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RESEARCH ARTICLE

A low-cost method for carrying loads during human walking

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ABSTRACT

Humans often perform tasks that require them to carry loads, but the metabolic cost of carrying loads depends on where the loads are positioned on the body. We reasoned that carrying loads at the arms' center of mass (COM) during walking might be cheap because arm swing is thought to be dominated by passive pendulum dynamics. In contrast, we expected that carrying loads at the leg COM would be relatively expensive because muscular actuation is necessary to initiate and propagate leg swing. Therefore, we hypothesized that carrying loads at the arm COM while swinging would be cheaper than carrying loads at the leg COM. We further hypothesized that carrying loads at the arm COM while swinging would be more expensive than carrying loads at the waist, where the mass does not swing relative to the body. We measured net metabolic power, arm and leg motion, and the free vertical moment while subjects (n=12) walked on a treadmill (1.25 m s^{-1}) without a load, and with 8 kg added to the arms (swinging versus not swinging), legs or waist. We found that carrying loads on the arms or legs altered arm swinging amplitude; however, the free vertical moment remained similar across conditions. Most notably, the cost of carrying loads on the swinging arms was 9% less than carrying at the leg COM (P<0.001), but similar to that at the waist (P=0.529). Overall, we found that carrying loads at the arm COM is just as cheap as carrying loads at the waist.

KEY WORDS: Arm swing, Metabolic cost, Energetics, Locomotion, Biomechanics

INTRODUCTION

It is quite common for animals to carry loads, but the metabolic rate associated with such activities will depend on where on the body the load is carried. It has been shown that if humans carry a 2 kg load at each foot (4 kg total), this will increase net metabolic rate of walking by 34% (Browning et al., 2007). It has also been shown that if dogs carry a 0.19 kg load at each foot (0.77 kg total), this will increase the gross metabolic rate of running by 10% (Steudel, 1990). Yet, in both of these studies (Browning et al., 2007; Steudel, 1990), it was found that if the load is positioned near the body's center of mass (COM), humans and dogs can carry the load without an appreciable increase in metabolic rate. Other load-carrying strategies that do not yield an appreciable increase in metabolic rate include women from Africa carrying loads on top of their heads (Maloiy et al., 1986) and female wallabies that can hop with their young in their pouch (Baudinette and Biewener, 1998). Thus, carrying loads can be expensive or cheap, but this will depend on its anatomical location.

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Finding ways to carry loads in the cheapest, and at times the most comfortable, way possible has led humans to develop clever load-carrying methods. Such methods include carrying loads suspended from springy bamboo poles by people in Asia (Kram, 1991). This inspired Rome and colleagues (2006) to invent an ergonomic backpack that uses springy rubber bands to suspend and carry loads. These load-carrying methods are quite clever because they exploit the physics of a spring–mass system, where the springy element is used to effectively decouple the body's up and down motion from being induced onto the load. The body still has to cope with the extra load that is carried because the average force over one period of oscillation will be equal to the added load. But, the metabolic cost of carrying a load that is suspended by a springy element is lowered. This is a classic example of using a springy element to resolve a situation whereby induced oscillations of a load can be controlled in a desirable manner (Den Hertog, 1985). Inspired by the physics of these load-carrying methods (Kram, 1991; Rome et al., 2006), we imagined a situation where loads could be carried on the arms while they swing. Allowing the arms to swing might be a way to exploit the pendulum-like swinging oscillations of the arms while walking, so that the cost of carrying the load remains relatively cheap. Arm-swinging oscillations are desirable during walking because they provide a mechanical benefit by means of reducing the free vertical moment that the feet exert on the ground during stance (Collins et al., 2009; Li et al., 2001). A reduction in the free vertical moment that the feet exert on the ground reflects a reduction in the body's twisting torque about the vertical axis; therefore, the mechanical benefit of arm swing comes from its ability to help stabilize whole-body rotation (Huang and Ferris, 2004).

One might suppose that swinging the arms while walking is caused by active, muscle contractile elements that do work to move the limb and thus exact a metabolic cost. Yet, the metabolic cost of swinging the arms appears to be relatively cheap because the forward and backward motion can be induced by a combination of minimal muscle actuation (Ballesteros et al., 1965; Goudriaan et al., 2014; Jackson et al., 1978) and passive dynamics (Arellano et al., 2012; Collins et al., 2009; Goudriaan et al., 2014; Jackson et al., 1978; Pontzer et al., 2009). However, the passive dynamics that underlie arm swing motion seems to dominate its behavior (Collins et al., 2009; Pontzer et al., 2009). For example, Pontzer et al. (2009) has proposed a passive arm swing model where the arms can be driven by forces that are transmitted to the torso/shoulders from the swinging legs. In a physical demonstration, Collins et al. (2009) showed that during walking, artificial pendulum-like arms attached to the shoulders can swing passively, i.e. without the need for muscle actuation. Inspired by these findings, Arellano et al. (2012) modeled the human arms as swinging pendulums driven by the horizontal motion arising at the shoulder joint. They found that at the energetically optimal walking speed (1.25 m s^{-1}), the arms swing at their natural frequency. That suggests that little mechanical energy input is needed from arm/shoulder muscle actuation, consistent with the finding that arm swing exacts a negligible cost (Collins et al., 2009).

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Is it possible that arm swing and its pendulum-like motion provides a means to carry loads with a low metabolic cost? Previous studies that have attached loads to the limbs have generally found that more distal locations increase the metabolic cost of load carrying (Browning et al., 2007; Rose et al., 1994; Royer and Martin, 2005; Soule and Goldman, 1969). For instance, when compared with walking without a load, carrying 4 kg on each foot increases the allocated cost per kilogram of added load by 42.6 ml O_2 min^{-1} kg^{-1} (Browning et al., 2007). Those values contrast with a modest increase of only 9.6 ml O_2 min^{-1} kg^{-1} for carrying an 8 kg load at the waist (Browning et al., 2007). Earlier research by Soule and Goldman (1969) showed that carrying 4 kg on each hand increased the cost of walking by only 16.5 ml O_2 min^{-1} kg^{-1}. Taken together, these findings suggest that carrying loads on the hands – the most distal portion of the arms – is much less expensive than carrying loads on the feet – the most distal portion of the legs. However, the reported cost of carrying loads on the arms may have been exaggerated in the study of Soule and Goldman (1969) because the location of the loads changes the equivalent length of the pendulum-like arm. If modeling the arm as a compound pendulum, the equivalent length is given as:

$$L_{\mathrm{eq}} = \frac{I}{md}, \tag{1}$$

where I is the moment of inertia of the pendulum about the shoulder's axis of rotation, m is the pendulum mass and d is the distance between the shoulder's axis of rotation and the pendulum's center of mass. More importantly, the equivalent length of the arm is the mechanical parameter that determines the arm's natural pendular frequency [$\omega_n=\sqrt{(\boldsymbol{g}/L_{\mathrm{eq}})}$] and thus will influence its swinging amplitude. A compound pendulum model of arm swing suggests that adding a 4 kg load to the hand would increase the equivalent length of the arm, which would decrease its natural frequency. Any shoulder excitations that drive the system with a frequency different from the arm's natural frequency would respond with much smaller swinging amplitudes. Furthermore, even if the arm swung with smaller amplitude, placing an added load at the hand would induce additional torques on the upper limb that are proportional to the weight that acts perpendicular to its motion. While speculative, these effects might have minimized the role of the arm's passive dynamics, requiring the arm to rely on greater muscular actuation to produce upper-limb torques that coordinate and control the arm's swinging motion, amplitude and frequency, which would exact a substantial metabolic cost.

Instead of adding loads to the hands, we tested the idea that humans could inexpensively carry loads if they are attached at the arm COM so as to minimize a change in the arm's pendulum length, and thus its natural frequency during walking. With such a strategic location of the load, the arms might swing passively and with sufficient amplitude, which might be key to minimizing the metabolic cost of arm swing and the overall cost of walking with loads. Therefore, we hypothesized that carrying loads on the arms would be cheaper than carrying loads on the legs during walking. The rationale underlying this hypothesis is that if carrying the loads on the arms is dominated by passive dynamics, then the cost of passively swinging the load should be small, as opposed to the substantial cost of actively swinging the load when carried on the legs (Browning et al., 2007; Royer and Martin, 2005). We also hypothesized that carrying loads on the swinging arms would more expensive than carrying loads near the body's COM (positioned about the waist in this study) because carrying loads near the body's COM has been generally observed to be the least expensive way to carry loads during walking (Browning et al., 2007; Royer and Martin, 2005; Soule and Goldman, 1969). While our main focus was on understanding the influence of load-carrying conditions on metabolic cost, we also wanted to understand how carrying loads on the arms and legs might influence arm and leg swing amplitude as well as the free vertical moment the foot exerts on the ground; therefore, we measured these variables as well. We also studied a condition where loads were carried on the arms while the arms were prevented from swinging. In line with previous studies, we expected that restricting arm swing would increase the cost of walking (Collins et al., 2009; Umberger, 2008), but we were particularly interested in determining the potential cost savings that would arise from allowing the arms to swing freely with an added load, presumably under conditions when arm swinging motion would benefit metabolically from being dominated by passive dynamics. We quantified this by comparing the relative increase in the metabolic cost of walking when carrying the load on the arms, both when the arms were freely swinging and when they were prevented from swinging.

MATERIALS AND METHODS

Experimental procedures

We tested these hypotheses on 12 subjects (9 men and 3 women, mean±s.d., age 22.0±1.76 years, mass 74.3±17.4 kg, height 1.76±0.12 m), who carried loads on the arms, waist and legs. Our sample size for this study was based on an *a priori* power analysis. We used data published by Soule and Goldman (1969), who compared the rates of oxygen consumption when human subjects walked at ∼1.1 m s^{-1} (4.0 km h^{-1}) without an added load and when carrying 4 kg on each hand (8 kg total). The effect size of 1.74 yielded a relatively low sample size of only 4 subjects. However, we decided to utilize a conservative approach (Arellano et al., 2009), by using an effect size of 0.8. With an effect size of 0.8, a one-tailed test, an alpha error probability equal to 0.05 and a type II error rate equal to or less than 0.20 (i.e. power ≥80%), the power analysis estimated an objective sample size of 12 subjects (G*Power v.3.1). Prior to experimental data collection, each subject read and signed an informed consent document approved by the University of Houston Institutional Review Board. All subjects completed a health screening form to ensure they met the study's participation criteria, wore their own shoes, were healthy and were experienced with treadmill walking.

Subjects visited the laboratory for a single experimental session. The experimenter first measured body segment lengths and then fitted each subject with reflective markers on the lower and upper extremities using a simple body marker set (methods similar to Arellano and Kram, 2014). Following the initial preparation, the subject stood quietly on a dual-belt force-measuring treadmill (Bertec Corporation, Cleveland, OH, USA) for 7 min. During this time, we measured their rates of oxygen consumption ($\dot{V}_{O_2}$) and carbon dioxide production ($\dot{V}_{CO_2}$) using expired gas analysis (ParvoMedics TrueMax2400, Sandy, UT, USA). We also recorded a 5 s standing calibration (12-camera system; Vicon, Oxford, UK) to use as an anatomical reference frame for each segment and as a baseline measure for the ground reaction forces and moments (Arellano and Kram, 2014). For the remaining trials, we simultaneously measured $\dot{V}_{O_2}$ and $\dot{V}_{CO_2}$, ground reaction forces/moments (1000 Hz), and 3D positions of reflective markers (100 Hz) while subjects completed five randomized trials (7 min each) of walking at 1.25 m s^{-1}. The trial conditions consisted of walking without a load (control), with a 4 kg load at the COM of each arm while freely swinging (8 kg total), with a 4 kg load at the COM of each arm while prevented from swinging

(8 kg total), with a 4 kg load on the COM of each leg (8 kg total) and with an 8 kg load around the waist (Fig. 1). To reduce any effects of fatigue, subjects were allowed a full recovery *ad libitum* with at least 5 min between each walking trial.

For the arm and leg conditions, we ensured symmetry in mass but also in location by placing and distributing the load at the COM of each arm or leg, which was calculated relative to each subject's shoulder or hip joint using published anthropometric data tables and formulas (Winter, 1990). Loads were positioned at the location of each subject's arm or leg COM so that the location of the arm or leg COM with the added load remained roughly the same. In general, we found that the COM location for each arm or leg was just proximal to the elbow or knee joint, respectively. It should be noted that using the position of the COM and measuring its distance from the shoulder joint stems from our original approach of modeling the arm and leg as a simple pendulum. Therefore, the experiments were performed by adding the load to the COM, but a more accurate way to test our hypotheses would have been to add the load to a point located at the equivalent length of the arm or leg when it is modeled as a compound pendulum. Following the methods of Browning et al. (2007), the loads consisted of 3.175 mm (0.125 inch) thick, lead rectangular strips wrapped in duct tape, which acted as an interface to prevent direct contact with the subject's skin. In addition, we secured the loads tightly to the arms and legs with athletic prewrap tape to prevent relative motion.

Data analysis

We calculated net metabolic power and the respiratory exchange ratio (RER) from the average $\dot{V}_{O_2}$ and $\dot{V}_{CO_2}$ during the last 3 min of each trial using the Brockway equation (Brockway, 1987). Consistent with previous methods (Arellano and Kram, 2011, 2012, 2014), we filtered the position data of the reflective markers with a 9th order zero-lag low-pass Butterworth filter with a cut-off frequency of 6 Hz. We also filtered the ground reaction force and moment data generated from the right and left foot using a 4th order zero-lag low-pass Butterworth filter at 20 Hz (Snyder and Farley, 2011).

From the filtered position data, we calculated the sagittal plane coordinates of the COM of the right and left arm and leg using published anthropometric data tables (Winter, 1990). Arm and leg angles for the right and left side were measured between a horizontal reference axis and a vector defining the orientation of the arm or leg COM. From the filtered vertical ground reaction force, we determined instances of initial and end foot–ground contact (Arellano and Kram, 2011) by using a 5 N threshold. In particular, instances of initial contact were used to quantify, over 30 consecutive strides, average stride frequency and average peak-to-peak amplitude of arm and leg COM motion. Peak-to-peak amplitude reflects the angular distance swept by the arm and leg COM during a full cycle. For the arm and leg, a full cycle in the sagittal plane takes place when the arm or leg COM initiates forward motion at the instant of maximum retraction,

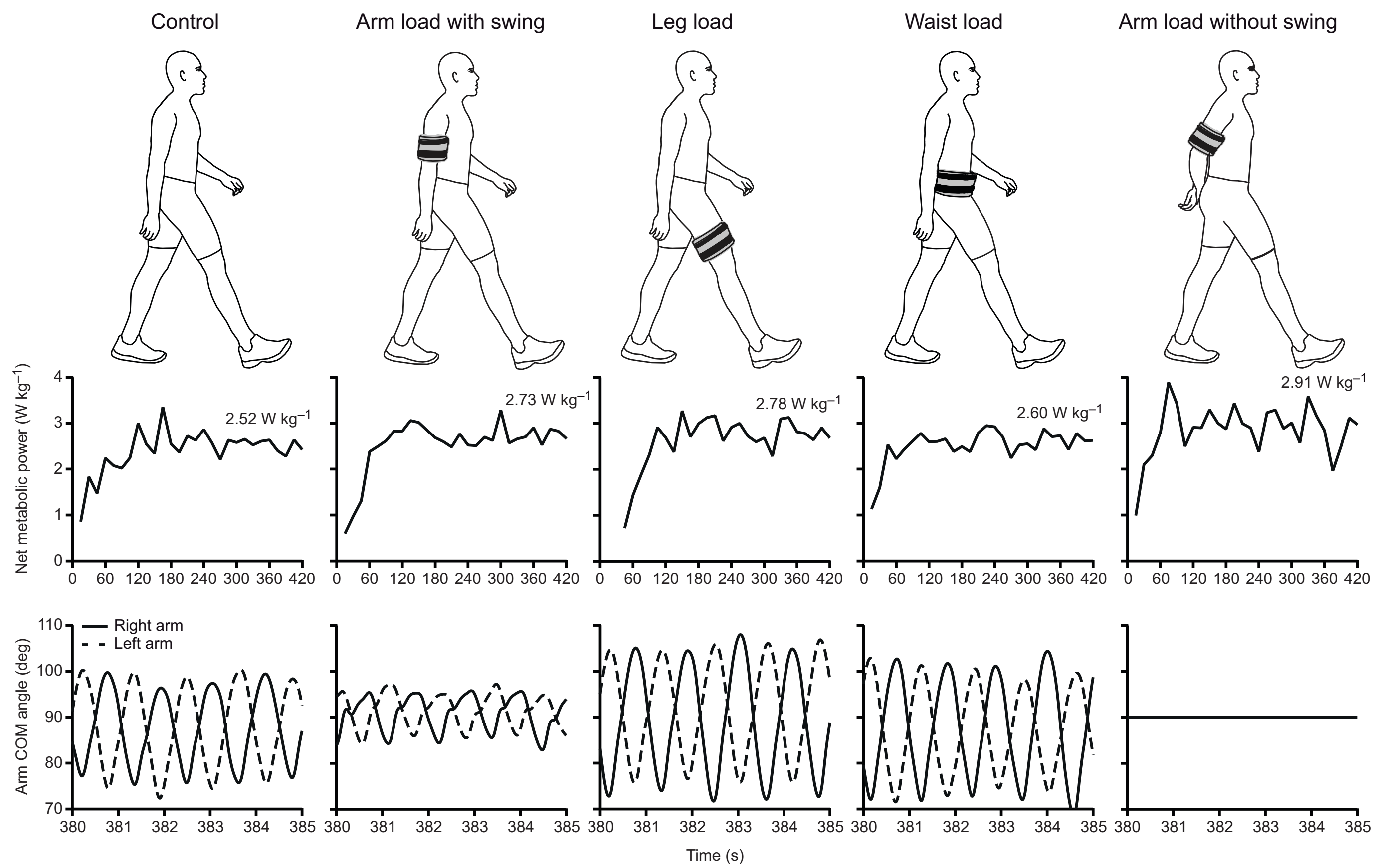

Fig. 1. Net metabolic power and arm swinging dynamics while carrying loads on various parts of the body during human walking. Top: subjects walked on a dual-belt instrumented treadmill while we measured their rates of oxygen consumption ($\dot{V}_{O_2}$) and carbon dioxide production ($\dot{V}_{CO_2}$), 3D position data for markers placed on the upper and lower body, and ground reaction forces and moments. Each subject completed five randomized trials (7 min each) of walking without a load (control), with an 8 kg load around both arms while naturally swinging, with an 8 kg load around both arms while prevented from swinging, with an 8 kg load around the waist and with an 8 kg load around both legs. For the arm and leg load trials, we split the load symmetrically by having subjects carry 4 kg on each arm or leg and by positioning the load at the same distance relative to the shoulder or hip joint. Bottom: representative time-series data (5 s) showing changes in the demand for net metabolic power and arm swinging dynamics across the control (no load) and load-carrying walking conditions.

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then reaches maximum protraction as it moves forward, and then again as it moves back until reaching maximum retraction.

And finally, instances of initial and end foot–ground contact were used to calculate the moment that is exerted on the surface of the treadmill about the vertical axis, which has also been referred to as the free vertical moment (Li et al., 2001; Umberger, 2008). Similar to Collins et al. (2009), the free vertical moment is defined here as the moment the foot exerts on the surface of the treadmill about the vertical axis that extends through the instantaneous center of pressure during the stance phase of walking. In line with the convention of Umberger (2008), negative and positive values signify an internal and external rotation moment, respectively. After inspecting the free vertical moment for the left and right feet, we chose to report the data for the right foot only because the force and moment signals in the medio-lateral direction were inherently noisy for the left foot, leading to unreliable calculations. The origin of the noise was a small corner of a wooden platform that was inadvertently touching the side of the left treadmill.

All computational calculations and descriptive analyses were performed using Matlab (MathWorks, Inc., Natick, MA, USA).

Statistical analysis

For each dependent variable, we performed a repeated measures ANOVA with 'load' as a within-subjects fixed factor and followed that with planned comparisons between normal arm swing without a load (the control) and each load-carrying condition (arm load with swing, arm load without swing, waist load and leg load). We also made planned comparisons between arm load with swing and leg load conditions and between arm load with swing and arm load without swing conditions. For all planned comparisons, we used Dunnett's multiple comparison method and published data table for a one-sided comparison against a control (Dunnett, 1955, 1964), as described previously (Arellano and Kram, 2011). With respect to our arm and leg swing amplitude measures, we performed a preliminary analysis to determine whether significant differences existed between left and right sides. We defined 'side' as a within-subjects fixed factor and found that right and left arm swing (P=0.276) and right and left leg swing (P=0.509) were not statistically different. Therefore, we decided to only perform repeated measures ANOVA for right arm swing and right leg swing amplitudes (P-values reported in Table 1). When feasible, exact P-values are reported within the text; otherwise, statistical significance is signified as P<0.05 or lower. For clarity, we also report statistical significance in Table 1 and in the figures as P<0.05 or lower. Statistical significance was set an alpha value of 0.05 (SPSS, Chicago, IL, USA) and all values are reported as means±s.d. unless otherwise noted.

RESULTS

Effects of load carrying on net metabolic power

As expected, the demand for net metabolic power during walking increased when carrying loads around the arms, legs and waist (Fig. 2A; $F_{4,44}$=33.37; 'load' main effect, P_{load}<0.001; all P<0.001 when each load-carrying condition compared against the control). The demand for net metabolic power while carrying loads on the swinging arms was 9% less than when carrying loads on the legs (P<0.001), but similar to that when carrying the load at the waist (P=0.529). In addition, the demand for net metabolic power while carrying loads around the arms was 7% less when the arms were swinging freely as opposed to not swinging (P=0.001). The RER across all conditions was always <1.0, indicating that metabolic energy was provided primarily by aerobic metabolism. When compared with the control, however, the RER increased under all load-carrying conditions ($F_{4,44}$=4.46, P_{load}=0.014; all P<0.05 when each load-carrying condition compared against the control), indicating a significant shift toward greater carbohydrate utilization when carrying the load.

Effects of load carrying on arm COM motion

During the control (no load) condition (normal walking), the arm COM swept an angular distance of ∼32±9 deg and statistical analyses indicated that carrying loads on the arms, legs and/or waist was coupled with changes in the arm COM motion (Fig. 2B; $F_{4,44}$=75.19, P_{load}<0.001). When compared with control, the angular distance swept by the arm decreased by half, i.e. to ∼16±5 deg, when carrying the same load at the arm COM (P<0.001). When carrying the load at the leg COM, the arm swept an angular distance of 38 deg, representing a 20% increase from walking without a load (P=0.002). The arm swept an angular distance of 30 deg when subjects carried the load at the waist, which was similar in amplitude to walking without a load (P=0.261).

Table 1. Average data for subjects walking without a load (control) and with an 8 kg load attached to the arms, legs and waist

	Control	Arm load with swing	Leg load	Waist load	Arm load without swing
Net metabolic power (W kg^{-1})	2.53±0.28	2.85±0.31 $P_{control}$<0.001 $P_{waist\ load}$=0.529	3.14±0.33 $P_{control}$<0.001 $P_{arm\ load\ with\ swing}$<0.001	2.90±0.32 $P_{control}$<0.001	3.05±0.27 $P_{control}$<0.001 $P_{arm\ load\ with\ swing}$=0.001
RER ($\dot{V}_{CO_2}/\dot{V}_{O_2}$)	0.85±0.04	0.87±0.04 $P_{control}$=0.001	0.88±0.03 $P_{control}$=0.001	0.87±0.05 $P_{control}$=0.016	0.87±0.04 $P_{control}$=0.018
Right arm COM peak-to-peak amplitude (deg)	31.60±8.65	15.79±5.44 $P_{control}$<0.001 $P_{waist\ load}$<0.001	38.19±12.52 $P_{control}$=0.002 $P_{arm\ load\ with\ swing}$<0.001	30.00±10.58 $P_{control}$=0.261	0±0 $P_{control}$<0.001 $P_{arm\ load\ with\ swing}$=0.001
Left arm COM peak-to-peak amplitude (deg)	32.29±9.40	14.81±3.53	41.04±12.23	31.11±9.54	0±0
Right leg COM peak-to-peak amplitude (deg)	39.12±2.18	40.06±2.52	39.39±2.85	40.70±2.39	39.38±6.18
Left leg COM peak-to-peak amplitude (deg)	39.05±2.67	40.18±2.73	39.14±2.85	40.86±2.46	40.92±2.51
Positive peak free vertical moment (%BW×LL)	0.019±0.008	0.018±0.006	0.020±0.004	0.018±0.008	0.018±0.006
Stride frequency (Hz)	0.92±0.04	0.93±0.04	0.91±0.04	0.93±0.04	0.93±0.05

Values are expressed as mean±s.d. and statistical comparisons against a particular condition are defined as $P_{condition}$ equal to or less than the stated numerical value. BW, body weight; LL, leg length.

Note: independent repeated measures ANOVA did not detect a significant main effect for the following variables: right leg COM peak-to-peak amplitude ($F_{4,44}$=0.851, $P_{condition}$=0.386); positive peak free vertical moment ($F_{4,44}$=0.346, $P_{condition}$=0.845); stride frequency ($F_{4,44}$=0.346, $P_{condition}$=0.057). As a significant main effect for load was not detected for any of these variables, we did not perform follow-up comparisons between conditions; thus, P-values are not reported.

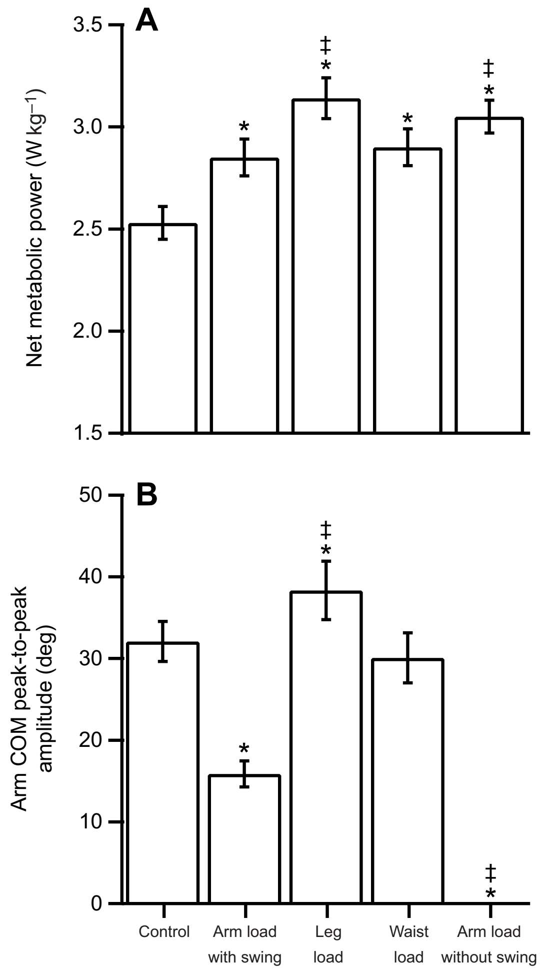

Fig. 2. Carrying an 8 kg load on the swinging arms is relatively inexpensive during human walking. Mean (±s.e.m.) values for (A) metabolic power demand and (B) peak-to-peak amplitude of arm swing (n=12). *Significantly greater or lower than control (no load) condition; ‡significantly greater or lower than arm load with swing condition (P<0.05).

Effects of load carrying on leg COM motion, stride frequency and free vertical moment

The leg COM motion ($F_{4,44}$=0.851, P_{load}=0.386) and stride frequency ($F_{4,44}$=3.47, P_{load}=0.057) remained similar across the control and all load-carrying conditions (Table 1). On average, the leg COM swept an angular distance of ∼40 deg and the legs maintained a stride frequency of 0.92 Hz. In addition, the positive peak values for the free vertical moment (normalized to body weight and leg length) were similar across the control and load-carrying conditions (Fig. 3; $F_{4,44}$=0.346, P_{load}=0.845).

DISCUSSION

Our findings support our first hypothesis that carrying loads on the arm COM is cheaper than carrying loads on the leg COM. The cost of carrying 4 kg on each arm, while swinging freely, was 12.6% greater than the cost of walking without a load. Yet, the cost of carrying the same load on the COM of each leg was 24% greater than the cost of walking without a load, nearly doubling the percentage increase in cost. Our findings do not support our second hypothesis, which predicted that carrying loads on the swinging arms would be more expensive than carrying loads on the waist. Instead, we found that carrying loads at the arm COM was just as costly as carrying the same total load at the waist, but this only holds true when the arms were allowed to swing freely.

Our data indicate that, when compared with carrying loads at other locations on the body, carrying modest loads at the arm COM while swinging freely is relatively cheap. It has been shown, in both bipedal and quadrupedal animals, that placing loads near the body's COM is one of the cheapest ways of carrying loads on the body (Browning et al., 2007; Royer and Martin, 2005; Steudel, 1990). The most expensive means of load carrying that we investigated was placing 4 kg at each leg's COM. The 24% increase in the net metabolic cost of walking observed here is similar to the 25% increase reported by Browning et al. (2007), where 4 kg loads (8 kg total) were placed on each thigh. Carrying the same load on the waist increased metabolic cost by ∼15%, which was similar to the increase in cost of ∼13% when carrying the same load on the freely swinging arms. Restricting the arms from swinging with the added load led to an ∼21% increase in cost. Therefore, we infer that as a load-carrying method, placing moderate loads on the swinging arms is just as good as placing loads on the waist.

Although carrying loads on the legs is expensive, we found that carrying loads on the arms is not. One explanation for the difference in load-carrying cost between the legs and arms is that the swinging amplitude of the leg COM was conserved, while the swinging amplitude of the arm COM was not. Leg COM motion and stride frequency remained the same when carrying the load on the arms,

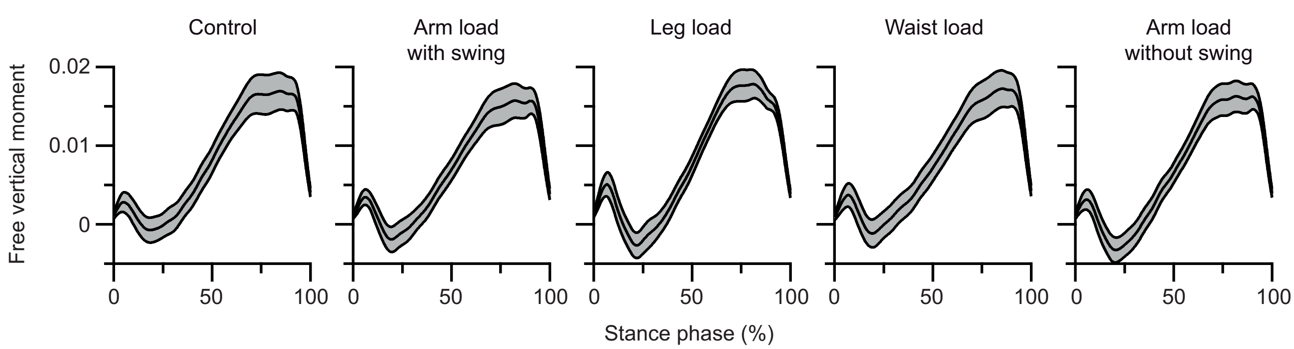

Fig. 3. Free moment exerted by the foot on the surface of the treadmill across control and load-carrying conditions. The free vertical moment (body weight×leg length) was calculated from the forces and moments applied by the right foot for 30 consecutive steps during walking (n=12). Centre lines and shaded regions represent the mean and standard error of the mean envelope, respectively. The general trend for the applied free vertical moment was similar across all conditions, with a short period of a small internal rotation moment followed by a large external rotation moment during mid-to-late stance. Statistical comparisons against the control (no load) condition did not reveal differences in peak external rotation moments when loads were carried on the arms, legs or waist. Following the convention of Umberger (2008), a negative value signifies an internal rotation moment and vice versa.

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waist or legs. However, when compared with walking without a load, the angular distance swept by the arm decreased by 50% when carrying the load at the arm COM. This raises the question: why conserve leg COM motion and not arm COM motion? One possible reason is that humans tend to exploit the passive pendulum dynamics that govern arm swinging motion during walking (Collins et al., 2009; Pontzer et al., 2009). One way to achieve this is by allowing the pendulum-like arm to be driven in part by the back and forth motion arising at the shoulder joint (Arellano et al., 2012), which would not exact a metabolic cost. While we found that carrying loads close to the arm COM is relatively cheap, it does reduce the arm's swinging amplitude. This observation is line with the passive arm swing model proposed by Pontzer et al. (2009), where the arms are treated as an auxiliary mass that acts to reduce torso rotation and thus shoulder translation. Therefore, it is possible that adding a 4 kg load to each arm as in this study had a similar effect on torso rotation. A reduction in torso rotation, we predict, would decrease the back and forth motion arising at the shoulder joint, which would be coupled with a lower driving force and would explain why we observed a decrease in the swinging amplitude of the arm COM with the added load.

From an angular momentum perspective (Herr and Popovic, 2008), a reduction in arm swinging amplitude as a result of added mass may have had little effect on the arm's ability to counteract the angular momentum generated by the swinging legs about the vertical axis. As the arms gain mass, they can swing less and still be effective in regulating whole-body angular momentum during walking. Given that we added a fixed 4 kg load to every subject's arm and that the body mass of our subjects varied by more than 30 kg, this will represent a different fraction of arm mass that was added by the load. Therefore, we would expect that as the fraction of the arm's mass increases, the arms will swing with lower amplitude. Indeed, a simple regression analysis reveals a modest, but positive correlation (Pearson's r=0.55) between the two variables (Fig. 4), indicating that an increase in the fraction of the arm's mass was coupled with greater reductions in arm swinging amplitude. Even though this was the case, we found that when the arms carried a 4 kg load, they still swung with an average peak-to-peak amplitude of

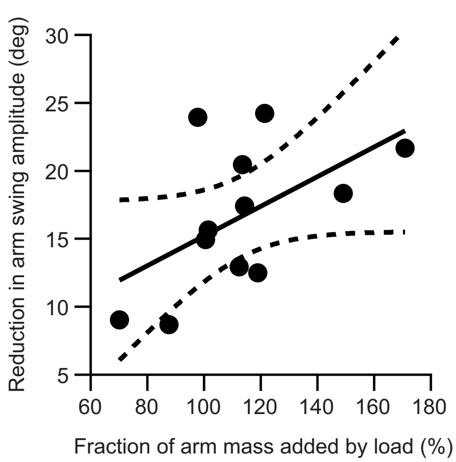

Fig. 4. Relationship between fraction of arm mass added by load and reduction in arm swing amplitude. A positive correlation (Pearson's r=0.55) indicates a tendency for greater reductions in arm swing amplitude to occur with greater fractions of arm mass added by the load. Data were fitted with a linear least-squares regression analysis: y=4.31x+0.11, r^2=0.30, P=0.03. The black solid line represents the best fit line and black dashed lines represent 95% confidence bands.

roughly 15 deg. Regardless of the underlying reason for the reduction in arm swinging amplitude and its effect on regulating whole-body angular momentum, the cost of walking with an added load at the waist (14.6%) was similar to the cost of walking with the added load on the swinging arms (12.6%), suggesting that swinging the added load did not exact a metabolic cost. This observation leads us to conclude that the arm's passive pendulum dynamics dominated the swinging motion of the added load.

An unexpected observation was that while adding loads to the legs did not alter the leg COM motion, it did alter the arm COM motion. When adding loads to the legs, the left and right arm COM swept an angular distance of roughly 40 deg on average, reflecting an 8 deg increase from the control condition. This might reflect a compensatory strategy that can be explained from the perspective of an angular momentum framework (Herr and Popovic, 2008). Swinging the arms with greater amplitude would help to counterbalance the angular momentum generated by the swinging legs, as it is expected that when the legs swing with an added load, the legs generate greater angular momentum about the vertical axis. Therefore, it seems reasonable to think that swinging the arms with greater amplitude was an attempt to regulate whole-body angular momentum during walking (Herr and Popovic, 2008).

Regulating whole-body angular momentum via the swinging arms may have helped minimize the cost of carrying loads on the legs. Minimizing the cost of carrying loads on the legs would have required subjects to correctly tune their arm swinging amplitude in an attempt to exploit a possible trade-off (Collins et al., 2009; De Graaf et al., 2019) between the cost incurred to swing the arms with greater amplitude and the reduction in cost that arises from minimizing the free vertical moment that the feet exert along the ground during each step (Park, 2008). As pointed out by others (De Graaf et al., 2019; Ferris et al., 2006), the free vertical moment that the feet exert along the ground is related to the angular momentum generated about the vertical axis. As the arms swing, they generate angular momentum about the vertical axis that is roughly equal but opposite in direction to that of the swinging legs. This balancing of upper and lower body angular momentum helps minimize whole-body rotation, which can be observed by a reduction in the magnitude of the free vertical moment that the foot exerts on the ground. It has been proposed that reducing the free vertical moment that the foot exerts on the ground reduces metabolic cost (Collins et al., 2009), as this minimizes the need for the leg muscles to generate torques that resist the reaction moments that the ground exerts back on to foot. Therefore, minimizing the cost of carrying loads on the legs would require that the cost of swinging the arms with greater amplitude is outweighed by the savings in cost that come from minimizing free vertical moments. If subjects did attempt and succeed in exploiting this trade-off, then this might explain why the free vertical moment was similar when walking without a load and when walking with leg loads (Fig. 3). Although speculative, our findings hint at the possibility that increasing arm swinging amplitude was a compensatory strategy that helped minimize the cost of carrying loads on the legs. While we were careful in our experimental design, we now recognize that we lacked a condition where subjects walked with leg loads while the arms were prevented from swinging. We also lacked a condition where equal loads were carried on the arms and legs. Such conditions would have allowed us to understand whether increasing arm swinging amplitude helped to minimize free vertical moments. In the absence of arm swing and its mechanical effect on the body during walking, we would expect that the feet would exert a greater free vertical moment when carrying loads on the legs, supporting

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the notion that the increase in arm swinging amplitude was a compensatory strategy to minimize metabolic cost.

In summary, we found that attaching loads to freely swinging arms provides a relatively cheap means to carrying loads while walking. Carrying the load around the swinging arms yields a similar cost to carrying loads around the waist; however, preventing the arms from swinging negates those savings.

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Competing interests
The authors declare no competing or financial interests.

Author contributions
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References
Arellano, C. J. and Kram, R. (2011). The effects of step width and arm swing on energetic cost and lateral balance during running. *J. Biomech.* **44**, 1291-1295. doi:10.1016/j.jbiomech.2011.01.002

Arellano, C. J. and Kram, R. (2012). The energetic cost of maintaining lateral balance during human running. *J. Appl. Physiol.* **112**, 427-434. doi:10.1152/japplphysiol.00554.2011

Arellano, C. J. and Kram, R. (2014). The metabolic cost of human running: is swinging the arms worth it? *J. Exp. Biol.* **217**, 2456-2461. doi:10.1242/jeb.100420

Arellano, C. J., O'Connor, D. P., Layne, C. and Kurz, M. J. (2009). The independent effect of added mass on the stability of the sagittal plane leg kinematics during steady-state human walking. *J. Exp. Biol.* **212**, 1965-1970. doi:10.1242/jeb.026153

Arellano, C. J., Beale, T. B. and Kram, R. (2012). Arm swing during human walking: active and passive contributions to a hybrid system. In Dynamic Walking Conference Proceedings. Pensacola Beach, Florida.

Ballesteros, M. L. F., Buchthal, F. and Rosenfalck, P. (1965). The pattern of muscular activity during the arm swing of natural walking. *Acta Physiol. Scand.* **63**, 296-310. doi:10.1111/j.1748-1716.1965.tb04069.x

Baudinette, R. V. and Biewener, A. A. (1998). Young wallabies get a free ride. *Nature* **395**, 653-654. doi:10.1038/27111

Brockway, J. M. (1987). Derivation of formulae used to calculate energy expenditure in man. *Hum. Nutr. Clinl. Nutr.* **41**, 463-471.

Browning, R. C., Modica, J. R., Kram, R. and Goswami, A. (2007). The effects of adding mass to the legs on the energetics and biomechanics of walking. *Med. Sci. Sports Exerc.* **39**, 515-525. doi:10.1249/mss.0b013e31802b3562

Collins, S. H., Adamczyk, P. G. and Kuo, A. D. (2009). Dynamic arm swinging in human walking. *Proc. Biol. Sci.* **276**, 3679-3688. doi:10.1098/rspb.2009.0664

De Graaf, M. L., Hubert, J., Houdijk, H. and Bruijn, S. M. (2019). Influence of arm swing on cost of transport during walking. *Biol. Open* **8**, bio039263. doi:10.1242/bio.039263

Den Hertog, J. P. (1985). *Mechanical Vibrations*. New York: Dover.

Dunnett, C. W. (1955). A multiple comparison procedure for comparing several treatments with a control. *J. Am. Stat. Assoc.* **50**, 1096-1121. doi:10.1080/01621459.1955.10501294

Dunnett, C. W. (1964). New tables for multiple comparisons with a control. *Int. Biomet. Soc.* **20**, 482-491. doi:10.2307/2528490

Ferris, D. P., Huang, H. J. and Kao, P-C. (2006). Moving the arms to activate the legs. *Exerc. Sport Sci. Rev.* **34**, 113-120. doi:10.1249/00003677-200607000-00005

Goudriaan, M., Jonkers, I., Van Dieen, J. H. and Bruijn, S. M. (2014). Arm swing in human walking: what is their drive? *Gait Posture* **40**, 321-326. doi:10.1016/j.gaitpost.2014.04.204

Herr, H. and Popovic, M. (2008). Angular momentum in human walking. *J. Exp. Biol.* **211**, 467-481. doi:10.1242/jeb.008573

Huang H. J. and Ferris D. P. (2004). Neural coupling between upper and lower limbs during recumbent stepping. *J. Appl. Physiol.* **97**, 1299-1308. doi:10.1152/japplphysiol.01350.2003

Jackson, K. M., Joseph, J. and Wyard, S. J. (1978). A mathematical model of arm swing during human locomotion. *J. Biomech.* **11**, 277-289. doi:10.1016/0021-9290(78)90061-1

Kram, R. (1991). Carrying loads with springy poles. *J. Appl. Physiol.* **71**, 1119-1122. doi:10.1152/jappl.1991.71.3.1119

Li, Y., Wang, W., Crompton, R. H. and Gunther, M. M. (2001). Free vertical moments and transverse forces in human walking and their role in relation to arm-swing. *J. Exp. Biol.* **204**, 47-58.

Maloiy, G. M. O., Heglund, N. C., Prager, L. M., Cavagna, G. A. and Taylor, C. R. (1986). Energetic cost of carrying loads: have African women discovered an economic way? *Nature* **319**, 668-669. doi:10.1038/319668a0

Park, J. (2008). Synthesis of natural arm swing motion in human bipedal walking. *J. Biomech.* **41**, 1417-1426. doi:10.1016/j.jbiomech.2008.02.031

Pontzer, H., Holloway, J. H., , IV, Raichlen, D. A. and Lieberman, D. E. (2009). Control and function of arm swing in human walking and running. *J. Exp. Biol.* **212**, 523-534. doi:10.1242/jeb.024927

Rome, L. C., Flynn, L. and Yoo, T. D. (2006). Biomechanics: rubber bands reduce the cost of carrying loads. *Nature* **444**, 1023-1024. doi:10.1038/4441023a

Rose, J., Ralston, H. J. and Gamble, J. G. (1994). Energetics of walking. In *Human Walking* (ed. J. Rose and J. G. Gamble), pp. 54-55. Baltimore, MD: Willaims and Wilkins.

Royer, T. D. and Martin, P. E. (2005). Manipulations of leg mass and moment of inertia: effects on energy cost of walking. *Med. Sci. Sports Exerc.* **37**, 649-656. doi:10.1249/01.MSS.0000159007.56083.96

Snyder, K. L. and Farley, C. T. (2011). Energetically optimal stride frequency in running: the effects of incline and decline. *J. Exp. Biol.* **214**, 2089-2095. doi:10.1242/jeb.053157

Soule, R. G. and Goldman, R. F. (1969). Energy cost of loads carried on the head, hands, or feet. *J. Appl. Physiol.* **27**, 687-690. doi:10.1152/jappl.1969.27.5.687

Steudel, K. (1990). The work and energetic cost of locomotion. I. The effects of limb mass distribution in quadrupeds. *J. Exp. Biol.* **154**, 273-285.

Umberger, B. R. (2008). Effects of suppressing arm swing on kinematics, kinetics, and energetics of human walking. *J. Biomech.* **41**, 2575-2580. doi:10.1016/j.jbiomech.2008.05.024

Winter, D. A. (1990). *Biomechanics and Motor Control of Human Movement*. New Work: John Wiley & Sons, Inc.