arc prescribed by the stance leg to the corresponding arc for the next step (Kuo, 2002). In normal human walking, much of this work occurs simultaneously during double support (Donelan et al., 2002b), with an approximately proportional metabolic cost (Donelan et al., 2002a). In dynamic walking models, curved feet reduce the directional change that the COM velocity must undergo (McGeer, 1990a; Ruina et al., 2005), reducing step-to-step transition work. The magnitude of step-to-step transition work theoretically will decrease with increasing radius of curvature of the feet, potentially leading to decreases in metabolic cost with the amount of work.

The purpose of this study was to quantify the effects of an imposed rolling foot curvature on the work performed on the COM during human walking, and on the associated metabolic cost. We imposed a rigid, curved foot surface on human subjects, manipulating the radius of curvature experimentally. We counteracted the human tendency to preserve a single effective roll-over shape by rigidly constraining the ankles. Subjects therefore rolled forward on the foot surface much like dynamic walking models (e.g. Kuo, 1999; McGeer, 1990a). We hypothesized that curved feet of small radius would result in high step-to-step transition costs, in terms of both work performed on the COM and metabolic energy consumption. We expected these costs to decrease with increasing radius of curvature. However, the theoretical dependency cannot apply to all radii, because it predicts the lowest cost at an impractically large radius of curvature equal to leg length. We therefore sought to test the hypothesis of step-to-step transitions, and to evaluate the limitations of the theory as applied to actual humans.

## Materials and methods

We designed an experiment to rigidly constrain ankle motion and impose different foot curvatures on subjects, and observed the impact of these changes on COM work and metabolic cost of walking. We constructed a simple boot apparatus to fix subjects' ankle joints in a neutral position. This allowed us to restrict the ankle's dynamic action and impose different static curvatures that could be manipulated experimentally. We measured ground reaction forces while subjects walked over force plates wearing different curves. We also measured metabolic rate during matched trials of treadmill walking. Finally, we compared the two data sets to elucidate how changes to curvature affect the work performed on the body center of mass (COM), and in turn how this work affects the metabolic cost of walking. Before describing the experiments in more detail, we use a simple model of walking to predict the effects of changes to radius of curvature.

### *Model*

A simple walking model illustrates the influence of foot curvature on step-to-step transitions (Fig. 1). This model is based on the Simplest Model of walking on level ground (Kuo, 2001), with the addition of arc-shaped feet. The model has a point mass at the pelvis, with infinitesimally small point masses at the bases of the feet (Fig. 1A). The model can be powered by an instantaneous push-off impulse applied under the stance foot just before contralateral heelstrike (Kuo, 2001). This push-off impulse performs positive work on the COM, of magnitude $W^+$. Immediately thereafter, the collision of swing leg with ground performs negative work, of magnitude $W^-$. For a periodic gait at steady speed, $W^+=W^-$.

The step-to-step transitions may be computed as a function of the foot's radius of curvature, $\rho$. Push-off and heelstrike impulses are directed from the ground contact points to the COM. These impulses successively redirect the COM velocity. The push-off impulse redirects the COM from its pre-transition velocity $v_{\mathrm{pre}}$ to a mid-transition velocity $v_{\mathrm{mid}}$; then the heelstrike impulse redirects the COM to a post-transition velocity $v_{\mathrm{post}}$. A curved foot reduces the directional change in COM velocity, and the work performed to redirect the COM (see Fig. 1B). For a leg at angle $\alpha$ with respect to vertical at the step-to-step transition, and a positive radius of curvature $\rho$, the pre-to-post angular direction change $\delta$ in COM velocity is less than the angle between the legs $2\alpha$. A periodic gait is produced (Kuo, 2002) if this net directional change is shared equally between the push-off and collision impulses (see Fig. 1C). From the geometry of these impulses,

$$\tan\frac{\delta}{2} = \frac{(1-\rho)\sin\alpha}{\rho+(1-\rho)\cos\alpha}\ . \tag{1}$$

A small angle approximation for $\alpha$ yields:

$$\tan\frac{\delta}{2} \approx \alpha(1-\rho)\ . \tag{2}$$

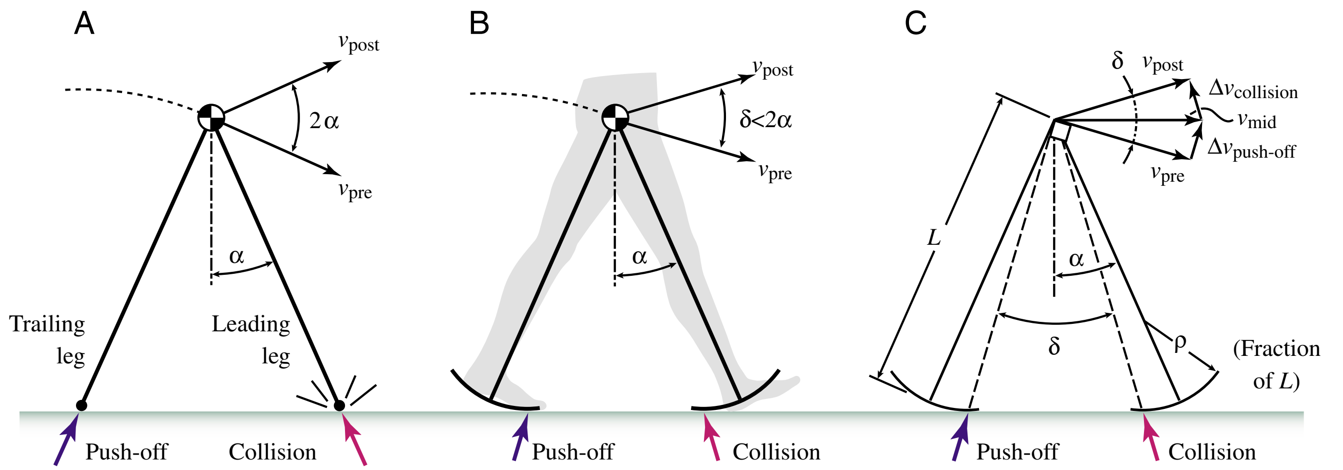


Fig. 1. A simple model demonstrates how a rolling foot can affect walking energetics. (A) Modeling the legs as pendulums supporting the body center of mass (COM), a step can be produced by passive limb dynamics with no energy input (McGeer, 1990a). Work is required, however, in the step-to-step transition to redirect the COM velocity. This can be accomplished with positive push-off work performed by the trailing leg, and negative collision work by the leading leg (Kuo, 2002). These leg actions redirect the pre-transition COM velocity $v_{pre}$ to a post-transition velocity $v_{post}$. For point feet, the net directional change in velocity is equal to the angle between the legs, $2\alpha$. (B) A model with arc feet applies collision at the heel of the leading leg, and push-off at the toe of the trailing leg. This reduces the directional change $\delta$ in COM velocity and therefore the step-to-step transition work. (C) COM velocity change may be understood geometrically. The pre-transition velocity $v_{pre}$ is directed perpendicular to the line from the trailing leg's rolling point of ground contact to the COM. Push-off, directed along this line (angle $\delta/2$ from vertical), causes a change in velocity ($v_{mid}=v_{pre}+\Delta v_{push\text{-}off}$). A periodic gait is achieved if push-off and collision velocity changes ($\Delta v_{push\text{-}off}$ and $\Delta v_{collision}$, respectively) are of the same magnitude, so that $v_{post}$ is equal in magnitude to $v_{pre}$ but directed according to rolling of the leading leg. Work is proportional to the square of each velocity change. As the arc foot radius ($\rho$, defined as a fraction of leg length $L$) increases, less step-to-step transition work is needed. There is no redirection of COM velocity for a radius equal to leg length, $\rho=1$.

The magnitude $W^-$ of the negative work performed each step by the heelstrike collision is equal to the change in kinetic energy:

$$W^- = \frac{1}{2}Mv_{mid}^2 - \frac{1}{2}Mv_{post}^2 \ . \tag{3}$$

The geometric relationship between $v_{mid}$ and $v_{post}$ (see Fig. 1C) yields:

$$W^- = \frac{1}{2}Mv_{post}^2 \tan^2\frac{\delta}{2} \ . \tag{4}$$

The overall trend is revealed by substituting Eqn 2 into Eqn 4:

$$W^- \approx \frac{1}{2}Mv_{post}^2\,\alpha^2(1-\rho)^2 \ . \tag{5}$$

The model therefore predicts the trends in COM velocity change and step-to-step transition work as a function of foot radius of curvature $\rho$. Keeping step length fixed, the step-to-step transition leg angle $\alpha$ is nearly constant (varying only by a few percent over the range of $\rho$ applied in our experiment). Keeping walking speed fixed, the post-transition velocity $v_{post}$ is also approximately constant. Again assuming small angles, the angular direction change $\delta$ in COM velocity then decreases approximately linearly with foot radius of curvature $\rho$:

$$\delta \propto (1-\rho) \ . \tag{6}$$

The trend in the magnitude of negative COM work performed is:

$$W^- \propto (1-\rho)^2 \ . \tag{7}$$

For a constant-speed gait, $W^+=W^-$, allowing Eqn 7 to predict the trend for positive COM work as well. This prediction forms the basis for comparisons to measured data.

We used numerical simulations to verify the analytical prediction of Eqn 7, and to quantify how well it holds for more realistic models (Fig. 2). The 'Simplest Model' (SM) analyzed above neglects leg mass and inertia to allow our closed-form energetic analysis. An 'Anthropomorphic Model' (AM) introduces more human-like mass distribution, but retains straight legs and curved feet that extend fore and aft from the legs (Kuo, 2001; McGeer, 1990a). A 'Forward-foot Model' (FM) moves the anthropomorphic model's feet forward from the leg axis, more like human feet [similar to (McGeer, 1990b), but without knees]. Finally, a 'Kneed Model' (KM) introduces a hinged knee joint to the forward-foot model, with a stop to prevent hyperextension (McGeer, 1990b). The Anthropomorphic and Kneed Models (AM and KM) both resemble physical machines constructed by McGeer (McGeer, 1990a; McGeer, 1990b). All of these models include springs about the joints in order to produce human-like step frequencies (Dean and Kuo, 2005; Kuo, 2001). We examined the gaits of all of these models as a function of $\rho$, keeping speed, step length and other parameters fixed.

These models have different absolute step-to-step transition costs, but all exhibit a net decrease in cost over the range of $\rho$ explored in our human experiment (Fig. 2). However, the

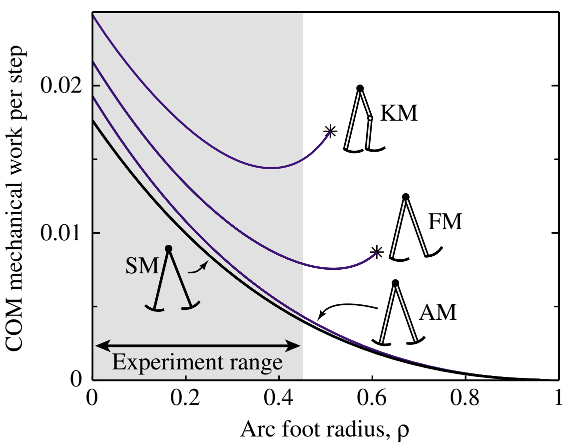


Fig. 2. Work performed on COM as a function of foot radius of curvature ρ, from various dynamic walking models. Models are powered by push-off to walk on level ground: the 'Simplest Model' (SM) with point mass pelvis and feet (Kuo, 2001), the 'Anthropomorphic Model' (AM) with human-like mass distribution (Kuo, 2001), the 'Forward-foot Model' (FM) with feet facing forward from the legs, and the 'Kneed Model' (KM) with knees and forward feet (after McGeer, 1990b). All simulations generally predict decreasing step-to-step transitions with increasing arc foot radius, roughly in proportion to $(1–\rho)^2$ as in Eqn 7. However, FM and KM have a slight upward trend for larger values of ρ, due to different foot geometry and introduction of knees. The SM is used as a prediction for experimental results. Over the range of arc radii studied experimentally, all other models match the trend of Eqn 7 reasonably well, with $r^2$ ranging 0.940–0.998.

decrease is monotonic only for SM and AM. Simulations show that SM closely follows the curve of Eqn 7 to a minimum of zero cost at ρ=1. AM follows the same trend remarkably well, despite the different mass distribution of the legs. The other models, FM and KM, exhibit a U-shaped curve, where step-to-step transition costs increase beyond a certain radius of curvature. FM has a minimum cost at approximately ρ=0.52, with an increasing cost due to the unfavorable mass distribution of the leg relative to the point of collision with ground. KM has a minimum at ρ=0.38 for the same reason, but with even higher costs due to increasing energy lost during knee lock. These latter models also do not yield walking gaits at larger radii (ρ>0.61 and ρ>0.51, respectively, marked with asterisks in Fig. 2), without a change in other parameters. Despite these significant differences in actual behavior, the simple trend of Eqn 7 applies remarkably well to all models over the experimental range of ρ (up to 0.45), with an $r^2$ value of at least 0.94. For this reason, we compared experimental data against the same single trend, predicting a general decrease in step-to-step transition work proportional to $(1–\rho)^2$.

We hypothesized that the mechanical work of step-to-step transitions would be accompanied by an approximately proportional metabolic cost. Work performed actively by muscle would be expected to exact a proportional metabolic cost. Indeed, both step-to-step transition work and metabolic cost measured in humans undergo changes proportional to the work predicted by models as a function of step length and step width (Donelan et al., 2001; Donelan et al., 2002a; Kuo et al., 2005).

There are, of course, many other factors likely to contribute to overall cost. Muscles incur metabolic cost due to their use in supporting body weight, controlling posture and stability, moving the legs, and moving other parts of the body such as the arms and trunk (Doke et al., 2005; Kuo, 2001). We assume that these other costs are relatively constant when only radius of curvature ρ is varied, and step length and frequency are fixed. Any relatively constant costs would contribute to an offset in mechanical work rate or metabolic rate, but would not affect the general trend of Eqn 7.

We also considered an alternative explanation that the metabolic cost of walking reflects work performed by muscles to raise the COM during each step (Saunders et al., 1953). Following this hypothesis, metabolic cost should be proportional to vertical displacement of the COM, with a muscular efficiency of approximately 25% relative to work against gravity (Margaria, 1976). The application of different radii ρ is predicted to cause small changes in vertical COM displacement, yet large changes in COM velocity direction. If raising the COM, rather than redirecting its velocity, is a more important contributor to metabolic cost, we would therefore expect much smaller changes in metabolic rate than predicted by Eqn 7. We therefore compared metabolic cost measured in subjects against their measured vertical COM displacement.

### *Experimental procedure*

We measured mechanical work performed on the COM and metabolic rate while 10 adult human subjects walked in rigid boots with soles of different curvature. Walking speed was fixed at 1.3 m $s^{-1}$ and step frequency was fixed across conditions for each subject. All subjects (5 male, 5 female; body mass 67.5±9.6 kg, mean ± s.d.; leg length 0.94±0.07 m, floor to greater trochanter) were healthy and had no known gait abnormalities. The study was approved by the local Institutional Review Board and subjects gave their informed consent to participate prior to the experiment.

The experimental apparatus consisted of a pair of rigid walking boots modified to accept circular foot surfaces in place of their standard soles (see Fig. 3). The boots were Aircast Pneumatic Walkers (Aircast, Inc., Summit, NJ, USA), with the bottom surface replaced by an aluminum plate with a pyramidal prosthesis adapter. These adapters allowed attachment of foot surfaces (referred to as 'arcs'), circular segments as viewed in the sagittal profile, cut from pine wood 0.086 m wide. Pairs of arcs were constructed with seven different radii of curvature (0.02, 0.05, 0.10, 0.15, 0.225, 0.30 and 0.40 m; see Fig. 3B). All arcs had sufficient fore–aft extent to ensure rolling contact with the ground throughout a normal stance phase. Arcs were matched in weight (1.1±0.1 kg each) and standing height (0.11 m), although moment of inertia could not be matched over this range of sizes. All arcs were attached to the same boots (0.85 kg medium, 1.05 kg large). Subjects walked with

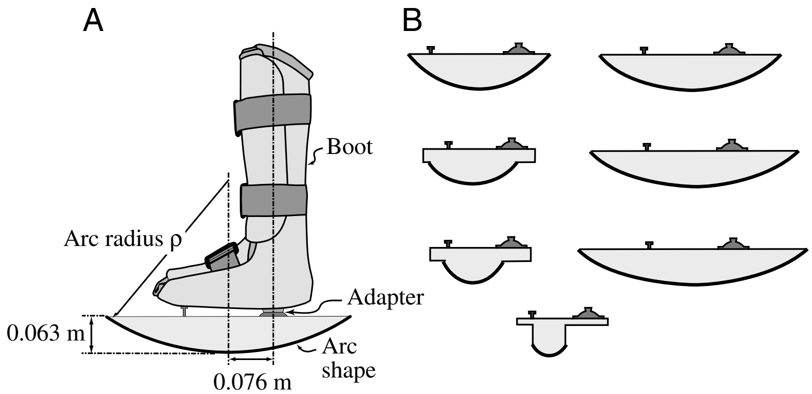


Fig. 3. Apparatus used to rigidly restrict human ankle motion and control rolling characteristics of the foot. (A) Subjects wore a boot and arc apparatus bilaterally, each consisting of a rigid walking boot modified to accept wooden arc shapes of varying radius. (B) 'Arc foot' shapes of varying arc radius ρ (defined as fraction of leg length) were rigidly attached with pyramidal prosthesis adapters. Arcs ranged in radius 0.02–0.40 m in absolute dimensions, and each subtended a sufficient range of angles to ensure continuous rolling ground contact throughout the stance phase. Arcs had matched mass of 1.1±0.1 kg, and boots had mass 0.85 or 1.05 kg, depending on size.

each pair of arcs and in normal street shoes ('normal walking'), with the order of arc conditions randomized for each subject.

Arcs were positioned relative to the leg so that the arc center was 0.076 m anterior to the tibial axis (Fig. 3A). Through trial and error experimentation we determined that the offset could affect walking comfort and metabolic cost. A zero offset (aligning the arc center directly with the tibial axis) caused the ground reaction force to pass behind the knee early in the stance phase. To prevent the knee from buckling, subjects counteracted this flexion moment with high quadriceps activity. A forward offset reduced the buckling moment, but larger offsets led to increasing discomfort due to a knee extension moment late in the stance phase. The offset of 0.076 m was found to provide reasonable compromise between these two factors.

Walking speed was held constant at 1.3 m s$^{-1}$ for all trials, with a subject-specific fixed step frequency. Step frequency was fixed to control for the cost of moving the legs, which increases with step frequency (Doke et al., 2005), and to match our constant step frequency simulations. The particular value chosen was dependent on each subject's preferred step frequency for large arcs. Subjects briefly practiced walking over ground and on a treadmill (Star-Trac, Irvine, CA, USA) with each arc until they felt comfortable with their gait. Prior to experimental trials, we measured each subject's preferred step frequency while they walked with the largest arcs, which were expected to be the most difficult to move quickly due to their inertia. We then tested whether subjects could comfortably maintain this same frequency on the smallest arcs. If not, we measured the lowest frequency they could achieve and used that as the enforced frequency. The mean step frequency thus chosen was 1.74±0.09 Hz (mean ± s.d.), slightly slower than the typical normal walking step frequency of about 1.8 Hz (Donelan et al., 2002a).

Trials were performed both over ground and on a treadmill for the same conditions. We measured ground reaction forces (GRFs, see Fig. 4) in the over-ground walking trials. Subjects walked across two sequential force plates (AMTI, Watertown, MA, USA) at the same speed and step frequency used in treadmill walking. Speed was measured using two photogates, positioned 2.5 m apart around the force plates, and the chosen step frequency was regulated by a metronome. Trials were discarded if speed was not within 5% of the nominal 1.3 m s$^{-1}$ speed. We assessed the net change in speed per trial to be +0.012±0.050 m s$^{-1}$ (mean ± s.d.) for normal walking and +0.017±0.052 m s$^{-1}$ for arc foot conditions. Both were statistically insignificantly different from zero ($P$>0.05), indicating that subjects walked at relatively steady speed. We recorded ten successful trials for each subject on each pair of arcs, and averaged the GRF from all ten trials. A step was defined as beginning with heelstrike and ending with opposite heelstrike.

We used GRF data to estimate the COM velocity changes and the average rate of negative mechanical work performed on the COM over the step cycle. We calculated COM

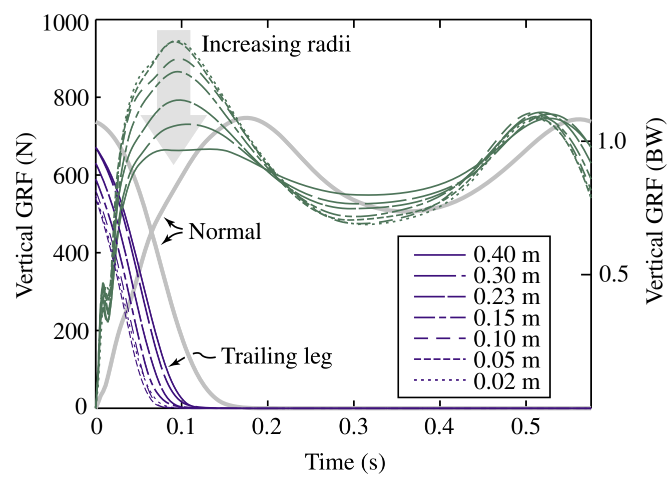


Fig. 4. Vertical ground reaction forces (GRF) *versus* time over one step, measured during walking with arcs of different radius and in normal shoes. Larger arc radii resulted in smoother collisions during the step-to-step transition. Small arc radii resulted in very large initial peaks in ground reaction force. With larger arcs this peak decreased to below its magnitude in normal walking, but it always occurred earlier in the step. Walking on arcs resulted in shorter double-support times, decreasing with smaller radii. Arc radius had little effect on the second peak in vertical force. Data shown are averaged over all subjects and plotted over the mean step period. A step begins at heelstrike and ends at opposite heelstrike, with double support occurring over the first 0.10–0.15 s. BW, body weight. Blue and green traces indicate forces under the trailing and leading legs, respectively.

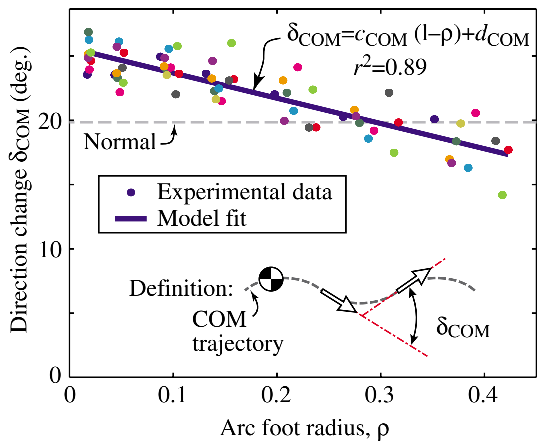


Fig. 5. The angular direction change $\delta_{\mathrm{COM}}$ in COM velocity decreased with increasing arc foot radius ρ. COM direction change was estimated as the angle between the steepest upward and downward velocities of the COM in the sagittal plane (defined below; compare with Fig. 1B). The relationship between $\delta_{\mathrm{COM}}$ and ρ is described well by the linear fit of Eqn 9, $r^2$=0.89. Different coloured symbols indicate different subjects.

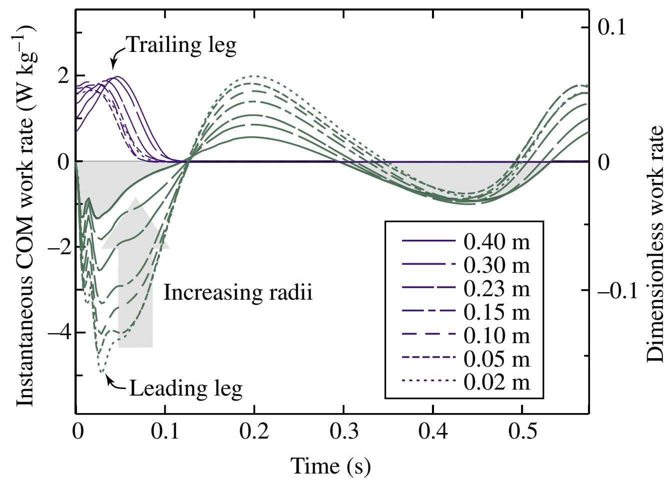


Fig. 6. Instantaneous COM mechanical work rate for each leg over one complete step, measured with arcs of different radii. The trailing leg performed positive work and the leading leg negative work to redirect the COM during the step-to-step transition. Leading leg negative work rate was highest in magnitude for small-radius arcs. Work rate magnitudes decreased with increasing arc radius for the leading leg during double support, and through most of single support. Average rate of negative work was computed by integrating the magnitude of negative regions of instantaneous work rate (shaded areas for 0.40 m arc) and dividing by step period. Data shown are averaged from all subjects and plotted over the mean step period. Blue and green traces indicate work rate of the trailing and leading leg, respectively.

kinematics (linear acceleration, velocity, and position) from three-dimensional GRF data, assuming periodic gait (Donelan et al., 2002b). The velocity data were then used to derive the maximum angular change $\delta_{\mathrm{COM}}$ in the direction of COM velocity in the sagittal plane (see Fig. 5). The instantaneous rate of mechanical work performed by each leg on the COM was calculated according to the ‘individual limbs method’ (Donelan et al., 2002b), as the dot product of each leg’s GRF and the COM velocity (Fig. 6). We integrated the combined negative portions of the individual limbs’ work rate (Fig. 6, shaded area) to find the total negative work $W^{-}_{\mathrm{mech}}$ (J) performed during one step. Finally, we multiplied this work by step frequency (Hz) to yield the average rate of negative mechanical work $\dot{W}^{-}_{\mathrm{mech}}$ (in W) performed by the subject on the COM. For comparison purposes, we also calculated the average rate of negative COM work performed during double support alone, $\dot{W}^{-}_{\mathrm{DS}}$.

We estimated metabolic energy expenditure rate from respiratory gas exchange data collected during treadmill trials. Subjects walked on the treadmill for at least 7 min while we collected data. Steps were again regulated by a metronome set to the chosen step frequency. We used an open-circuit respirometry system (Physio-Dyne Instrument Corp., Quogue, NY, USA) to measure the volume rates of oxygen consumption and carbon dioxide production ($\dot{V}_{\mathrm{O_2}}$ and $\dot{V}_{\mathrm{CO_2}}$, ml s$^{-1}$). Following a 3.5-min transient period to allow subjects to reach steady state, we collected and averaged volume rates over at least 3 min of each trial. Metabolic energy expenditure rate $\dot{E}_{\mathrm{met}}$ (W) was estimated using the formula:

$$\dot{E}_{\mathrm{met}} = 16.48\dot{V}_{\mathrm{O_2}} + 4.48\dot{V}_{\mathrm{CO_2}}\ , \qquad (8)$$

where constants 16.48 J ml$^{-1}$ and 4.48 J ml$^{-1}$ arise from analysis of substrate metabolism [after (Brockway, 1987) and (Weir, 1949)]. Finally, we calculated net metabolic rate by subtracting the metabolic rate of quiet standing. The quiet standing data collection procedure was similar to that of the walking tests, except that it was administered after at least 5 min of seated rest. Two subjects reported discomfort when walking on some arcs. These conditions were terminated early, and data were not collected for a total of five trials.

*Data analysis*

We used angular change in COM velocity, average COM work rate, and metabolic rate to test the simple model’s predictions for changes in arc radius. First, we performed a least-squares fit to the model of Eqn 6, regressing velocity direction change $\delta_{\mathrm{COM}}$ against arc radius ρ according to:

$$\delta_{\mathrm{COM}} = c_{\mathrm{COM}}(1-\rho) + d_{\mathrm{COM}}\ . \qquad (9)$$

Coefficients $c_{\mathrm{COM}}$ and $d_{\mathrm{COM}}$ accommodate differences between humans and the model, such as knee flexion and duration of step-to-step transition, that can affect measured $\delta_{\mathrm{COM}}$.

We regressed subjects’ mechanical and metabolic costs against arc radius according to:

$$\text{Simplest Model Fit: } C(1-\rho)^2 + D\ . \qquad (10)$$

$C$ is an unknown scaling coefficient, and $D$ is a constant offset due to costs not affected by arc radius. We applied the same form of fit to three costs: $\dot{W}^{-}_{\mathrm{mech}}$, $\dot{W}^{-}_{\mathrm{DS}}$ and $\dot{E}_{\mathrm{met}}$, applying subscripts 'mech', 'DS' and 'met', respectively, to $C$ and $D$ to distinguish the various coefficients.

We also performed a more general quadratic fit for metabolic rate $\dot{E}_{\mathrm{met}}$. To allow for a minimum cost not occurring at $\rho$=1, we performed a least-squares fit to a general quadratic,

$$\text{Empirical Fit: } \dot{E}_{\mathrm{met}} = C_{\mathrm{EF}}(B_{\mathrm{EF}}-\rho)^2 + D_{\mathrm{EF}}\,, \qquad (11)$$

where coefficient $B_{\mathrm{EF}}$ locates the $\rho$ value of the curve minimum. Unlike Eqn 10, which is based on the dynamic walking model of Eqn 7, the Empirical Fit is a purely mathematical curve fit.

We also tested the hypothesis that metabolic cost will increase in proportion to step-to-step transition work. We predicted the metabolic cost of step-to-step transitions by scaling the Simplest Model Fit to COM work rate according to the 25% maximum expected efficiency of muscle work (Margaria, 1976) with an arbitrary offset. We then compared this prediction against the observed Empirical Fit, using the difference between the two as an indication of residual costs not predicted by the simple model.

We compared our results against the idea that metabolic cost arises from work performed against gravity in raising the COM. We calculated the vertical displacement of the COM as the difference between its highest and lowest positions during the step. We multiplied vertical displacement by body weight ($\boldsymbol{Mg}$) to determine the work performed against gravity during each step, and multiplied this work by step frequency to estimate the average rate of work ostensibly performed to raise the COM, $\dot{W}_{\mathrm{raise}}$. We then formed a predicted metabolic cost due to COM raising. We computed a best-fit line to the $\dot{W}_{\mathrm{raise}}$ *versus* $\rho$ data, and divided this by the expected 25% efficiency to obtain a prediction of metabolic rate according to the COM raising explanation.

To account for differences in subjects' body size, we performed all analyses with non-dimensionalized variables. We used base units of total mass $M$ (body plus apparatus), gravitational acceleration $\boldsymbol{g}$, and total standing leg length $L$ (including boots and arc feet). Work rate and energy rate were therefore made dimensionless by the divisor $\boldsymbol{Mg}^{1.5}L^{0.5}$, work and energy by $\boldsymbol{Mg}L$, and force by $\boldsymbol{Mg}$. Arc radius was non-dimensionalized by $L$. Work rate and energy rate graphs and model fits are presented in both dimensionless units and in the more common units of W kg$^{-1}$. Conversion between these units was performed with the mean factor $\boldsymbol{g}^{1.5}L^{0.5}\approx$29.8 W kg$^{-1}$. We also accounted for inter-subject kinematic and energetic variations by computing offsets $d_{\mathrm{COM}}$, $D$ and $D_{\mathrm{EF}}$ separately for each subject and then averaging them.

## Results

The mechanics and energetics of walking changed significantly as a function of arc radius of curvature. The angular direction change in COM velocity occurring in each step decreased with increasing radius $\rho$. The average rate of negative mechanical work performed on the COM also decreased significantly with increasing $\rho$. Net metabolic rate decreased with small increasing values of $\rho$, but increased again after reaching a minimum. Results for ground reaction forces, COM velocity direction change, COM work rate, and metabolic rate during normal walking and walking with arcs are compared below.

We first verified that the measured mechanical work rate and metabolic rate of normal walking were comparable to values found in previous literature. In normal walking at 1.3 m s$^{-1}$ with preferred step frequency, the angular direction change $\delta_{\mathrm{COM}}$ in COM velocity was 19.7°. Subjects performed negative COM work at an average rate of 0.595 W kg$^{-1}$ (non-dimensional value, 0.020). This is equivalent to 0.343 J kg$^{-1}$ per step, comparable to estimates of 0.33 and 0.38 J kg$^{-1}$ per step from two previous studies (Donelan et al., 2002a; Donelan et al., 2002b). Average net metabolic rate for normal walking was 2.71 W kg$^{-1}$ (non-dimensional value, 0.091), also comparable to previously published results (Donelan et al., 2002a).

Measured ground reaction forces changed with arc radius, and differed from those of normal walking. Vertical forces (Fig. 4) exhibited greater overlap with higher radius, expanding the duration of double support from about 6.5% of the stride (two steps) for 0.02 m arcs to 10% for 0.40 m arcs. The early force peak, about 1.4 BW (body weight) for small arcs, decreased to about 1.0 BW for large arcs, possibly because the opposite leg contributed higher forces throughout double support. The second peak's magnitude was about 1.1 BW for all experimental conditions and for normal walking, but its duration was longer for larger arcs. Reflecting the relative rigidity of the boot-arc apparatus compared to a normal foot and ankle, loading of each new stance limb occurred very quickly. Peak load was reached in as little as 8.5% of a stride, compared with about 15% for normal walking.

The observed angular direction change $\delta_{\mathrm{COM}}$ in COM velocity decreased with increasing arc foot radius $\rho$ ($P$<0.05, Fig. 5). These data were fit well ($r^2$=0.89) by the linear prediction of Eqn 9, with coefficients $c_{\mathrm{COM}}$=19.6±3.0° (mean ± 95% Confidence Interval, CI), $d_{\mathrm{COM}}$=6.0±2.8°. The COM direction change for normal walking intersected with the observed trend at an arc radius of about 0.3.

The relative distribution of COM work throughout the step also changed with arc radius (Fig. 6). We define the collision as the first phase of negative COM work in a step, and push-off as the first phase of positive work starting near the end of the preceding step and extending until the end of double support (Kuo et al., 2005). There was a dramatic increase in collision negative work with decreasing $\rho$, occurring over a relatively fixed duration of about 0.13 s. But the duration of double support decreased with smaller arcs, so that the collision tended to extend beyond double support in those conditions. The amount of push-off COM work remained relatively fixed, but tended to occur earlier before heelstrike with smaller arcs. Subjects performed less work during push-off than during collision, making up for the deficit with more positive work in the single-support leading leg prior to mid-stance.

In relation to normal walking, walking on arc feet resulted

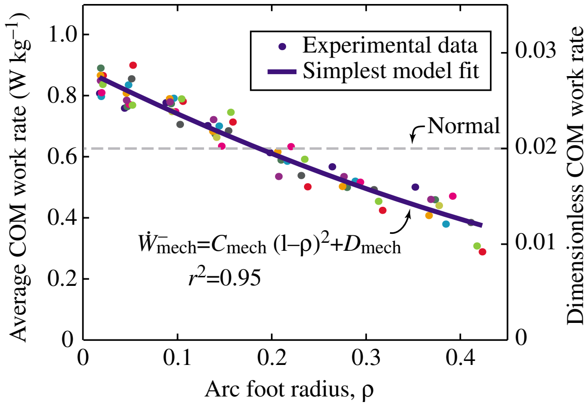


Fig. 7. The average rate at which negative work is performed on the COM ($\dot{\bar{W}}_{\mathrm{mech}}$, see shaded areas in Fig. 6) fell with increasing arc foot radius ρ. The Simplest Model Fit of Eqn 10 predicted the trend well ($r^2$=0.95). The magnitude of work rate was greater for small arcs than for normal walking (broken line), and lower for arcs of approximately ρ>0.2. Less work is needed to redirect the COM velocity with larger arcs, due to a smaller directional change during the step-to-step transition. The work rate observed with the smallest arcs was 2.37 times that for the largest arcs. Different coloured symbols indicate different subjects.

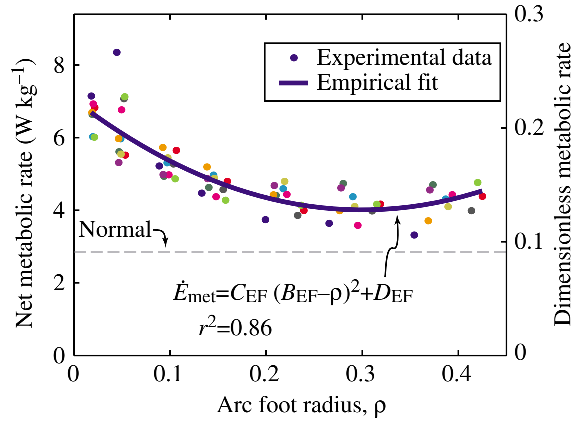


Fig. 8. Net metabolic rate $\dot{E}_{\mathrm{met}}$ exhibited a U-shaped curve as a function of arc radius ρ. For small radii, metabolic rate decreased with ρ much as predicted by the Simplest Model (Fig. 2). However, metabolic rate reached a minimum at ρ=0.30 according to the Empirical Fit of Eqn 11, $r^2$=0.86, and began to increase with larger ρ. The energetic cost of walking was 59% higher than the minimum for the smallest arcs, and higher for all arc radii compared to normal walking (broken line). Different coloured symbols indicate different subjects.

in a roughly comparable average COM work rate but a considerably higher metabolic rate. COM work rate with arcs at 1.3 m s$^{-1}$ ranged from a high of 0.774 W kg$^{-1}$ (dimensionless 0.026) for the smallest arcs to a low of 0.327 W kg$^{-1}$ (0.011) for the largest arcs (Fig. 7). Arcs of radius 0.225 m and greater actually resulted in lower average negative COM work rates than normal walking. However, the Empirical Fit to metabolic rate for walking on arcs was always at least 45% higher than the rate for normal walking (Fig. 8). Net metabolic rate ranged from 6.25 W kg$^{-1}$ (0.210) for the smallest arcs to 3.93 W kg$^{-1}$ (0.132) for the second-largest arcs, and demonstrated a minimum near ρ=0.300.

The amount of negative COM work performed ($\dot{\bar{W}}_{\mathrm{mech}}$) agreed well with the decreasing trend predicted by the Simplest Model (Fig. 7). Overall negative work rate decreased with increasing ρ ($P$<0.05), fitting the Simplest Model fit of Eqn 10 with an $r^2$ value of 0.95. The model fit showed a decline in overall negative COM work rate from 0.774 to 0.327 W kg$^{-1}$ (dimensionless 0.026–0.011) as arc radius increased from 0.02 to 0.42 (Fig. 7). The coefficients are $C_{\mathrm{mech}}$=0.700±0.050 W kg$^{-1}$ and $D_{\mathrm{mech}}$=0.110±0.047 W kg$^{-1}$ (mean ± CI, dimensionless 0.024±0.001 and 0.004±0.001, respectively). A similar trend was observed for double-support work rate $\dot{\bar{W}}_{\mathrm{DS}}$ ($r^2$=0.92), with coefficients $C_{\mathrm{DS}}$=0.617±0.059 W kg$^{-1}$ and $D_{\mathrm{DS}}$=–0.093±0.055 W kg$^{-1}$ (dimensionless 0.0207±0.0020 and –0.0031±0.0019, respectively).

Metabolic rate $\dot{E}_{\mathrm{met}}$ also fell with increasing radius of curvature ($P$<0.05), although with a U-shaped rather than a monotonically decreasing curve (Fig. 8). The Simplest Model (Eqn 10) predicted a decreasing curve with minimum at ρ=1, but the resulting fit to data for $\dot{E}_{\mathrm{met}}$ gave a relatively poor $r^2$=0.77. Metabolic rate was matched better by the purely Empirical Fit of Eqn 11, $r^2$=0.86. The coefficients are $B_{\mathrm{EF}}$=0.300±0.108 (mean ± CI), $C_{\mathrm{EF}}$=32.02±9.40 W kg$^{-1}$ (dimensionless 1.074±0.316), and $D_{\mathrm{EF}}$=3.81±1.65 W kg$^{-1}$ (0.128±0.055).

The predicted metabolic cost for the COM raising hypothesis was far below the observed metabolic cost. Vertical COM displacement decreased approximately linearly from 0.045 m (dimensionless 0.048) for ρ=0.02 to 0.035 m (0.037) for ρ=0.42. The rate of work $\dot{W}_{\mathrm{raise}}$ needed to raise the COM through these displacements therefore ranged from 0.831 W kg$^{-1}$ (dimensionless rate 0.028) to 0.614 W kg$^{-1}$ (0.021). This yields expected metabolic rates of 3.3 W kg$^{-1}$ (0.111) to 2.5 W kg$^{-1}$ (0.083), a range far smaller than observed (Fig. 8). The change in vertical COM displacement could only account for about 24% of the observed change in metabolic rate.

## Discussion

We investigated the effects of arc foot radius ρ on the mechanical and metabolic costs of walking. Our model of walking with arc-shaped feet predicted an energetic cost based on the work performed on the center of mass (COM) in each step-to-step transition. We predicted that the average rate of COM work would fall with increasing arc radius according to Eqn 7. We also predicted that metabolic cost would change in proportion to mechanical work.

The observed downward trend in negative COM work (Fig. 7) indicates that arc radius influences step-to-step transition mechanics much as predicted (Eqn 7). Even with no change in walking speed or step length, less work is needed to walk on larger-radius arcs. This is due to the smaller angular direction change in COM velocity for step-to-step transitions associated with larger radii (Fig. 5). Small radii result in larger directional changes, greater impact forces and more negative work. Subjects compensated for collisions with more positive work, not during double support but during single support (see Fig. 6), perhaps by performing positive work with the hips. Regardless of when work was performed, overall work rate decreased in proportion to the predicted $(1-\rho)^2$ trend.

Curiously, larger-radius arcs actually resulted in less step-to-step transition work than occurred in normal walking. COM work rate was lower for all radii greater than about $\rho$=0.2; the trend exhibited the greatest difference of about 45% at the upper limit of radius, $\rho$=0.42. By the criterion of center of pressure progression, human walking may appear to have an effective roll-over radius of $\rho$=0.3, but by the criterion of step-to-step transition work, normal walking is closer to $\rho$=0.2. The actions of the human ankle-foot complex appear not to perfectly mimic a static arc. Some of the difference reflects active motion in the normal ankle and foot articulations, performed with mechanical work. Passive deformation of soft tissues may also contribute to normal COM work, with inelastic deformations dissipating energy. Passive elastic deformation, for example in the Achilles tendon, may also contribute to normal COM work (Donelan et al., 2002b; Kuo et al., 2005) without dissipating energy. These ankle and foot motions, whether active or passive, elastic or inelastic, are reduced considerably by the arc foot apparatus used in this experiment.

Metabolic rate generally decreased with increasing arc radius, but only to about $\rho$=0.3 (Fig. 8). For larger radii, $\dot{E}_{met}$

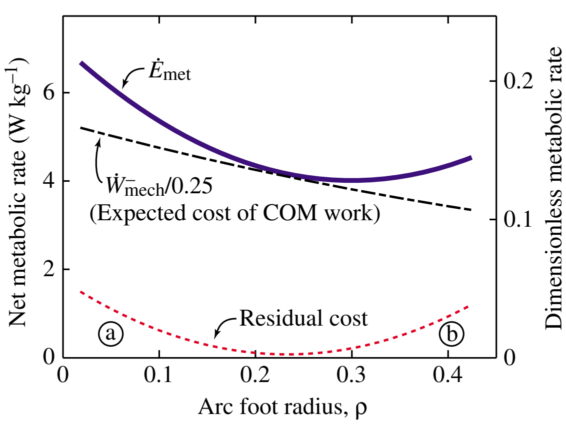


Fig. 9. Comparison of net metabolic rate $\dot{E}_{met}$ with expected cost based on step-to-step transition work. Assuming a peak efficiency of 25%, the observed work performed on the COM (Fig. 7) would be expected to yield a strictly decreasing metabolic rate with increasing $\rho$. Subtracting the expected cost from observed yields a residual cost not explained by the Simplest Model. The residual cost is substantial for arcs of smallest and largest radii. (Region a) The high cost for small radii may be caused by the effort of balancing on a small contact patch through large collisions in the step-to-step transition. (Region b) The cost for large radii may be associated with stabilizing the knee joint against a hyperextension moment caused by the ground reaction force late in single support.

increased with $\rho$. The measured metabolic rate exceeded that predicted by COM work (assuming 25% muscle efficiency) for both low and high values of $\rho$ (Fig. 9). It appears that changes in metabolic cost were largely proportional to COM work rate, but with additional costs that are not captured by the step-to-step transition model. These unmodeled factors affect the cost of walking on unusually large and small arcs. Subjective observations suggest that there may be separate explanations for the increased metabolic cost measured for small or large arcs.

We consider two possible explanations for the unexpectedly high metabolic cost of walking on small-radius arc feet (Fig. 9, region a). First, subjects found it difficult to balance while walking with all arcs, especially the smallest ones. The small radius afforded a small ground contact patch and resulted in short impact durations with little time spent in double support (see Fig. 4). More effort may have been expended to maintain balance, with an added metabolic cost. Second, small arcs resulted in greater collisions at heelstrike, which subjects found jarring and uncomfortable. Their preference would have been to walk at faster step frequencies, had frequency not been controlled. Faster and shorter steps would have reduced the collisions, trading high step-to-step transition costs for forced motion of the legs (Kuo, 2002). Instead, subjects appeared to expend effort to maintain joint stability, particularly for the knee, through the greater collisions. The additional muscle activity for co-contraction or other stabilizing actions may have incurred a metabolic cost.

Other explanations may apply to the high cost of walking on large-radius arc feet (Fig. 9, region b). Late in stance, larger arcs produced a longer moment arm between the knee joint axis and the ground reaction force's line of action, resulting in an extension moment tending to hyperextend the knee during late stance. Subjects reported high activity in knee flexors, presumably to counteract hyperextension. Some subjects also reported high activity in plantarflexor muscles, which may have been used to counteract the bending moment the boot applied to the shank, as well as to stabilize the foot within the boot. Stabilization of the knee and ankle may have contributed to the higher metabolic cost on large arcs.

We also consider the higher rotational moments of inertia of larger arcs. Larger arcs might theoretically require greater effort to swing through a step, depending on their contribution to overall moment of inertia about the medio-lateral axis. The arcs had central moments of inertia of about 0.002–0.013 kg m$^2$ (despite all being matched in mass), compared to a total inertia of about 0.90 kg m$^2$ for the entire lower leg and boot-arc apparatus about an axis passing through the knee. Even the largest arcs therefore contributed less than 2% to overall rotational inertia. This difference cannot explain the cost of walking with large arcs.

Higher step-to-step transition costs for walking with large arcs were also observed indirectly in models with knees. Forward-facing feet (FM and KM of Fig. 2) and a passive knee joint (KM) alter the collision geometry, resulting in higher step-to-step transition costs for larger foot radii. These costs are a function of joint spring stiffnesses in the models. If KM were given infinite knee stiffness, its step-to-step transition work would be identical to that of FM. For a human to stiffen a joint in the same manner, muscle activity would presumably incur some metabolic cost. KM also loses more energy at heelstrike for larger arcs. These phenomena may have affected the human subjects metabolically without appearing in COM work rate estimates.

There was also an overall higher metabolic cost of walking on arc feet independent of arc radius. The constant offset was such that metabolic rate was at least 45% higher for arc foot walking than for normal walking (see Fig. 8), despite the arcs' advantage in terms of mechanical work. One constant factor is that the weight-matched arcs and boot apparatus added about 2.0 kg at the end of the leg in each arc condition. Many studies (Burse and Pandolf, 1979; Inman et al., 1981; Martin et al., 1997; Miller and Stamford, 1987; Skinner and Barrack, 1990) have quantified the metabolic impact of adding mass to the ankles, measuring increases equivalent to 11–24% over normal walking per kilogram added. One study (Royer and Martin, 2005) incrementally varied the location of the mass, and found steadily increasing metabolic costs with more distal placement due to changes in moment of inertia. In our current experiment, the added mass is greater, and it is placed more distally than in any of these studies. Extrapolating from these and the results of other studies (Griffin et al., 2003), a hypothetical 2.0 kg mass centered near the bottom of the foot may increase the net metabolic cost of normal walking by up to 44%.

An additional factor may have been the novelty of walking on arc feet. After brief practice sessions, subjects may not have fully adapted to the added mass, restricted ankle motion, smaller ground contact patch, and rigid arcs. We performed a repeatability test on two subjects, and found roughly a 10% decline in cost from their first arc condition to a post-experiment re-test of the same condition. Practice may help subjects to improve balance and control, reducing metabolic cost. Novelty may therefore have contributed to the overall cost of walking on arcs, but not to the observed trends in cost due to randomized trial order. Factors such as added mass, increased moment of inertia, decreased double-support time, difficulty of balancing on the arcs, the need to compensate for restricted ankle motion, and incomplete adaptation could all contribute to the higher overall cost we measured for walking with arc feet.

The metabolic cost of walking on arc feet is not well explained by the alternative hypothesis of raising the COM against gravity. Based on the measured changes in vertical displacement of the COM, work performed against gravity (at 25% efficiency) would account for only about 24% of the observed changes in metabolic rate as a function of $\rho$. This hypothesis is also at odds with the inverted pendulum analogy for the stance leg, because a pendulum can conserve mechanical energy, gaining height by conversion of kinetic energy to potential energy. Work is therefore not needed to raise a pendulum, which will have the same energy and speed at the beginning and end of single support. Even with a conservative pendulum, however, work is needed to restore energy lost in collisions. We find the explanation based on step-to-step transitions to be more helpful than that based on raising the COM.

Arc feet allow rolling during single support and reduce step-to-step transition costs. For rolling, a rigid convex curved shape will dissipate less energy than one that is deformable, because deformations cause rolling resistance. Polygonal or concave shapes (e.g. a rigid cast without a cast shoe) are poor choices because each corner produces a collision as it contacts the ground (Ruina et al., 2005). However, the circular shape we examined is not necessarily optimal. An inverted pendulum can theoretically roll atop any smooth convex curve. Longer (fore–aft) curves reduce the directional change in COM velocity and therefore step-to-step transition work (Eqn 6, Eqn 7). For longer curves, some attention must be paid to alignment with respect to the tibial axis, and to induced moments about the knee. Such factors would warrant further study for possible application to rocker bottom shoes, which evidently already employ them to advantage but without quantitative, energetics-based design principles.

The human plantigrade gait appears to use the feet to behave approximately like rigid arcs. The effective roll-over shape ($\rho$=0.3 based on center of pressure progression) appears to take advantage of reduced step-to-step transition costs compared to a point foot ($\rho$=0), subject to the limitations apparent with larger arc radii. The disadvantages of larger arcs might stem from side effects such as moments induced about the knee. For animals that walk exclusively on flat ground, it might be preferable to have rigid legs with curved feet of radius equal to leg length, and without ankles or knees. However, animals that wish to sit, stand, climb, or use ankles or knees for any other purpose must compromise the efficiency of high-radius rolling gait with the body's structural limits and versatility constraints.

## List of symbols

| | |
|---|---|
| AM | Anthropomorphic Model |
| B, C, D | coefficients of energetics curve fits |
| BW | body weight |
| $c_{\mathrm{COM}}$, $d_{\mathrm{COM}}$ | coefficients for $\delta_{\mathrm{COM}}$ curve fit |
| CI | confidence interval |
| COM | center of mass |
| $\dot{E}_{\mathrm{met}}$ | average metabolic rate |
| FM | Forward-foot Model |
| $\boldsymbol{g}$ | gravitational acceleration |
| GRF | ground reaction force |
| KM | Kneed Model |
| $L$ | leg length |
| $M$ | total mass |
| SM | Simplest Model |
| $\dot{V}_{\mathrm{CO_2}}$, $\dot{V}_{\mathrm{O_2}}$ | rate of carbon dioxide production/oxygen consumption |
| $v_{\mathrm{mid}}$, $v_{\mathrm{post}}$, $v_{\mathrm{pre}}$ | mid-, post-, pre-transition velocity |
| $W^{-}$ | negative work |

| | |
|---|---|
| $W^{+}$ | positive work |
| $\dot{W}^{-}_{\mathrm{DS}}$ | average double support negative COM work rate |
| $W^{-}_{\mathrm{mech}}$ | total negative work performed during one step |
| $\dot{W}^{-}_{\mathrm{mech}}$ | average negative COM work rate |
| $\dot{W}_{\mathrm{raise}}$ | average rate of work performed to raise the COM |
| $\alpha$ | leg angle (model) |
| $\delta$ | COM angular direction change (model) |
| $\delta_{\mathrm{COM}}$ | COM angular direction change (human) |
| $\rho$ | radius of curvature |

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