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RESEARCH ARTICLE

Mechanical and energetic consequences of rolling foot shape in human walking

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SUMMARY

During human walking, the center of pressure under the foot progresses forward smoothly during each step, creating a wheel-like motion between the leg and the ground. This rolling motion might appear to aid walking economy, but the mechanisms that may lead to such a benefit are unclear, as the leg is not literally a wheel. We propose that there is indeed a benefit, but less from rolling than from smoother transitions between pendulum-like stance legs. The velocity of the body center of mass (COM) must be redirected in that transition, and a longer foot reduces the work required for the redirection. Here we develop a dynamic walking model that predicts different effects from altering foot length as opposed to foot radius, and test it by attaching rigid, arc-like foot bottoms to humans walking with fixed ankles. The model suggests that smooth rolling is relatively insensitive to arc radius, whereas work for the step-to-step transition decreases approximately quadratically with foot length. We measured the separate effects of arc-foot length and radius on COM velocity fluctuations, work performed by the legs and metabolic cost. Experimental data (*N*=8) show that foot length indeed has much greater effect on both the mechanical work of the step-to-step transition (23% variation, *P*=0.04) and the overall energetic cost of walking (6%, *P*=0.03) than foot radius (no significant effect, *P*>0.05). We found the minimum metabolic energy cost for an arc foot length of approximately 29% of leg length, roughly comparable to human foot length. Our results suggest that the foot's apparently wheel-like action derives less benefit from rolling *per se* than from reduced work to redirect the body COM.

Key words: metabolic energy, locomotion, biomechanics, rocker bottom foot, arc foot, round foot, foot length, fixed ankle, rigid ankle, rollover shape.

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INTRODUCTION

The human ankle and foot act similar to a rolling wheel during each stance phase. This is evident both from the progression of the center of pressure upon the ground (McGeer, 1990), and from the arc traced by successive ground contact points when viewed from the shank's perspective (Hansen and Childress, 2004). Both are consistent with a wheel radius of approximately 30% of leg length. Although rolling presents clear benefits to the wheel, it is less obvious why this might benefit a leg, which otherwise behaves much like an inverted pendulum. After all, a pendulum conserves mechanical energy regardless of whether it rolls upon a curved surface or about a fixed pivot point. Perhaps the wheel-like motion has other consequences for walking, separate from rolling itself. Those consequences, if explained, might reveal fundamental mechanisms of human walking.

The human foot's wheel-like motion is facilitated by its plantigrade posture. An alternative configuration would place the foot more vertically, in line with and extending the leg. A longer leg traverses greater distance for the same angular excursion of the leg, potentially improving locomotion economy or speed. Indeed, digitigrade and unguligrade foot postures appear more prevalent in nature (Alexander, 1990). If there is no benefit from its rolling motion, there remains the question of whether the human foot's plantigrade posture has any advantage, or is even a disadvantage, for walking economy.

One potential benefit to wheel-like behavior other than rolling is for the transition between steps. The body center of mass (COM) follows a curved path atop the stance leg, and its velocity must be redirected from forward-and-downward at the end of one pendulum-like step to forward-and-upward at the beginning of the next step (Kuo et al., 2005). This redirection requires mechanical work, and exacts a metabolic cost (Donelan et al., 2002a). Simple models show that arc-shaped, rolling feet reduce the directional change in COM velocity, and thus the work of the step-to-step transition (Adamczyk et al., 2006; McGeer, 1990), which appears to have a proportional metabolic cost (Donelan et al., 2002a). Moreover, experiments with humans wearing curved, rigid foot attachments fixed to the leg show that both work and energetic cost change systematically with the arc radius (Adamczyk et al., 2006). Very small arcs are costly due to increased work to redirect the COM velocity, and this cost decreases with greater arc radii.

Closer examination of these models reveals how larger arc feet gain their advantage. The amount of COM redirection, and hence the step-to-step transition work, increases with the relative distance between the ground contact points of the two feet (see Fig. 1A). For a fixed angular excursion of the legs, this distance may be reduced with longer feet in a plantigrade posture. As long as the pendulum is allowed to move freely, the details of the foot's curvature are otherwise immaterial (Ruina et al., 2005). Thus, arc feet of greater radius may be economical primarily because they are longer, and plantigrade feet need not present an energetic disadvantage relative to digitigrade or unguligrade feet.

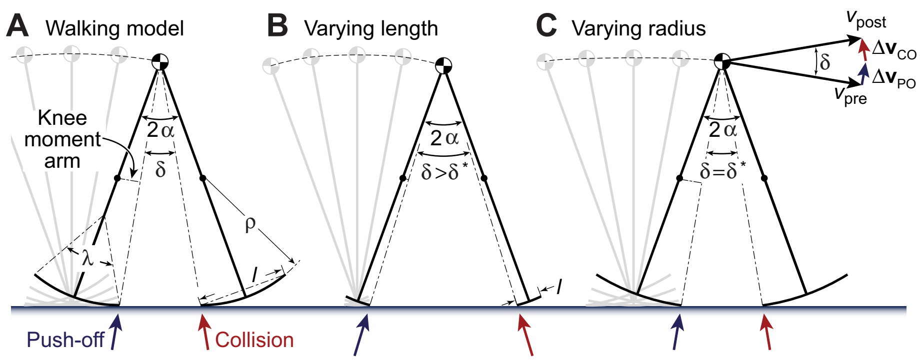

Fig. 1. The effect of varying radius and length of an arc-shaped foot, in a simple walking model. (A) The model has radius ρ and length l, and performs positive work with a push-off impulse P and negative work with a collision impulse C at the step-to-step transition, to redirect the COM velocity. The amount of redirection (represented by angle δ) is determined by the distance between the points of ground contact for the two arcs. More work is needed to perform greater amounts of redirection. (B) Varying foot length, keeping radius fixed, greatly affects the amount by which the velocity must be redirected. (C) In contrast, varying arc radius, keeping foot length fixed, has substantial effect on the COM trajectory, but no effect on redirection. The velocity change due to push-off and collision are shown as $\Delta \mathbf{v}_{PO}$ and $\Delta \mathbf{v}_{CO}$, respectively. The step length, kept constant, determines the angle between the legs, 2α. Keeping step length constant across conditions, the redirection angle is at a minimum (δ^*) when the feet are long enough to roll without pivoting on the heel and toe. The moment arm of GRF about the knee, typically small in human walking, is affected strongly by foot length but only slightly by radius; this effect may be partly responsible for the increased cost caused by long feet of large radius (Adamczyk et al., 2006).

Previous studies have not considered foot length independently of radius. We previously varied the two together parametrically (Adamczyk et al., 2006), and others have varied the radius while keeping length constant (with ankles not fixed) (Wang and Hansen, 2010). But to discern the distinct effects of rolling *versus* COM redirection, foot radius and length should ideally be treated as separate independent variables. For a given foot length, the foot's radius may be increased arbitrarily, which should affect rolling without affecting the step-to-step transition (Fig. 1B). Similarly, foot length may be varied for a given foot radius (Fig. 1C). We expect that foot length will have relatively greater effect on step-to-step transition work, total COM work and energetic cost than will foot radius.

There are, of course, other contributors to energetic cost in humans. For example, we have observed metabolic cost to increase for arc radii ρ greater than approximately 0.3 (expressed as a fraction of leg length) for arc feet with fixed ankles (Adamczyk et al., 2006). This is not explained by work requirements of simple dynamic walking models, or by COM work data, both of which steadily decrease. An increase in cost has also been observed in the free-ankle case, for radii above ρ=0.4 (Wang and Hansen, 2010). Additionally, our subjective experience with larger arcs revealed them to cause discomfort in the knee flexors after long bouts of walking (Adamczyk et al., 2006). We speculated that this was due to the need to actively avoid or compensate for a large external knee moment induced by the ground reaction force (GRF) acting on a very long foot (Fig. 1). Such a disadvantage would also be expected to depend more on foot length than radius. These and other factors are not necessarily predicted from first principles, but may nonetheless cause significant increases in energy cost for larger arcs.

The purpose of the present study was to test for separate effects of varying arc-foot length and radius. For simplicity, we considered walking with the ankles fixed, because it is difficult to predict the compensations that humans might perform with the ankles left free (Wang and Hansen, 2010). We hypothesized that, for a fixed foot length, variations in radius of curvature would have relatively little effect on mechanical work for the step-to-step transition and metabolic cost. For a fixed foot radius, we hypothesized that an increasing foot length would result in steadily decreasing work. We further hypothesized that metabolic cost would follow the same trend, but only to a point, beyond which we expected metabolic cost to increase due to an increasing external knee moment. These predictions were then tested with human subjects walking on curved foot bottom surfaces of varying length and radius. Such an experiment may thus separate the possible effects of rolling *versus* redirection of the COM.

MATERIALS AND METHODS

We experimentally tested a range of arc shapes attached to subjects' feet, and observed the effect on walking. We used a simple boot apparatus to fix subjects' ankles in a neutral position, restricting their dynamic action and allowing the attachment of different static shapes on the foot bottom. We measured GRFs, body motion and metabolic rate while subjects walked on an instrumented treadmill wearing different foot shapes. We estimated the work performed on the body COM by each leg, as well as the metabolic cost of walking, in response to independent changes in foot length and radius. We also investigated changes in joint mechanics to understand human adaptations to changing rollover shape. Before describing the experiments in more detail, we use a dynamic walking model to predict the effects of changes to foot length and radius.

Model

A simple, dynamic walking model explains the influence of foot length and radius on step-to-step transitions (Fig. 1). This model is very similar to a previous model with arc-shaped feet (Adamczyk et al., 2006), based on the 'simplest model' of walking (Kuo, 2001) (Fig. 1A). The model has a point mass at the pelvis and infinitesimally small point masses at the end of each leg (Fig. 1A). Arc-shaped feet of radius ρ (as a fraction of leg length) are rigidly attached to the leg. The present model limits the fore-aft foot length, l (also as a fraction of leg length), and hence the range of center of pressure excursion. At the limit of this range, the foot pivots about its heel or toe (Fig. 1B,C). Here we treat foot length l and radius ρ independently, revealing that work requirements decrease with longer feet, and are almost completely independent of changes in radius.

The model is powered by an instantaneous push-off impulse applied under the stance foot just before contralateral heelstrike (Fig. 1A) (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^-. The push-off and collision together constitute the step-to-step transition, and have equal magnitudes of work for a steady gait. Push-off and heelstrike impulses are directed from the ground contact points to the COM. The push-off impulse redirects the COM from its pre-transition velocity $\mathbf{v}_{\mathrm{pre}}$ to a mid-transition velocity $\mathbf{v}_{\mathrm{mid}}$; the heelstrike impulse then redirects the COM to a post-transition velocity $\mathbf{v}_{\mathrm{post}}$. A foot of nonzero length $l=2\rho\sin(\lambda/2)$ (where λ is the angle subtended by the foot; see Fig. 1A) reduces the directional change in COM velocity and the work performed to redirect the COM (Fig. 1A,B). Over the step-to-step transition the pre-to-post angular direction change δ in COM velocity is less than the angle between the legs, 2α. A periodic gait is produced (Kuo, 2002) if this net directional change is shared equally between the push-off and collision impulses (Fig. 1C). From the geometry of these impulses, we formulate the angular redirection of COM velocity and form a small-angle approximation:

$$\tan\frac{\delta}{2} = \frac{\rho\sin\left(\alpha-\frac{\lambda}{2}\right)+\left(1-\rho\right)\sin\alpha}{\rho\cos\left(\alpha-\frac{\lambda}{2}\right)+\left(1-\rho\right)\cos\alpha} \approx \alpha - \frac{l}{2}. \quad (1)$$

Note that the foot length term subsumes the effect of arc radius ρ in this linear approximation.

The effects on step-to-step transition work are as follows. The magnitude W^- of the negative work performed each step by the heelstrike collision is equal to the change in kinetic energy (Fig. 1C):

$$W^- = \frac{1}{2}Mv_{\mathrm{mid}}^2 - \frac{1}{2}Mv_{\mathrm{post}}^2 = \frac{1}{2}Mv_{\mathrm{post}}^2\tan^2\frac{\delta}{2}. \quad (2)$$

The overall trend is revealed by substituting Eqn 1 into Eqn 2:

$$W^- \approx \frac{1}{2}Mv_{\mathrm{post}}^2\left(\alpha-\frac{l}{2}\right)^2. \quad (3)$$

The model therefore predicts the trends in COM velocity change and step-to-step transition work as a function of foot length l.

Keeping step length fixed, the step-to-step transition leg angle α is nearly constant over the range of l and ρ applied in our experiment. Again assuming small angles, Eqn 1 reduces to show that the angular direction change δ in COM velocity decreases approximately linearly with foot length l, with a constant offset C_δ:

$$\delta \propto (C_\delta - l). \quad (4)$$

Keeping walking speed fixed, the post-transition velocity v_{post} is also approximately constant. Thus the trend in the magnitude of negative COM work performed simplifies to a similar form:

$$W^- \propto (C_{\mathrm{W}} - l)^2, \quad (5)$$

where C_{W} is the foot length at minimum negative COM work. For a steady gait, $W^-=W^+$, allowing Eqn 5 to predict the trend for positive COM work as well. This prediction forms the basis for comparisons with measured data.

We used computational simulations to test the analytical predictions with a more human-like mass distribution. The 'anthropomorphic model' has roughly human-like leg mass and inertia (after McGeer, 1990), as well as a spring about the hip joint in order to produce human-like step frequencies (Kuo, 2001). We examined the model's gait across variations in foot length l and radius ρ, keeping speed, step length and all other model parameters except spring stiffness fixed. We limited our model investigation to foot lengths short enough to lead to pivoting about the heel or toe, beyond which the model is identical to our previous model (Adamczyk et al., 2006). The model exhibits a consistent decrease in work (i.e. energy cost) with increasing foot length l (Fig. 2). For

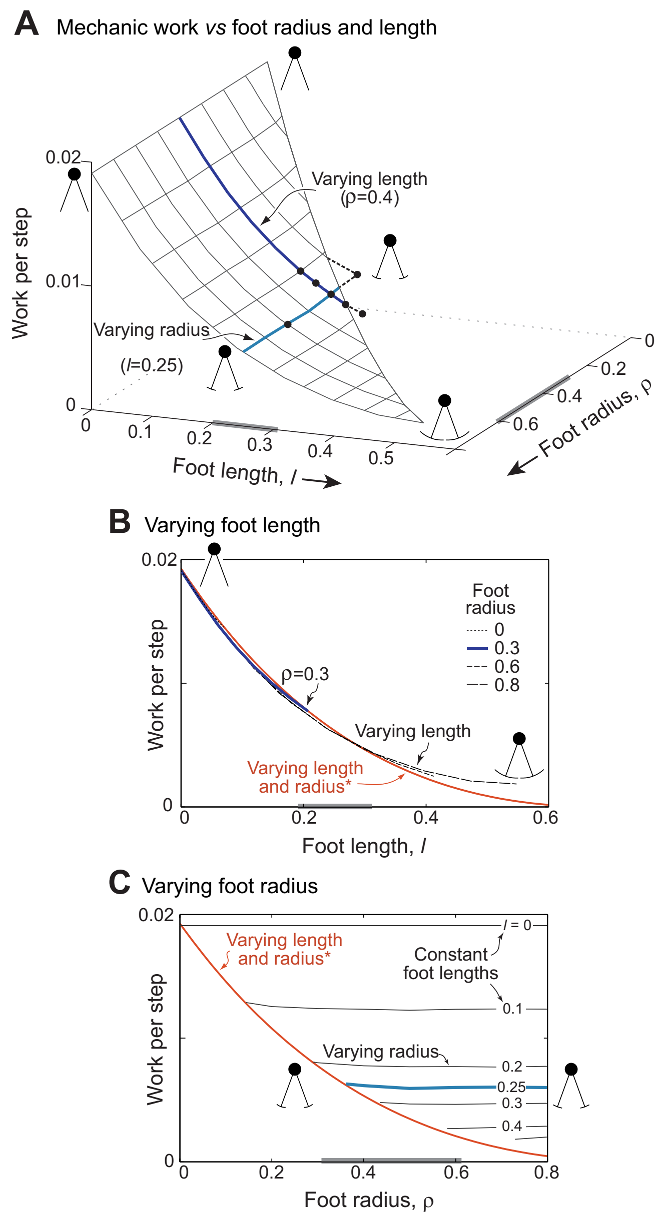

Fig. 2. Dynamic walking model predictions for step-to-step transition work as a function of foot radius and length. (A) Step-to-step transition work per step (vertical axis) depends on both arc radius and length, the latter having a much greater effect. (B) Work decreases nearly quadratically as a function of foot length, as demonstrated for various foot radii. (C) In contrast, work per step varies little with foot radius, as shown for various foot lengths. The approximate range of experimental conditions is denoted by the shaded area on the horizontal axes. Filled circles in A show the parameters studied experimentally (see Fig. 3). In B and C, the boundary of the parameter space for the current model shows results that agree closely with a previous model, in which length and radius varied together (Adamczyk et al., 2006).

a constant foot radius ρ, the model closely follows the curve of Eqn 5 (Fig. 2B) as foot length increases up to the maximum limit imposed by ρ, which it cannot exceed. But there is no substantial change with varying foot radius at constant foot length (Fig. 2C).

We expected that work by humans would show trends similar to the model. Despite the much greater complexity of humans, we expected COM work to decrease with increasing foot length, but change little with increasing radius. We also expected that metabolic energy expenditure would largely follow the same trend as COM work. Of course, there are metabolic costs beyond the work performed on the COM. Based on our speculation regarding the external knee moment, we expected to observe a minimum in metabolic cost for a foot of intermediate length, beyond which cost would increase. We expected no trend in metabolic cost *versus* foot radius of curvature for the range tested.

Experiment

We measured GRFs, body motion and respiratory gas exchange while eight adult human subjects walked in rigid boots with soles of different length and curvature. Walking speed was fixed at 1.275 m s^{-1} using a split-belt force-measuring treadmill. All subjects (four male, four female; mean ± s.d. body mass 72.8±12.5 kg; leg length 0.904±0.062 m) were healthy and had no known gait abnormalities. Subjects gave their written informed consent according to Institutional Review Board procedures.

The experimental apparatus consisted of a pair of rigid walking boots (PneumaticWalker, Aircast, Summit, NJ, USA) modified to accept interchangeable bottom surfaces (Fig. 3) (see Adamczyk et al., 2006). Seven pairs of these foot arcs were constructed from pine wood (0.086 m wide) and covered on the bottom surface with SoleFlex shoe sole material (0.0015 m thick; SoleTech, Salem, MA, USA). Five pairs had a foot radius of 0.40 m, with different foot lengths (0.203, 0.229, 0.254, 0.279 and 0.305 m, measured heel to toe). Two additional pairs had a foot length of 0.254 m, with different foot radii (0.30 and 0.60 m). Arcs were matched in mass (0.45 kg) and standing height (0.037 m), although moment of inertia could not be precisely matched. Although fewer than length conditions, the three radius conditions nevertheless span a twofold range of radii, sufficient to test the hypothesized sensitivity. All arcs were attached to the same pair of boots with the same alignment for a given subject (each 0.85 kg for medium size, 1.05 kg for large). Arcs were positioned relative to the leg so that the arc center was 0.058 m anterior to the tibial axis (Fig. 3). This dimension is slightly less than in our previous study (Adamczyk et al., 2006) because of the different geometry of the arcs.

Subjects walked at 1.275 m s^{-1} on a custom-built split-belt instrumented treadmill (Collins et al., 2009) wearing each pair of arcs, and also in normal street shoes (termed normal walking), with the order of arc conditions randomized for each subject. We did not control step frequency, which was observed to range from 94% to 100% of normal as a function of arc shape here. We assumed that this variation did not substantially affect metabolic cost for forced motion of the legs (Doke et al., 2005).

We measured GRFs (Fig. 4) and lower body motion for each condition. We recorded multiple 30 s trials of GRF and motion capture data to ensure recording of several clean ground contact periods of each foot on a single side of the treadmill at steady state. We analyzed the first seven clean strides in MATLAB (The MathWorks, Natick, MA, USA). The GRF data were used 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 kinematics (linear acceleration, velocity and position) from three-dimensional GRF data, assuming periodic gait (Whittle, 1997; Donelan et al., 2002b). The velocity data were then used to derive the maximum angular change δ_{COM} in the direction of COM velocity in the sagittal plane (Fig. 1, Fig. 6A,B) (Adamczyk et al., 2006). The forward and vertical COM velocities were also plotted against each other as the COM hodograph (Adamczyk and Kuo, 2009) for each condition (Fig. 5A,B). The instantaneous rate of mechanical work performed by each leg on the COM (Fig. 5C,D) was calculated as the dot product of each leg's GRF with the COM velocity (Donelan et al., 2002b). We integrated the negative portion of this COM work rate to find the total COM work W^{-}_{mech} (J) performed during one step. Finally, we multiplied this work by step frequency to yield the average COM work rate $\dot{W}^{-}_{mech}$ (W) performed on the COM through each limb (Fig. 6C,D).

We estimated metabolic energy expenditure rate from respiratory gas exchange data collected during the treadmill trials (Fig. 6E,F). We used an open-circuit respirometry system (Max-II, Physio-Dyne Instrument Corporation, Quogue, NY, USA) to measure the volume rates of oxygen consumption and carbon dioxide production ($\dot{V}_{O_2}$ and $\dot{V}_{CO_2}$, ml s^{-1}). Following a 3-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 $\dot{E}_{met}$ was estimated using standard formulas (Adamczyk et al., 2006), after Brockway (Brockway, 1987) and Weir (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, but was performed before any other 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 foot length and arc radius. First, we performed a least-

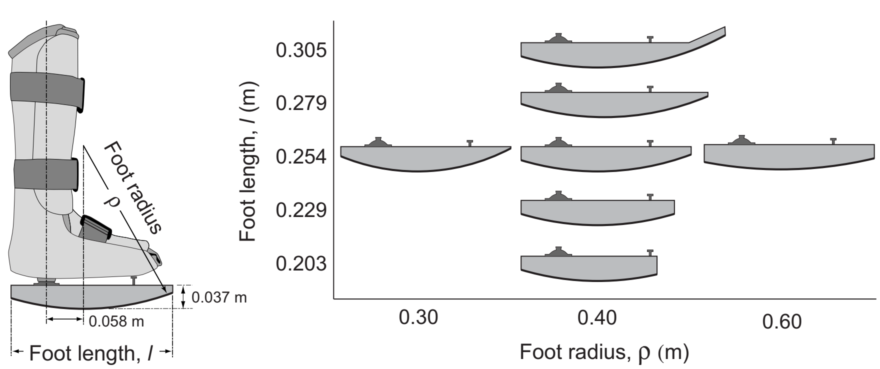

Fig. 3. Apparatus used to impose an arc-like shape to the foot bottom, while keeping the ankles fixed (see also Adamczyk et al., 2006). Arcs were constructed in five lengths and three radii of curvature.

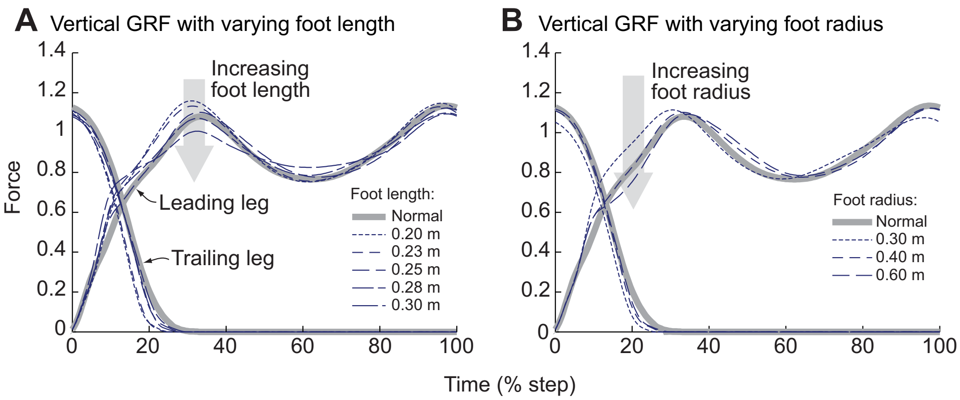

Fig. 4. Average ground reaction force data over a step cycle, for varying (A) foot length keeping radius fixed and (B) foot radius keeping length fixed. Curves are averaged across all (*N*=8) subjects. A step was defined as starting at one heelstrike and ending at opposite heelstrike.

squares fit to the model of Eqn 4, regressing COM velocity direction change δ_{COM} against foot length l according to:

$$\delta_{COM} = c_{COM} l + d_{COM} \, . \tag{6}$$

The coefficients c_{COM} and d_{COM} accommodate differences between humans and the model, such as knee flexion and duration of step-to-step transition, that can affect measured δ_{COM} (Adamczyk and Kuo, 2009).

We regressed the subjects' mechanical and metabolic costs against foot length (for conditions with constant arc radius 0.40 m) using a general second-order curve fit inspired by the model (Eqn 5):

$$\text{Curve fit: } a_l l^2 + b_l l + c_l \, . \tag{7}$$

We also regressed mechanical and metabolic costs against arc radius (for conditions with constant foot length 0.254 m) in the same manner, using coefficients a_ρ, b_ρ and c_ρ. We applied the same form of fit to both mechanical and metabolic costs, $\dot{W}^{-}_{mech}$ and $\dot{E}_{met}$, adding subscripts 'mech' and 'met', respectively, to distinguish the various coefficients.

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 g and barefoot standing leg length L. Work rate and energy rate were therefore made dimensionless by the divisor $Mg^{1.5}L^{0.5}$; work, energy and moment by MgL; and force by Mg. Foot length and arc radius were non-dimensionalized by L. Work rate and energy rate graphs are presented in both dimensionless units and in the common units of $\mathrm{W\,kg^{-1}}$. Conversion between these units was performed with the mean factor $g^{1.5}L^{0.5} \approx 29.2\,\mathrm{W\,kg^{-1}}$. We also accounted for inter-subject kinematic and energetic variations by computing curve fit offsets d_{COM}, c_l and c_ρ separately for each subject and then averaging them. Accompanying r^2 values were computed without these offsets, to quantify only the trends explained by the independent variable rather than the offsets.

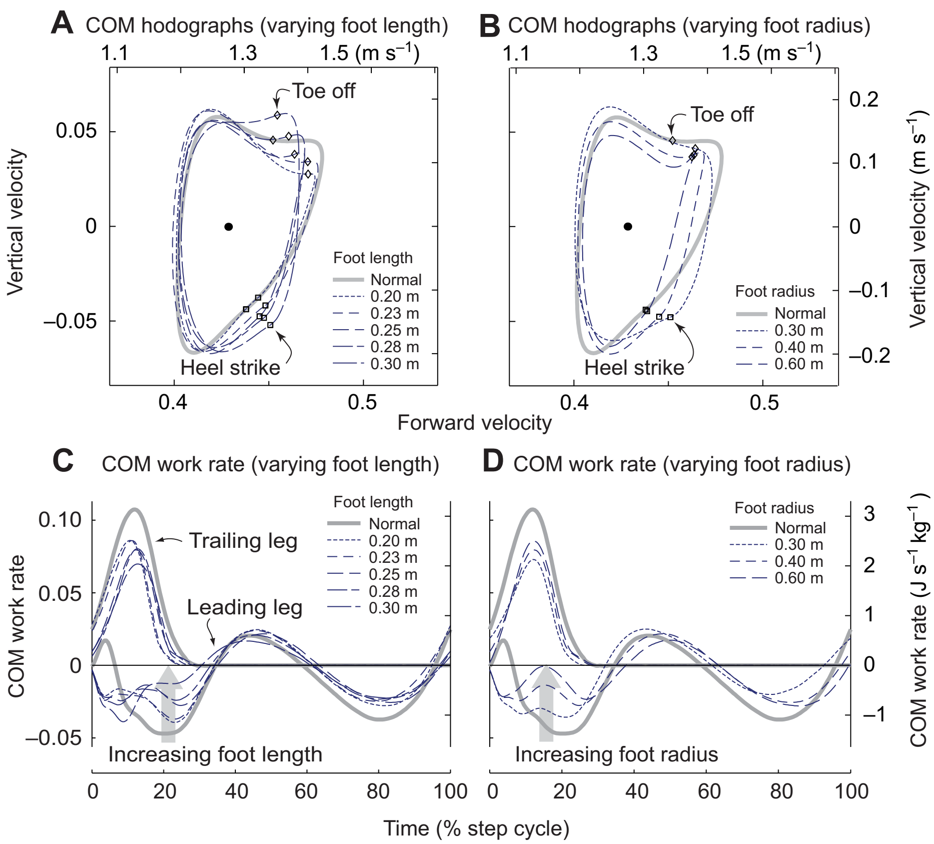

Fig. 5. (A,B) Vertical *versus* forward components of center of mass (COM) velocity data, termed COM hodographs, for arc feet of varying (A) length and (B) radius. (C,D) COM work rate *versus* step cycle for varying (C) foot length and (D) foot radius. Curves are shown for both legs (labeled as leading and trailing leg during step-to-step transition). All curves are averaged across subjects (*N*=8).

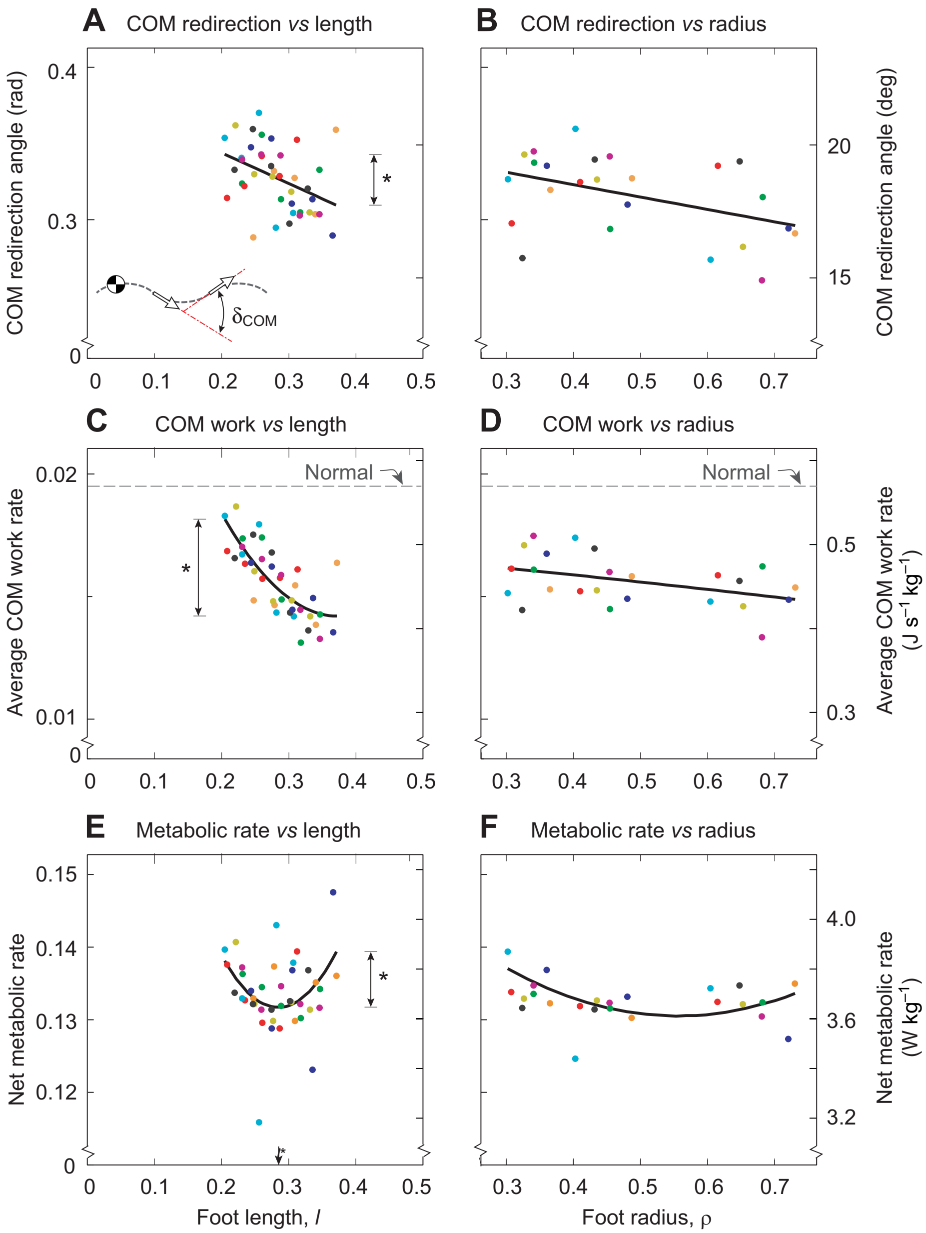

Fig. 6. Experimental results for varying foot length (left column) and foot radius (right column). (A,B) Average redirection angle of COM velocity (δ_{COM}) decreased significantly (**P*<0.05) as function of (A) foot length (*P*=0.03), but not (B) foot radius (*P*=0.08). Angular redirection of COM velocity is a strong predictor of COM work (Adamczyk and Kuo, 2006; Adamczyk et al., 2009). (C,D) Average COM work rate changed as predicted by the model, decreasing significantly across (C) foot length (*P*=0.04 for linear term) but not (D) foot radius (*P*=0.33). Average COM work was considerably lower than normal in all cases. (E,F) Net metabolic rate also changed significantly with (E) foot length but not (F) foot radius. Quadratic curve fits (solid lines) yield minima at l=0.285 for foot length (*P*=0.03) and ρ=0.553 for foot radius (*P*=0.31). All trials from all subjects (*N*=8, filled symbols) are shown along with curve fits (solid dark lines) to equations predicted by the model.

RESULTS

The mechanics and energetics of walking changed significantly as a function of foot length, and to a much smaller degree with arc radius. The average rate of negative mechanical work performed on the COM decreased significantly with increasing foot length, but not with increasing radius. Net metabolic rate was at a minimum at intermediate foot length (curve fit minimum at l=0.285), but was relatively unaffected by radius. GRFs were affected systematically by both foot length and arc radius, but in different ways. Finally, the angular direction change in COM velocity over the step-to-step transition decreased with increases in both foot length and foot radius, although the effect was much smaller for arc radius. Results for GRFs, COM velocity direction change, COM work rate and metabolic rate during normal walking and walking with arcs are detailed below.

Baseline measures of mechanical work rate and metabolic rate of normal walking were as follows. In normal walking at 1.275 m s^{-1} with preferred step frequency 1.86±0.09 Hz, the angular direction change δ_{COM} in COM velocity was 17.3±2.6 deg [mean ± 95% confidence interval (CI)]. Subjects performed negative COM work $\dot{W}^{-}_{mech}$ at an average rate of 0.543 W kg^{-1} (dimensionless 0.019). This is equivalent to 0.291 J kg^{-1} per step, which is slightly lower than estimates of 0.31 to 0.36 J kg^{-1} per step from previous studies (Donelan et al., 2002a; Donelan et al., 2002b), possibly because of differences between over-ground and treadmill walking during mechanics trials. Average net metabolic rate $\dot{E}_{met}$ for normal walking was 2.96 W kg^{-1} (dimensionless 0.101).

Measured vertical GRFs changed in qualitatively different ways with alterations in foot length and radius (Fig. 4A,B). Foot length had important effects on peak values and timing, whereas radius appeared to alter the shape of the GRF curve more subtly. With increasing foot length at constant radius (ρ=0.40 m), a linear regression shows that the double-support period increased with greater length, expanding from approximately 16% to 20% of the step, over dimensionless foot lengths from 0.20 to 0.37 (*P*=2e–11; curve fit $-0.24l$+0.11, r^2=0.77). Vertical GRF peaks also declined with longer feet, from 1.21 to 1.02 (dimensionless; *P*=7e-10; curve fit $-1.19 \cdot l + 1.46$, r^2=0.71) for the first peak, and from 1.18 to 1.09

(P=3e–5; curve fit $-0.56l+1.30$, r^2=0.44) for the second peak. In addition, the mid-stance force minimum became shallower with increasing foot length, rising from 0.70 to 0.79 (P=4e–7; curve fit $-0.56l+0.59$, r^2=0.57). With increasing radius (at constant foot length l=0.254 m), the only change was a negligible increase in double-support period, from 18% to 19% of the step (P=2e–3). The initial rise of the force curve appeared qualitatively to have a steeper slope than normal, perhaps because of the relative rigidity of the boot-arc apparatus compared with a normal foot and ankle. The first peak in vertical GRF occurred with roughly normal timing. The vertical GRF most similar to normal walking occurred with arc feet of length 0.254 m and radius 0.40 m.

The fluctuations in COM velocity also changed in different ways with foot length and radius, described here qualitatively through COM hodographs (Fig. 5A,B). With increasing foot length, the hodograph showed overall smaller changes in forward COM velocity over each step (Fig. 5A). The shape of the hodograph also changed, with smoother COM velocity (i.e. less sharp inflections) near toe-off for longer feet. Increasing foot radius (Fig. 5B) also led to reduced forward COM velocity fluctuations, but changed the shape differently.

The relative distribution of COM work throughout the step also changed with foot length (Fig. 5C,D), also examined here qualitatively. We define the collision as the first region of negative COM work in a step, and push-off as the first region of positive work starting near the end of the preceding step and extending through double support (Kuo et al., 2005). Qualitatively, collision negative work by the leading leg tended to decrease with increasing foot length l, with late collision work decreasing such that the collision period ended earlier with longer feet (Fig. 5C). Push-off positive work rate by the trailing leg also tended to decrease and shift later with increasing foot length l, overlapping more with the late collision COM work in the leading leg. Subjects performed about the same amount of work during push-off and during collision, with little apparent variation in single-support positive and negative work in the stance leg. Collision negative work also tended to decrease with increasing foot radius (Fig. 5D). In this case, however, the work was merely shifted to the later ‘preload’ phase of negative work. Positive push-off work increased with increasing foot radius, the opposite effect from increasing foot length.

The observed angular direction change in COM velocity δ_{COM} decreased with increasing foot length and radius (Fig. 6A,B). For varying foot length l and constant arc radius ρ=0.40 m, data exhibited roughly linear trends (P=0.03, r^2=0.14; Fig. 6A), with coefficients $c_{l\text{-COM}}$=–11.5±10.3 deg (mean ± 95% CI) and $d_{l\text{-COM}}$=22.0±3.1 deg (Eqn 6). The COM direction change for normal walking intersected with the observed trend at a foot length of approximately l=0.38. The trend with increasing arc radius ρ at constant foot length l=0.254 m was much shallower, with no significant trend (P=0.08, r^2=0.19; Fig. 6B). The best-fit line had coefficients $c_{\rho\text{-COM}}$=–4.7±5.3 deg (mean ± 95% CI) and $d_{\rho\text{-COM}}$=20.7±3.3 deg (Fig. 6B). Angular direction change was comparable to normal walking at an arc radius of approximately ρ=0.65.

The total amount of negative COM work performed ($\dot{W}^{-}_{\mathrm{mech}}$) agreed well with the decreasing trend across foot length predicted by the dynamic walking model (Fig. 6C, Eqn 5), and did not change significantly across arc radius. The curve fit (Eqn 7, P=0.04 for linear term, r^2=0.57) showed a significant decline in overall negative COM work rate as foot length l increased from 0.20 to 0.37 (Fig. 6C). The coefficients of the curve fit are $a_{l\text{-mech}}$=0.140±0.169 (mean ± 95% CI, dimensionless), $b_{l\text{-mech}}$=–0.104±0.097 and $c_{l\text{-mech}}$=0.032±0.014. This curve fit has a minimum at l=0.373, though this value is tentative because the quadratic term was not different from zero (P=0.1). The slight decrease in negative work rate with increasing arc radius of curvature ρ (Fig. 6D) was not statistically significant (P=0.33).

Metabolic energy expenditure rate $\dot{E}_{\mathrm{met}}$ also exhibited a minimum at intermediate foot length, as expected (Fig. 6E,F, Eqn 5). The empirical quadratic curve fit (Eqn 7, P=0.03, r^2=0.15) showed a significant decline, with coefficients $a_{l\text{-met}}$=1.009±0.894 (mean ± 95% CI, dimensionless), $b_{l\text{-met}}$=–0.574±0.511 and $c_{l\text{-met}}$=0.213±0.072. The minimum suggested by this curve is 0.132 (3.84 W kg^{-1}) at l=0.285 (Fig. 6E). Metabolic rate did not change significantly with increasing arc radius of curvature ρ (P=0.31, r^2=0.12), yielding a curve fit with coefficients $a_{\rho\text{-met}}$=0.102±0.206 (mean ± 95% CI, dimensionless), $b_{\rho\text{-met}}$=–0.113±0.212 and $c_{\rho\text{-met}}$=0.162±0.051. This curve fit suggests a minimum cost of 0.130 (3.81 W kg^{-1}) at ρ=0.553 (Fig. 6F).

DISCUSSION

We have investigated the differing effects of foot length l and foot radius ρ on the mechanical and metabolic costs of walking. Our model and experimental human data demonstrate that foot length has a greater effect on walking, as quantified by GRF peaks, COM velocity fluctuations and COM work rate. Varying foot radius over a twofold range produces some changes in rolling mechanics of the stance phase, but has far less of an effect on work and energetic cost. Prior studies have focused more on the radius, either as a covariate of length in arc feet (Adamczyk et al., 2006), or in terms of an effective rollover shape for the foot–ankle complex (Hansen and Childress, 2004; Hansen and Childress, 2005; Hansen et al., 2004; Wang and Hansen, 2010). The present results show that arc foot length has a greater effect on gait mechanics than foot radius.

Arc foot length is important because it facilitates the step-to-step transition. The work dissipated at collision is determined by the change in COM velocity from the end of one step to the beginning of the next one (Fig. 1B). A longer foot reduces the directional change in velocity (Fig. 6A), which is accompanied by a reduction in work (Fig. 6C). This is consistent with the simple model introduced here and, in a different form, by Ruina et al. (Ruina et al., 2005). We had previously found that energy expenditure is greatly affected when length and radius vary together (Adamczyk et al., 2006). A comparison with the present results (Fig. 7) indicates that similar effects are observed when varying arc length alone, but not radius. Indeed, the curve fit of metabolic rate *versus* foot length reached a minimum metabolic cost for l=0.285, similar to that observed previously [0.300 (Adamczyk et al., 2006)] and to the ratio of human foot length to leg length (Dumas et al., 2007).

Arc foot radius has a comparatively subtler effect on walking. Smaller radius feet do appear to trend toward an increasing cost, and there is a minimum radius required to achieve a convex foot bottom while maintaining a given arc foot length. At the other extreme, a very high radius of curvature – essentially a flat foot bottom – may be quite unfavorable. If rigid, such a shape might cause the center of pressure to advance very rapidly, requiring compensatory forces by the muscles to be produced at a high rate and magnitude. It appears most advantageous for the foot bottom to have a curved, convex shape, with less sensitivity to the particular curvature.

It is curious that the minimum metabolic cost found here was approximately 30% higher than the cost of normal walking (Fig. 7B), despite the fact that all arc conditions required less COM work than normal. Much of this increase might be explained by the 1.5 kg added

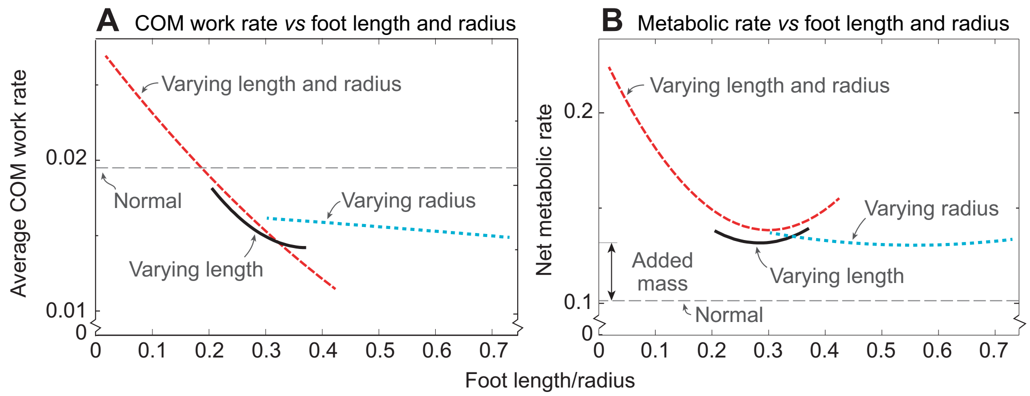

Fig. 7. Comparison of best-fit quadratic curves for (A) COM work rate and (B) metabolic rate *versus* foot length (solid black lines) and foot radius (dotted cyan lines) from the present study and from a previous experiment varying length and radius together (dashed red lines) (Adamczyk et al., 2006). The close match between the current 'varying length' curve and the previous result emphasizes that mechanical and metabolic cost changes attributable to arc foot shape are primarily due to changes in foot length. The 'varying radius' curve from the present study departs substantially from the previous result, suggesting that foot radius *per se* was not responsible for the changes observed in that study (Adamczyk et al., 2006). Note that length and radius share a common axis in the previous study (dashed red lines) because geometric design constraints made the two parameters numerically equal within a reasonable small-angle approximation.

mass of each arc foot. An estimate of the energetic penalty due to adding mass below the feet also yields about a 30% increase (e.g. Inman et al., 1981; Martin et al., 1997; Royer and Martin, 2005; see Adamczyk et al., 2006). Indeed, when normal shoes are weight-matched to rigid arc foot boots, there is no difference in energy expenditure (Vanderpool et al., 2008). Another possible explanation for the energetic penalty is work performed by the knee and hip to substitute for the ankle. If that work were performed less economically, it may explain the added cost. Other possible costs include relatively poor coordination with walking on arc-shaped feet, as subjects may not have fully adapted to their added mass, restricted ankle motion, smaller ground contact patch and rigid arcs. Although the added cost is not fully explained, we believe that it does not affect the relative effects we observed with changing foot length and radius.

More difficult to explain is the energetic penalty of longer arc feet. Based on previous experience, we had expected that metabolic cost would increase at high foot length, due to factors not attributable to COM work. We had proposed that the energetic increase may stem from compensations to counter or avoid the high external knee moment induced by longer feet late in stance (Adamczyk et al., 2006). After all, longer feet can cause the center of pressure to fall well in front of the knee late in stance. But qualitative inverse dynamics analysis (see the Appendix, Fig. A1, Fig. A2) does not indicate increasing knee moment or work late in stance (Fig. A1). It is possible that subjects managed to avoid a higher knee moment by making subtle gait adjustments that reduced the moment arm of GRF about the knee. If not for kinematic adjustments, the GRF would then induce a large external extension moment. The details of this adjustment remain unexplained by COM work, joint work or joint moments, but it is nonetheless clear that longer feet are energetically costly.

Inverse dynamics analysis offers qualitative support for the hypothesis that mechanical energy lost from push-off is restored elsewhere. The COM work measure does not capture work at the joints, but examination of inverse dynamics results demonstrates the consequence of reduced push-off work. With reduced ankle motion, the hip appeared to compensate with more work late in push-off. This was accompanied by a concurrent, increased burst of positive work by the contralateral hip, observed in all arc conditions (Fig. A1, Fig. A2), suggesting that it also compensates for the reduced push-off work. In some circumstances, adaptations shifting load or work to other joints may be less economical than normal gait (Sawicki et al., 2009), so this shift toward hip work may also contribute to the overall higher metabolic cost of all the arc foot conditions.

These results suggest a functional feature of the human plantigrade foot posture. Digitigrade or unguligrade feet might seem advantageous for locomotion economy, because such feet contribute to greater leg length. This allows for a greater step length and a lower cost of transport for a given angle between the legs at double support. A plantigrade foot would seem to sacrifice leg length and gain little in return. (There might be some advantage in stability for bipedal standing, but there is little evidence that humans are more adept at balance than, say, birds.) Our findings suggest that a plantigrade foot is not at a disadvantage, because it can decrease the angular redirection required of COM velocity. It allows the foot to apply a more vertical impulse at push-off, decreasing the energy dissipated in the leading leg collision (Adamczyk and Kuo, 2009). A plantigrade foot posture could be quite uneconomical, but the timing and amount of ankle push-off appear to make humans quite competitive with other animals of similar size (cf. Tucker, 1975).

It is interesting to consider how humans might benefit from the normal, articulated ankle. The active ankle tends to produce an effective foot rollover shape that agrees with the optimum rigid shape observed here, regardless of factors such as walking speed, incline and shoe geometry (Hansen and Childress, 2004; Hansen et al., 2004; Hansen and Childress, 2005; Wang and Hansen, 2010). That tendency makes it difficult to vary the shape experimentally, except by enforcing it with a rigid arc shape. Our results indicate that, however humans might produce the effective rollover shape, its nominal length and shape appear to minimize energy expenditure for the rest of the body. This is especially the case at slower speeds where the ankle produces little work (Hansen et al., 2004), making it most similar to the rigid case examined here. Another implication is the potential disadvantage of perturbations to the nominal effective rollover shape. As an example, prosthetic feet with shorter structural keels result in higher GRF peaks (Hansen et al., 2006), consistent with our model and potentially implying poorer walking economy. These findings may also be relevant to cases of partial foot amputation (Hansen, 2007). Returning to the normal case, it appears that the human foot uses its length and rollover shape to reduce the energetic penalty of redirecting the body COM between steps, making up for an apparent disadvantage in leg length relative to digitigrade or unguligrade feet.

APPENDIX

Inverse dynamics

We performed inverse dynamics analysis to estimate lower body mechanics during each condition. We attached reflective markers to the segments of the lower body: three markers to each thigh and

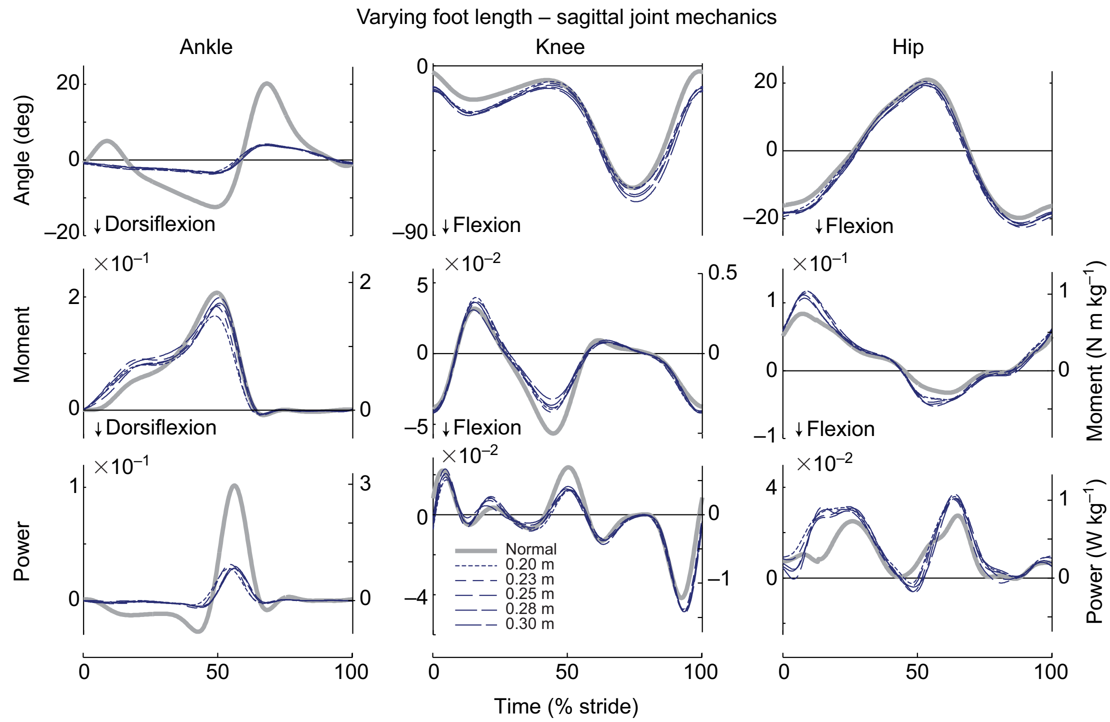

Fig. A1. Sagittal joint mechanics (angle, moment, power output) across variations in foot length. Curves are averaged across all eight subjects. Values are nondimensionalized on the left-hand axes; values on the right-hand axes units are normalized to body mass.

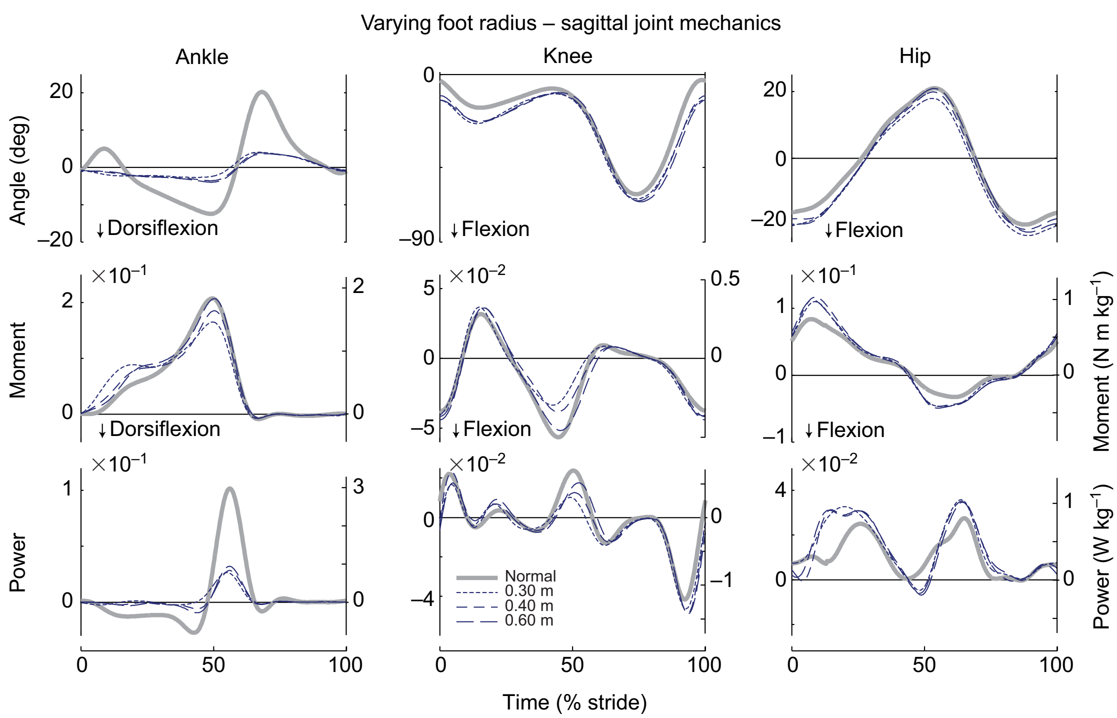

Fig. A2. Sagittal plane joint mechanics (angle, moment, power output) across variations in foot radius. Curves are averaged across all eight subjects. Note the trend toward increasing knee moment in late stance, which may reflect the altered timing of GRF center of pressure advancement due to changing foot radius. Values are nondimensionalized on the left-hand axes; values on the right-hand axes units are normalized to body mass.

shank, plus one superficial to each greater trochanter, lateral epicondyle, lateral malleolus, calcaneus and fifth metatarsal head, and one over the sacrum. We tracked these markers using an eight-camera system (Falcon) and software (EVaRT) from Motion Analysis Corporation (Santa Rosa, CA, USA). For trials using the fixed-ankle apparatus, all shank and foot markers were attached to the plastic boot, over the approximate locations of the underlying landmarks. We performed a standard inverse dynamics analysis using Visual3D software (C-Motion, Germantown, MD, USA) to estimate joint angle, moment and power output curves across the stride cycle, including additional mass and inertia for the boot and arc apparatus. Because separate models were required for the normal and arc-foot conditions, absolute joint angles may appear different artificially, though changes in angle over time should be accurate. Also, note that ankle mechanics in fixed-ankle conditions represent the combined effects of the physiological ankle and the apparatus.

Results for sagittal plane joint mechanics showed qualitative changes at the knee and hip with changes in both foot length (Fig. A1) and foot radius (Fig. A2). The most apparent trend was the emergence of a greater and more sustained knee internal flexion moment in late stance as foot radius increased (Fig. A2). With larger-radius arcs, this knee flexion moment was sustained into the period of pre-swing knee flexion displacement, with the result that the knee exhibited a considerable positive power peak during pre-swing as arc radius increased. In contrast, this same period of knee moment actually decreased with increasing foot length, though in that case there was no change in timing or knee power output because the concurrent knee displacement was very small. Finally, flexion moment and negative power output of the knee appeared to increase in magnitude during swing termination in all experimental conditions in comparison to normal. We interpret this increase as a simple effect of the added mass of the apparatus on the lower leg.

At the hip joint, sagittal moment and power peaks appeared amplified in all experimental conditions in comparison to normal, but with no clear trends across conditions. We broadly interpret this increase as an attempt to use the hip to replace some of the mechanical work lost to ankle fixation in the boots. Such a strategy is most strongly suggested by the extended period of positive hip power in early stance. The other peaks in late stance may also indicate the increased use of hip flexors required to accelerate the stance leg into swing.

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AUTHOR CONTRIBUTIONS

P.G.A. and A.D.K. conceived and designed the study. P.G.A. performed the simulations and experiments. P.G.A. and A.D.K. wrote the manuscript.

COMPETING INTERESTS

No competing interests declared.

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