Dynamic arm swinging in human walking

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Humans tend to swing their arms when they walk, a curious behaviour since the arms play no obvious role in bipedal gait. It might be costly to use muscles to swing the arms, and it is unclear whether potential benefits elsewhere in the body would justify such costs. To examine these costs and benefits, we developed a passive dynamic walking model with free-swinging arms. Even with no torques driving the arms or legs, the model produced walking gaits with arm swinging similar to humans. Passive gaits with arm phasing opposite to normal were also found, but these induced a much greater reaction moment from the ground, which could require muscular effort in humans. We therefore hypothesized that the reduction of this moment may explain the physiological benefit of arm swinging. Experimental measurements of humans ($n = 10$) showed that normal arm swinging required minimal shoulder torque, while volitionally holding the arms still required 12 per cent more metabolic energy. Among measures of gait mechanics, vertical ground reaction moment was most affected by arm swinging and increased by 63 per cent without it. Walking with opposite-to-normal arm phasing required minimal shoulder effort but magnified the ground reaction moment, causing metabolic rate to increase by 26 per cent. Passive dynamics appear to make arm swinging easy, while indirect benefits from reduced vertical moments make it worthwhile overall.

Keywords: locomotion; upper extremities; biomechanics; energetics; simulation

1. INTRODUCTION

It is curious that humans swing their arms as they walk, because the arms play no obvious role in locomotion. The shoulder muscles contribute to active swinging (Fernandez-Ballesteros *et al.* 1965), indicating that some metabolic energy is expended in the process. This cost may, however, be countered by other effects for an overall benefit. A number of beneficial effects of arm swinging have been hypothesized, for example reduced vertical excursion of the centre of mass (Murray *et al.* 1967; Hinrichs 1990; Umberger 2008), improved mechanical stability (Ortega *et al.* 2008) or reduced vertical ground reaction moments (Witte *et al.* 1991; Li *et al.* 2001). These and other effects have each been proposed in separate studies, making it difficult to compare their relative merits directly. Multiple effects could, however, be compared in computational models that simulate the dynamic interactions between the arms and the rest of the body. The costs and benefits of arm swinging might then be examined by combining such models with human subject experiments that treat arm swinging as a controlled variable.

The muscles appear to exert some amount of effort to swing the arms. Biomechanists initially speculated that the arms might swing purely as a result of the movements of the shoulders during gait, behaving as passive pendulums (Gerdy 1829; Weber & Weber 1836; Morton & Fuller 1952). Other measurements indicated significant

shoulder torques (Elftman 1939), although with reported peaks varying as much as three-fold (e.g. 3.8 N m reported by Jackson *et al.* 1978; 12 N m by Hinrichs 1990). Muscular effort is also suggested by observations of significant electromyographic activity, which led Fernandez-Ballesteros *et al.* (1965) to propose that muscular forces dominate arm swinging. More recently, Hinrichs (1990) observed lower peak shoulder muscle activations (4–9 per cent of maximum across a variety of gait speeds), and concluded that the relative contributions of pendulum dynamics and muscle forcing remain unresolved.

There are a number of possible benefits to arm swinging. Suggested effects include reduced vertical displacement of the centre of mass (Murray *et al.* 1967; Hinrichs 1990; Umberger 2008), and reduction of angular momentum (Elftman 1939; Hinrichs 1990; Bruijn *et al.* 2008; Herr & Popovic 2008; Park 2008), angular displacement (Fernandez-Ballesteros *et al.* 1965; Murray *et al.* 1967; Pontzer *et al.* 2009) or ground reaction moment (Witte *et al.* 1991; Li *et al.* 2001), all about the vertical axis. Other possible effects include prevention of uncontrolled arm motions (Jackson *et al.* 1983) and increased walking stability (Ortega *et al.* 2008). It has even been proposed that arm swinging may be an evolutionary relic from quadrupedalism that serves little or no purpose (e.g. Murray *et al.* 1967; Jackson *et al.* 1983). Arm swinging nevertheless appears to have some physiological benefit, as evidenced by reports of increases in the energetic cost of walking when the arms are prevented from swinging (Ortega *et al.* 2008; Umberger 2008). The mechanical explanation of this benefit remains unclear.

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These issues may be addressed in part through dynamical analyses. Mathematical models indicate that arm swinging may arise simply from translations of the shoulders (Kubo *et al.* 2004), although muscle inputs may be necessary for sustained rhythmic swinging (Jackson *et al.* 1983). Most such models study how walking may induce arm motion, but not how arm motion affects walking, thus giving little indication of possible benefits. Some dynamic walking robots, however, have been found to benefit from arm swinging by prevention of de-stabilizing twisting motions (Collins *et al.* 2001; Collins *et al.* 2005), related to the proposed effects of balanced vertical angular momentum (Elftman 1939) and reduced ground reaction moment (Li *et al.* 2001). An integrative model, in which arm and leg motions interact dynamically during gait, might be helpful for determining which of the proposed mechanical effects are most physiologically relevant.

The purpose of the present study is to combine an integrative model with an experimental study of arm swinging. The model integrates three-dimensional passive dynamic walking with arm motion, which we propose can be sustained by passive dynamics alone. The model also quantifies many of the mechanical effects on the rest of the body, such as angular momentum, stability and mechanical energy losses. We hypothesize that arm swinging requires little muscular effort, but might yield a net energetic benefit from mechanical effects such as reduction of vertical ground reaction moment. We test this by human subject experiments in which multiple types of arm swinging are used to test and separate possible mechanical effects, which are then compared with metabolic energy expenditure. We propose that active arm swinging may require very little effort yet indirectly provide a substantial metabolic benefit.

2. MODEL AND EXPERIMENTAL METHODS

This study comprised three main components: a passive dynamic walking model with free-swinging arms, physical demonstrations of passive dynamic arm swinging, and experiments to measure mechanical and metabolic effects on human walking. A common theme throughout these components was the study of multiple types of arm swinging, comparing walking with normal arm motion (referred to as Normal) to walking without arm motion and to swinging with phase opposite to normal (Anti-Normal). Anti-Normal swinging was intended to preserve many of the proposed mechanical effects, but to reverse those concerned with rotation about the vertical. We studied two ways to prevent arm motion, volitionally holding the arms still (Held) and binding them to the sides of the body (Bound), to test whether passive dynamics actually makes it easier to allow the arms to swing than not. The experiments quantified the effect of these conditions on joint mechanics, centre of mass mechanics, and vertical ground reaction moments, as well as metabolic energy expenditure.

(a) *Dynamic walking model*
We developed a simple dynamic walking model with free-swinging arms. The model (figure 1*a*) was based on a three-dimensional passive dynamic walking model described by Kuo (1999), modified to include free-swinging arms. The model consisted of two curved feet, two straight legs, a pelvis

and two arms. Each joint had a single degree of freedom: inversion–eversion at the ankle, flexion–extension at the hip and at the shoulder. The arms were attached at the hip so as to minimize additional parameters and degrees of freedom, similar to the robot described by Collins *et al.* (2001). The stance foot rolled along the ground, which had a slight downward slope. This slope provided the model with a small amount of energy during the single-support phase of each step (from gravitational potential energy), which balanced the energy that was dissipated in the collision that occurred during each step-to-step transition. In order to allow human-like step frequencies, we also included a passive inter-leg hip spring (Dean & Kuo 2008). The hip spring was not actively controlled and did not add energy to the system. Mass properties of the limbs were based on human anthropometry from Winter (1990), with each arm comprising 4 per cent of body weight, each leg comprising 16 per cent of body weight, and the pelvis/trunk comprising 60 per cent of body weight.

We used standard computational methods (e.g. Kuo 1999) to find walking gaits with different modes of arm swinging. Periodic motions, also known as limit cycles, were found by searching for initial conditions that, when applied to a simulation for an entire walking step, yielded the same initial conditions for the next step (McGeer 1990). A gradient search method was used to iteratively reduce the difference between initial conditions for one step and the next, thereby refining initial conditions toward a periodic gait. We searched for specific modes of arm swinging using a technique wherein we first rigidly constrained the arms to move in the desired mode, then tracked the initial conditions for the limit cycle as the constraints were gradually relaxed and finally removed altogether (using a differential damped-spring and inertia on the shoulder joints in a manner similar to that described by Gomes & Ruina 2005). This enabled us to find sets of initial conditions for a variety of periodic gaits with different arm swinging modes.

We observed several distinct modes of passive arm swinging, including a motion similar to that observed in typical human gait (Normal, figure 1*c*), one with phasing opposite from normal (Anti-Normal), one in which the arms and legs were nearly 90° out of phase (Mid-Phase), and one with the arms swinging together (Parallel). We also constrained the model's arms to remain vertical (Bound), and found a gait very similar to that of the model without arms. All of these motions were found with the same model parameters. The gaits were found to have similar stability, as indicated by eigenvalues of the linearized step-to-step mapping (e.g. McGeer 1990). The fore-aft motion of the legs was stable and the lateral motion unstable, similar to the findings of Kuo (1999), while the eigenvalues associated with arm motions indicated neutral stability. All modes were tested for the same slope of 0.03 rad, and so had equal mechanical cost of transport (energy per unit distance). Non-dimensional gait speeds were 0.293, 0.258 and 0.295 for Normal, Bound and Anti-Normal gaits, with Normal being roughly equivalent to a slow walking speed of 0.87 m s^{-1} for humans. We also found that slower speeds (caused by shallower slopes) yielded an unusual gait in which the arms swung at very low amplitude and twice the frequency of the legs, similar to what has been observed in humans (Craik *et al.* 1976). (Animations of these arm swinging modes, mode characteristics and model parameters may be found in the electronic supplementary material.)
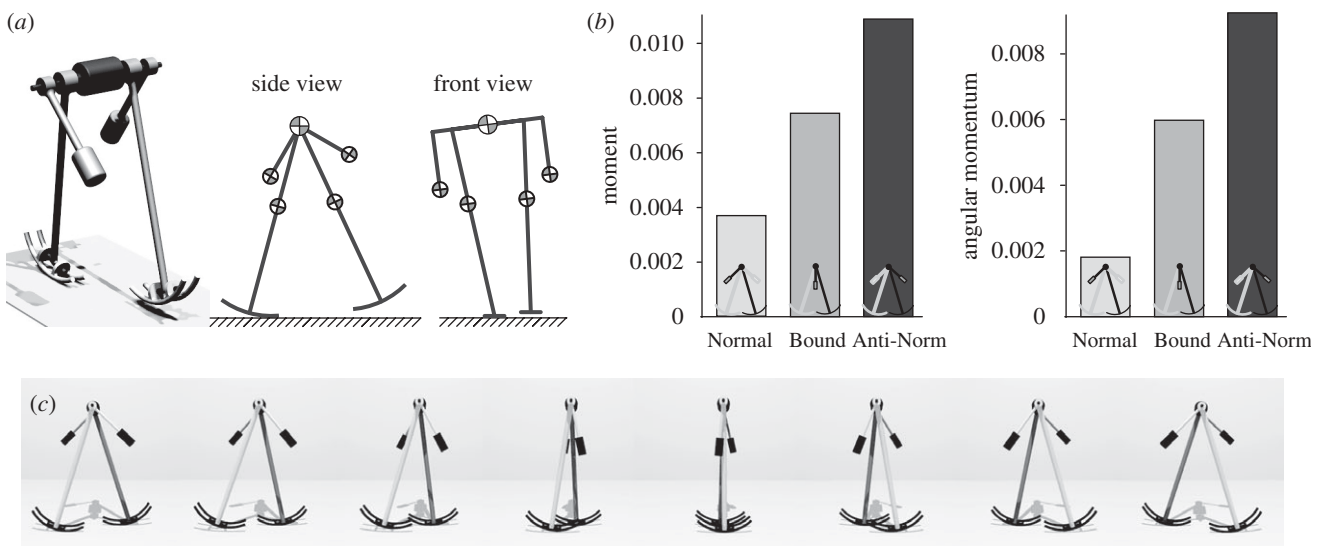
(*a*)

side view   front view

(*b*)

(*c*)

Figure 1. (*a*) Dynamic walking model with free-swinging arms, powered only by descending a gentle slope (see electronic supplementary material for detailed schematics and animations). (*b*) Simulation results for peak vertical ground reaction moment and vertical angular momentum in the Normal, Bound and Anti-Normal arm-swinging modes. Normal and Anti-Normal refer to passive dynamic gaits with the arms moving similar to normal human walking and the opposite phasing, respectively. Bound refers to a model walking with the arms mechanically constrained against rotation. Units are dimensionless, using body mass *M*, leg length *L* and gravitational acceleration *g* as normalizing factors. (*c*) Frame-by-frame rendering of the Normal gait (walking from left to right).

The mode of arm swinging had substantial effect on vertical angular momentum and ground reaction moments in our model. Peak vertical ground reaction moments and peak vertical angular momentum more than doubled when arms were Bound and increased still further with Anti-Normal swinging (figure 1*b*). Vertical excursions of the centre of mass changed much less between Normal, Bound and Anti-Normal modes (0.046, 0.053, and 0.045 leg lengths, respectively).

These model findings were used to adopt three primary hypotheses for the mechanical effects of arm swinging. Of the effects proposed in the literature, the only ones supported by the model were reductions in vertical angular momentum, vertical ground reaction moments, and to a lesser degree vertical displacement of the centre of mass. The experiments were therefore designed to test for these effects as a function of arm swinging.

(b) *Physical demonstrations of passive dynamic arm swinging*
We used freely swinging artificial arms attached to a person's shoulders to test for arm-swinging modes similar to those observed in simulation. The physical model used artificial, freely rotating arms made of rope or sticks attached to a person through a yoke placed on the shoulders (figure 2). The person's arms were bound to the body.

The artificial arms corroborated the passive motions predicted by the simulation. We found that with appropriate initial conditions, the arms could easily oscillate in the Normal and Anti-Normal modes found in simulation, as well as the Parallel and Mid-Phase modes (see electronic supplementary material for videos).

(c) *Experimental procedures*
We measured mechanical effects and metabolic energy expenditure in able-bodied human subjects as they walked with four different arm swinging modes (figure 3), chosen
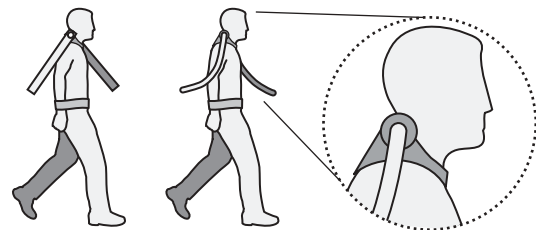

Figure 2. Artificial arms made of wood (left) or rope (right) provide physical demonstrations of passive dynamic arm swinging during walking. The artificial arms were allowed to swing freely from a yoke (inset) worn by individuals as they walked with their arms bound to their sides (see electronic supplementary material for videos).
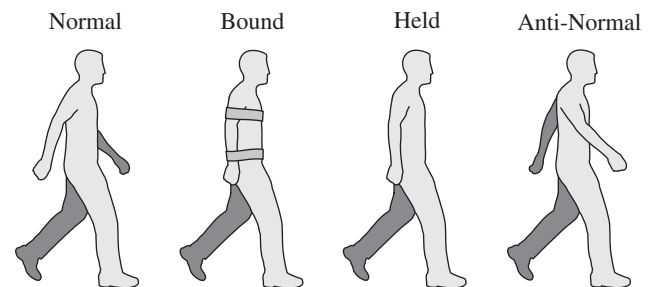

Figure 3. Walking conditions tested experimentally. Human subjects walked with arms swinging normally (Normal), physically bound to the sides (Bound), volitionally held to the body (Held) and volitionally swinging with phase opposite to normal (Anti-Normal). Gait mechanics and metabolic energy expenditure were measured under these conditions.

based on our simulation results. The conditions were Normal, Bound, Held and Anti-Normal arm swinging, expected to produce the following effects. First, we expected vertical ground reaction moment (about the centre of pressure of each foot) and vertical angular momentum

(about the centre of mass) to increase in the order Normal, Bound/Held (no difference between the two) and Anti-Normal, as they did in simulation. In contrast, we expected vertical centre of mass displacement to be relatively unaffected, and to be nearly the same for Normal and Anti-Normal conditions, the latter therefore serving to isolate this effect from the others. Because the vertical ground reaction moment is transmitted upward from the foot through the leg and the body, we hypothesized that muscular effort may be required to resist that moment. We therefore expected the metabolic cost to increase in the same order of Normal, Bound/Held and Anti-Normal. Finally, the Held condition was applied because simulations suggested that it may actually require effort to prevent the arms from swinging naturally, implying that the Held condition might require greater energy expenditure than Bound.

We compared these conditions at a walking speed of 1.25 m s^{-1}, with metabolic data recorded during treadmill walking and gait mechanics measured during overground walking. A total of 10 healthy adult subjects ($n = 10$; seven males and three females, aged 23–47 years) participated in the study. All subjects provided informed consent according to university procedures. All subjects (body mass 70.5 ± 11.3 kg, leg length 0.902 ± 0.074 m, mean $\pm$ s.d.) participated in energetic trials, but only a subset of subjects ($n = 7$, body mass 75.0 ± 10.2 kg, leg length 0.931 ± 0.073 m) participated in the gait mechanics trials.

Arm-swinging conditions were tested as follows (figure 3). During Normal walking trials, subjects were instructed to walk as naturally as possible. During Held walking trials, subjects were instructed to hold their arms loosely at their sides such that their wrists remained slightly posterior to the greater trochanter at the hip. The same arm posture was enforced during Bound trials through the use of two wide elastic sports bandages. This posture was chosen so as to minimize the interference of the hands with body motions and prevent the arms from obscuring motion tracking markers at the greater trochanter. During Anti-Normal conditions, subjects were instructed to swing their arms in phase with the ipsilateral leg and with swing amplitude approximately equal to normal. Subjects typically required an adaptation period of a few minutes to become comfortable with this condition, after which they reported no difficulties. All conditions were conducted in random order.

We measured metabolic energy expenditure and gait mechanics as subjects performed the experimental conditions. We estimated metabolic energy expenditure from the rate of oxygen consumption and carbon dioxide production recorded using open-circuit respirometry (Brockway 1987). The metabolic rate for quiet standing was subtracted from the gross rate for walking to yield net metabolic rate. For recording gait mechanics, we measured ground reaction forces and kinematics as subjects walked over ground-embedded force plates (calibrated as described in Collins *et al.* 2009). These were processed to yield vertical ground reaction moment, defined as the moment between foot and ground about a vertical axis extending through the centre of pressure of individual foot contacts (also referred to as 'free vertical moment', e.g. Li *et al.* 2001). During double support, we measured separate vertical ground reaction moments for each foot. Vertical angular momentum was defined as the vertical component of the angular momentum of the body about the centre of mass (e.g. Elftman 1939), calculated from segment kinematics. We also calculated joint torques and work rates and work performed on the

centre of mass. To obtain joint torques and work rates, standard inverse dynamics analyses were performed in three dimensions (e.g. Winter 1990; Siegler & Liu 1997). As an indicator of mechanical energy losses, we also estimated the work performed on the centre of mass by each leg ('COM work'), defined as the time-integral of the vector dot product of each leg's ground reaction force with the centre of mass velocity (Donelan *et al.* 2001, 2002) (see electronic supplementary material for details on experimental measurements).

Data trials from each condition were averaged for each subject. Averages were performed with each trial normalized in time to per cent gait cycle. All torque, work rate and work quantities were analysed in dimensionless form to help account for variations in subject size. The base units were each subject's body mass M, leg length L and gravitational acceleration g. Averages, s.d.s and statistics were computed in dimensionless quantities, but we report some variables in more familiar dimensional units such as W kg^{-1}, converted using average normalization factors. The average normalization factors used were: $MgL = 685 \text{ kg m}^2 \text{ s}^{-2}$ for torque and mechanical work, $Mg^{1.5}L^{0.5} = 2220 \text{ kg m}^2 \text{ s}^{-3}$ for mechanical work rate, $Mg^{1.5}L^{0.5} = 2060 \text{ kg m}^2 \text{ s}^{-3}$ for metabolic rate and $Mg^{0.5}L^{1.5} = 2110 \text{ kg m}^2 \text{ s}^{-1}$ for angular momentum.

We statistically compared outcome variables that captured the primary energetics and mechanics results. We compared net metabolic rate, peak vertical ground reaction moment, peak vertical angular momentum, peak upper-limb joint torque, upper-limb joint work, COM work and vertical excursion of the centre of mass. Peak vertical angular momentum, peak vertical ground reaction moment, and peak upper-limb joint torques were each calculated as the maximum absolute value during a single stride. We integrated each joint power over the intervals of the stride for which it was positive to obtain joint work, then divided by stride period to yield a term referred to here simply as 'joint work rate'. Similarly, positive COM work rate refers to positive COM work during a stride divided by stride period. Vertical excursion of the centre of mass was calculated as the difference between the maximum and minimum vertical component of the centre of mass position during a stride. Values for each outcome variable were obtained for the averaged stride of each subject in each condition. Statistical comparisons were made with repeated measures analysis of variance (ANOVA) for each variable, with a significance level of 0.05. Where differences were significant, post hoc comparisons were performed using paired t-tests, controlling for multiple comparison errors with the Sidak-Holm step-down procedure (Glantz 2005).

3. EXPERIMENTAL RESULTS
In human subjects, the arm swinging modes yielded significant differences in three main outcome variables: vertical angular momentum, vertical ground reaction moment and metabolic rate. Bound, Held and Anti-Normal modes caused significantly greater ground reaction moments and vertical angular momentum when compared with Normal, accompanied by significant increases in metabolic rate. The conditions had much less effect on other variables. For example, work performed by the leg joints and overall COM work were nearly the same across the Bound, Held and Anti-Normal conditions,

with some work quantities slightly higher than Normal. Vertical excursion of the centre of mass did not change significantly across conditions. Arm swinging appeared nearly passive in both Normal and Anti-Normal conditions, as evidenced by low torque and work in the shoulder and elbow joints.

Vertical ground reaction moments were strongly affected by arm-swinging mode. As expected, peak vertical ground reaction moments increased from Normal (3.0 ± 1.2 N m) to Bound (5.0 ± 1.5 N m or 63% greater than Normal, $p = 2 \times 10^{-4}$) and from Held (4.8 ± 2.0 N m) to Anti-Normal (8.8 ± 1.7 N m or 83% greater than Held, $p = 0.001$, mean $\pm$ s.d., figure 4*a*,*b*). This was a direct result of arm swinging. In the Normal condition, the angular momentum of the arms was of opposite phase to that of the legs (with peak momentum -1.1 ± 0.2 N m s), while in both Bound (-0.1 ± 0.1 N m s) and Held (-0.2 ± 0.1 N m s) conditions, the arms had negligible angular momentum and in Anti-Normal (0.9 ± 0.2 N m s) the arms' angular momentum was in phase with that of the legs (figure 4*c*,*d*). However, subjects did not appear to compensate for this effect. The angular momentum of the rest of the body, dominated by leg angular momentum, remained nearly constant across conditions (figure 4*c*). The net effect was that whole-body angular momentum increased significantly over Normal (1.0 ± 0.2 N m s) in the Bound condition (1.7 ± 0.2 N m s) and Held condition (1.7 ± 0.3 N m s, 77% greater than Normal, $p = 2 \times 10^{-4}$), and further increased in the Anti-Normal condition (2.8 ± 0.1 N m s or 65% greater than Held, $p = 2 \times 10^{-6}$; figure 4*e*,*f*). These increased fluctuations in angular momentum necessitated increased ground reaction moments, as were observed.

There were significant differences in metabolic energy expenditure for all conditions (figure 5). As hypothesized, metabolic rate in the Normal condition was lowest at 3.09 ± 0.12 W kg^{-1}, and increased by 7 per cent in the Bound condition (3.31 ± 0.22 W kg^{-1}; $p = 7 \times 10^{-4}$). Metabolic rate was 12 per cent greater than Normal in the Held condition (3.45 ± 0.25 W kg^{-1}), and 26 per cent greater than Normal in the Anti-Normal condition (3.93 ± 0.30 W kg^{-1}). Increases among all conditions were statistically significant ($p = 0.004$ for comparisons of Held to Bound; and $p = 1 \times 10^{-5}$ for Anti-Normal to Held).

Upper-limb joint torques and work rates were quite low during both Normal and Anti-Normal conditions. Peak shoulder joint torques averaged 2.2 ± 0.6 N m in Normal, and 1.6 ± 1.0 N m in Anti-Normal (figure 6); for reference, both of these are less than 25 per cent of the torque required to hold the arm in a horizontal posture (approx. 11 N m). Peak elbow torques were similarly low, measuring 2.5 ± 0.6 N m in Normal, and 2.2 ± 0.5 N m in Anti-Normal. Mean positive work rates for both the shoulder and elbow were always less than 0.013 W kg^{-1}, or less than 1 per cent that of the lower limbs. There were no significant differences in upper-limb peak torques or work rates between Normal and Anti-Normal conditions ($p > 0.2$). Upper-limb torques and work rates could not be calculated reliably in Bound or Held conditions due to the unknown forces from elastic restraints and intermittent contact between the arms and trunk.

There were no significant changes in the vertical excursion of the centre of mass. Vertical excursions were 0.050 ± 0.010 m for Normal, 0.054 ± 0.006 m for Bound, 0.052 ± 0.007 m for Held and 0.054 ± 0.008 m for Anti-Normal (no differences, ANOVA, $p = 0.15$). Lower-limb joint angles, joint torques and joint work rates did not vary consistently across conditions (figure S1 and table S1 in the electronic supplementary material). Positive COM work was 0.53 ± 0.08 W kg^{-1} for Normal, 0.59 ± 0.05 W kg^{-1} for Bound, 0.58 ± 0.04 W kg^{-1} for Held, and 0.59 ± 0.06 W kg^{-1} for Anti-Normal. This increase over Normal was significant for Bound ($p = 0.004$), Held ($p = 0.04$) and Anti-Normal ($p = 0.001$), while other comparisons were not statistically significant ($p > 0.3$). Lateral excursion of the centre of mass was also unaffected by arm condition.

4. DISCUSSION

We performed simulations and human subject experiments aimed at determining the function of arm swinging and the mechanisms controlling it during human gait. We had proposed that arm motions might primarily be the result of passive dynamics, and accordingly found fully passive gaits in simulation exhibiting an array of modes of arm swinging. In agreement with these simulations, human subjects produced minimal shoulder and elbow joint torques during the Normal and Anti-Normal gaits. Normal arm swinging therefore appears to require little effort. We had also hypothesized that arm swinging may play an important role in gait by offsetting the motion of the legs, reducing vertical ground reaction moments and attendant muscle forces, thereby reducing metabolic energy expenditure. We found that Bound, Held and Anti-Normal arm conditions all resulted in increased ground reaction moments and increased metabolic cost. These findings are consistent with the hypothesis that the arms are easily swung by exploiting natural dynamics, with significant benefits to gait economy due to reduced ground reaction moments.

The net energetic effect of arm swinging is aided by a low cost for swinging the arms. We found several indications of low cost, starting with passive dynamic arm swinging in both simulations and physical demonstrations with artificial arms. Passive dynamics indicate that little active torque is needed to sustain swinging. Indeed, the peak torques observed at human shoulder and elbow were only a few per cent of those at the leg joints, even with Anti-Normal swinging. Recent electromyographic evidence also indicates little muscular effort (Pontzer *et al.* 2009). As suggested by others (Murray *et al.* 1967; Jackson *et al.* 1983; Hinrichs 1990), the relatively small amount of work accompanying arm swing may even be partially due to elastic tendon work. The physical demonstration of sustained, passive swinging was also quite robust, whether the artificial arms were made of wood or rope. This suggests that uncontrolled motions (Jackson *et al.* 1983) do not pose a substantial problem, although the neutral stability of the model's arm motions indicate that some control might still be helpful. All of these indicators suggest that little energy is expended to actively swing the arms.

Some previously hypothesized explanations of arm swinging are unsupported by our results. One example
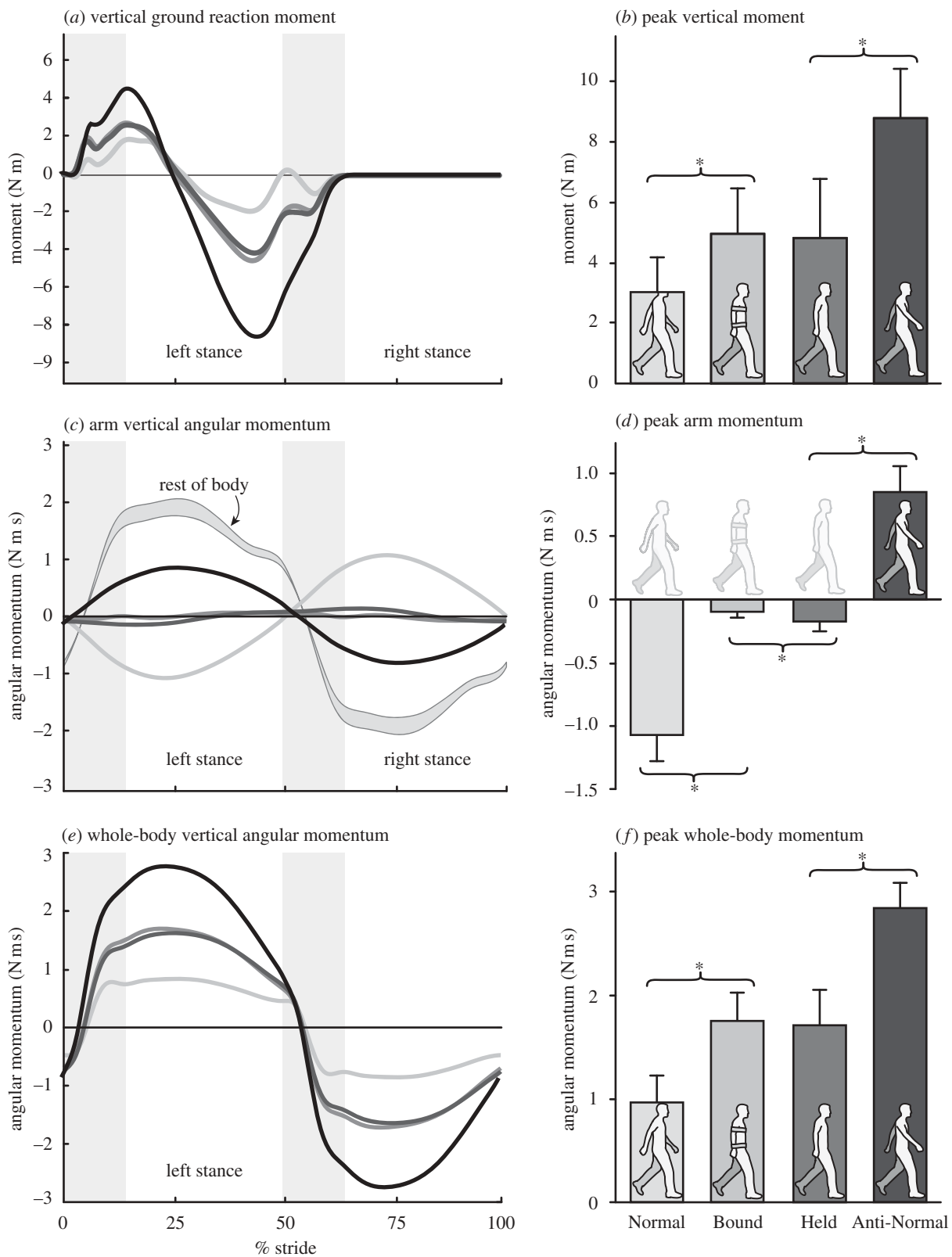
Figure 4. Experimental measurements of the mechanical effects of arm swinging during human walking. (*a*) Vertical ground reaction moment about the centre of pressure of the stance foot plotted versus time and (*b*) the peak moments over a stride; (*c*) the arms' contribution to vertical angular momentum versus time and (*d*) corresponding peak values; (*e*) whole-body angular momentum about the vertical versus time and (*f*) corresponding peak values. Trajectories show mean across subjects per condition, and bar graphs show peaks averaged across subjects. Error bars show 1 s.d. and asterisks indicate statistical significance ($p < 0.05$). Double support is denoted by a shaded region in plots. In (*c*), the band labelled 'rest of body' represents the vertical angular momentum of the body not including the arms; the range contains all mean trajectories, which were found to be dominated by the legs. Arm angular momentum increased in the order Normal, Bound/Held, Anti-Normal as expected while angular momentum of the rest of the body remained roughly constant, resulting in significant increases in whole-body angular momentum. Increased peak vertical moments were necessitated by increased rates of change in whole-body vertical angular momentum. (*a,c,e*) Light grey, Normal; mid grey, Bound; dark grey, Held; black, Anti-Normal.
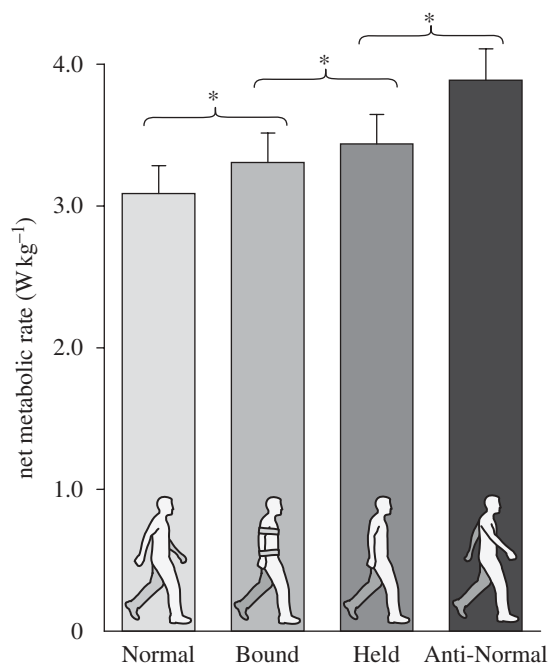
Figure 5. Net metabolic rate increased significantly in the order of Normal, Bound, Held and Anti-Normal. Bars indicate mean energy expenditure rate across subjects. Error bars denote s.d.'s, and asterisks indicate statistical significance ($p < 0.05$).

is reduction of vertical displacement of the body centre of mass (Murray *et al.* 1967; Hinrichs 1990; Umberger 2008). In our experiment this displacement was not found to change significantly with arm-swinging condition, perhaps due to the relatively low mass of the arms. Both Normal and Anti-Normal arm swinging could have produced the proposed benefits, yet had opposite effects on metabolic rate. This is not surprising, considering that other studies have found that reduction of centre of mass displacement can actually increase metabolic energy expenditure (Ortega & Farley 2005; Massaad *et al.* 2007; Gordon *et al.* 2009). Our results also contradict the hypothesis that arm swinging is merely an evolutionary relic of quadrupedalism (Murray *et al.* 1967; Jackson *et al.* 1983), since they corroborate previous reports of a significant energetic benefit (Park *et al.* 2000; Ortega *et al.* 2008; Umberger 2008). Another hypothesized explanation is reduced trunk rotation (e.g. Fernandez-Ballesteros *et al.* 1965), but our results showed no change in trunk angular momentum with different arm-swinging conditions. The parts of the body with greater 'rotation' about the vertical direction with increasing angular momentum were the legs and arms, rather than the trunk. Finally, we also found no clear trends in lower extremity joint work and torques. These were generally unchanged between Bound, Held and Anti-Normal conditions, and therefore appear not to explain the differences in energetic cost.

The more promising explanations are those associated with whole-body rotation about the vertical. The arms were found to counter the legs, resulting in reductions in angular momentum and ground reaction moment about the vertical. Of the two, we propose that the vertical moment about the stance foot is more directly related to energetic cost. This moment is transmitted upward

from the foot, through the leg and to the pelvis, apparently resisted by internal/external rotation moments produced by muscle along the way and thus requiring metabolic energy expenditure.

In contrast, there is not an obvious physiological penalty for high angular momentum. It has been proposed that humans may have an intrinsic goal of minimizing whole-body angular momentum (e.g. Herr & Popovic 2008) or reducing angular motion about the vertical (e.g. Elftman 1939) during walking. But walking with pendular dynamics generally produces non-zero angular momentum and motion, with little adverse effect. The human subjects in this study also did not minimize vertical angular momentum, which could have been reduced further—theoretically to zero—by appropriate motions of the arms and other parts of the body. Angular momentum is certainly convenient for quantifying the motions of the arms and legs (figure 4), but we believe its physiological relevance to be indirect: its rate of change is related to the vertical ground reaction moment, which we believe to be more directly associated with muscular energy expenditure.

Our results may be used to infer an energetic cost–benefit balance for arm swinging in walking. Arm swinging might have two separate effects: a direct cost for driving arm swing and an indirect benefit for reduced ground reaction moments. Direct costs might include the metabolic requirements of shoulder muscles for driving the arms, presumably minimized when allowing the arms to swing as naturally as possible. Swinging the arms in other ways, such as with greater or lesser amplitude, or even holding them still, could all increase the direct costs. Our results suggest that the direct cost of holding the arms in place might be approximately 5 per cent of the energy used in walking, based on the observed difference in energy use between Bound and Held conditions (for which lower-limb mechanics were nearly identical).

Indirect benefits of arm swinging might include reduced metabolic requirements for leg muscles in producing torques that resist vertical ground reaction moments. Our results suggest an indirect benefit of approximately 7 per cent for normal arm swinging, based on the observed difference in energy use between Normal and Bound (where direct costs were presumably zero). This value is consistent with the recently observed reductions of 8 per cent by Umberger (2008) and 6 per cent by Ortega *et al.* (2008) in similar conditions. Taking the direct costs and indirect benefits together, we would expect overall energy requirements to be minimized when ground reaction moments are small and arm motions are close to a passive mode of swinging. Perhaps it is such an optimum that humans seek in normal gait.

The simulation model used in this study has a number of limitations. The model does not provide a direct means for estimating increases in metabolic energy use in human subjects, but rather predicts trends in ground reaction moment and angular momentum. Metabolic consequences of these changes must be inferred from a separate understanding of physiology. The model was also limited in that arms were represented in a simplified form, without elbows and with shoulders at the hip. It is possible that an intervening trunk or a two-link arm could change some of the arm-swinging dynamics, but the
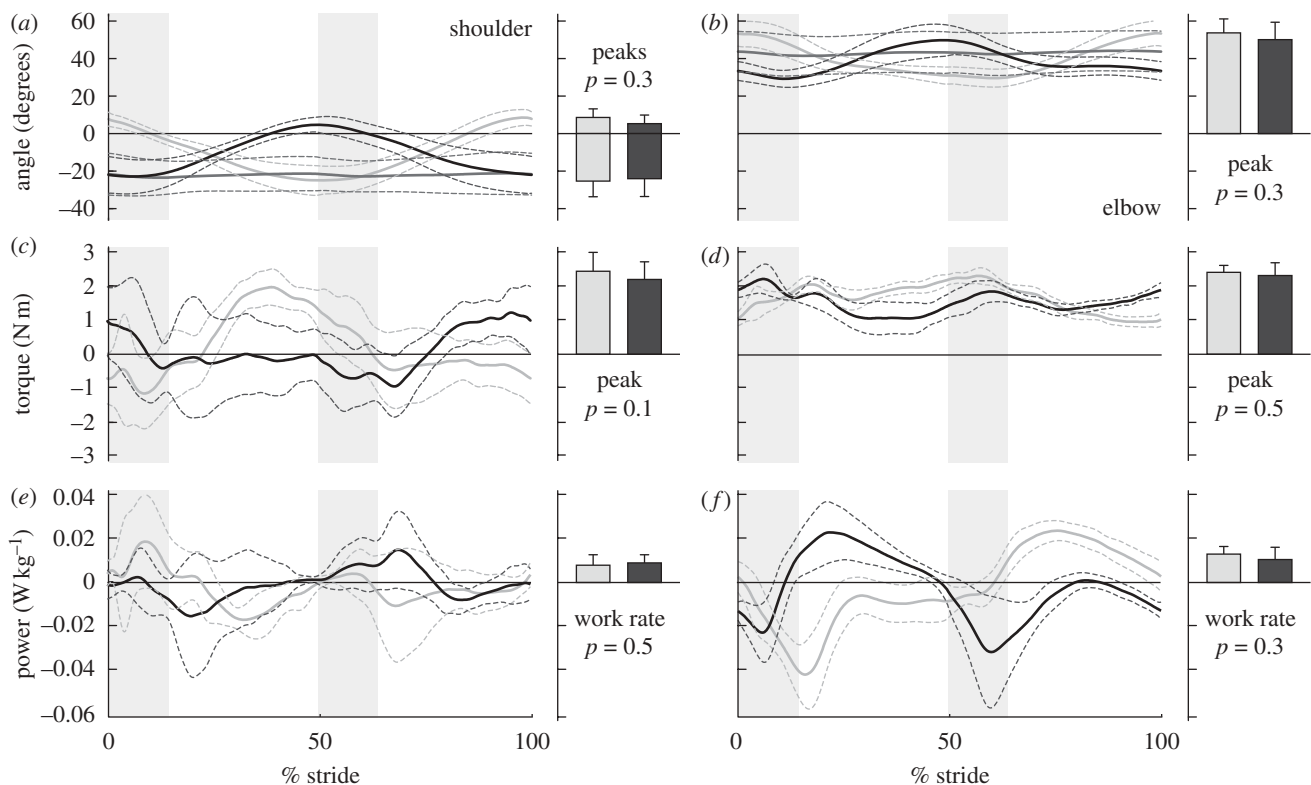
Figure 6. Upper-limb joint mechanics. (*a*) Shoulder and (*b*) elbow angles versus time (left) and corresponding peak angles (bars at right). (*c*) Shoulder and (*d*) elbow torque versus time (left) and corresponding peak torques (bars at right). (*e*) Shoulder and (*f*) elbow joint power versus time (left) and corresponding positive work rate (bars at right). Peak torques and work rates were quite small, and did not significantly differ in magnitude between Normal and Anti-Normal conditions, suggesting that arm motions were predominantly passive. Light grey, Normal; dark grey, Held; black, Anti-Normal; dashed lines, ±s.d.

physical demonstrations with artificial arms suggest that passive arm swinging is quite robust to such changes. The simulation results (figure 1) also show a reasonably similar trend in ground reaction moment and angular momentum as a result of arm-swinging condition.

Our experiment was limited by the fact that during both Held and Bound conditions, inverse dynamics were not available due to contact between the arms and trunk. Subjects' hands were also held in a position slightly posterior to the position where they would hang naturally, so as to prevent the hands from interfering with leg motions and hip markers. This could have led to a slight increase in shoulder torque during Held trials, which might slightly exaggerate the effort of suppressing arm swing. Another possible limitation was that we focused on vertical moments that occurred during single support, but some portion of the change in vertical angular momentum occurred during double support. These issues limit the accuracy with which we can estimate the physiological costs and benefits of arm swinging, but not the overall conclusions drawn here.

We also did not experimentally study the effects of walking speed, which has a strong effect on human arm swinging (e.g. Murray *et al.* 1967). Humans may swing their arms higher and faster with increasing gait speed, in part to cancel the effects of longer, quicker steps by the legs. In simulation, we found that high speeds or fast leg swing frequencies required actuation (for example by a shoulder spring) for sufficiently

rapid arm motions. The direct costs of shoulder effort would therefore increase with speed, but the indirect benefits of reduced vertical moments would increase as well, since arm motions would cancel larger effects from faster leg motions. By contrast, at very low speeds, the legs induced only small vertical moments, and so counter-motions of the arms may yield little benefit. In simulation, we found qualitatively different modes of oscillation at very low speeds, including a mode resembling 'double-swing' observed by others (Craik *et al.* 1976; Webb & Tuttle 1994; Wagenaar & van Emmerik 2000). However, there would appear to be little benefit for choosing or maintaining any particular arm motion at slow speeds, which may explain why there is usually more variation in arm motions (Donker *et al.* 2001).

Although we did not independently test subjects' stability, our results provide some insight into the possible role of arm swinging. Ortega *et al.* (2008) proposed that arm swinging may improve stability, because arm swinging had no effect on the energetic cost of walking with external lateral stabilization. Our dynamic walking simulation did not reveal any inherent relationship between arm swinging mode and step-to-step gait stability. Perhaps external lateral stabilization also helps to resist vertical ground reaction moments, making the arms less beneficial in that regard. There may yet be a benefit to making the arms available to thrust in various directions in response to perturbations, but in the Anti-Normal condition (where the arms were available for thrusting), this was apparently outweighed by other indirect costs. In

terms of step-to-step stability, our results show little evidence of a benefit to arm swinging.

Arm motions during gait seem to result primarily from passive dynamics, with muscles used to initiate motion, correct errors as they arise, and provide increased torques at higher speeds. Although arm swinging is relatively easy to achieve, its effect on energy use during gait is significant. Arm swinging can reduce ground reaction moment requirements, leading to overall decreased energy expenditure, perhaps in the muscles of the lower limbs. Rather than a facultative relic of the locomotion needs of our quadrupedal ancestors, arm swinging is an integral part of the energy economy of human gait.

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