Allowing the Load to Swing Reduces the Mechanical Energy of the Stance Leg and Improves the Lateral Stability of Human Walking

Lianxin Yang, Yuning Xu, Kuangen Zhang, Ken Chen, *Member, IEEE*, and Chenglong Fu, *Member, IEEE*

***Abstract*— Loaded walking with typical rigid backpack results in a significant increase in the mechanical energy of the stance leg and a decrease in lateral stability. Allowing the load to swing, which has been applied in shoulder pole, a tool widely used in Asia for load carriage assistance, may attenuate these effects. This paper theoretically analyzes and experimentally validates the biomechanical and energetic effects of the swinging loads. When walking with a 30 kg load, allowing the load to swing reduces the fore-aft leg impulses by over 19% and further reduces the mechanical energy of the stance leg by 12.9% compared to the typical rigid backpack. The whole-body metabolic cost has no significant change, which may be attributed to the increase in the muscle work of the upper body and the leg swing. Moreover, the load movement out of phase to the human in the lateral direction reduces the lateral excursion of extrapolated center-of-mass by 27.2%, indicating an increase in the lateral margin of stability and implying an improvement in lateral stability. The results demonstrate that allowing the load to swing reduces the horizontal leg impulses and the mechanical energy of the stance leg, and improves the lateral stability of human walking.**

***Index Terms*— Load carriage, swing, human walking, biomechanics, energetic cost.**

I. Introduction

CARRYING loads with a backpack is a common and important task in human society. However, the additional load mass results in a significant increase in ground reaction forces (GRF) [1], metabolic cost [2], muscle activity [3], and fatigue [4], [5], which can increase the risk of musculoskeletal injuries [6], and increases step width variability [7], perhaps as an indication of increased balance demands of human walking. Aiming at attenuating these effects, people have developed a variety of load carriage tools: elastically suspended backpacks [8]–[13] regulate the temporal distribution of load pressure induced by the vertical acceleration of loads; exoskeletons and exosuits [14]–[19] provide additional torque to human joints; and supernumerary robotic limbs [20]–[22] transfer the forces caused by the load mass directly to the ground.

The elastically suspended backpack regulating the vertical load movement passively [8] or actively [11] has been shown to reduce the metabolic cost of loaded walking by 6.2% or 8.02%, respectively, by minimizing the vertical acceleration of the load, which further affects the vertical GRFs [23]–[25]. Although the horizontal GRFs are relatively low compared to the vertical forces, they account for 47% of the walking's metabolic cost in walking [26], [27], indicating that external horizontal force may prominently affect the energetic cost. Besides, external lateral forces that are out of phase with the lateral displacement of the human center of mass (CoM) have been shown to improve the lateral stability and the energy efficiency of walking [28], [29]. Moreover, the backpack with load compliance in the lateral direction via an inverted pendulum mechanism allows the load to oscillate out of phase with the carrier laterally, and reduce the peak GRFs while walking [30]. Overall, the horizontal relative load movement should result in different interaction forces between the load and the carrier compared to the typical rigidly-attached backpack, further affecting the biomechanics and energetics of loaded walking.

The spherical pendulum is a simple structure that allows for horizontal load movement relative to the human CoM. The swing motion can be generated naturally by the stimulation of the periodic oscillation of the human CoM, such as a ponytail [31], [32], or arm swing motion [33]–[35]. In contrast to the forced vibration of the spring-mass-damper system in the present elastically suspended backpacks, the forced swing

Manuscript received September 24, 2020; revised December 28, 2020 and January 23, 2021; accepted January 25, 2021. Date of publication January 29, 2021; date of current version March 2, 2021. This work was supported in part by the National Natural Science Foundation of China under Grant U1913205, in part by the National Key Research and Development Program of China under Grant 2018YFC2001601, in part by the Guangdong Innovative and Entrepreneurial Research Team Program under Grant 2016ZT06G587, in part by the Shenzhen and Hong Kong Innovation Circle Project under Grant SGLH20180619172011638, and in part by the Centers for Mechanical Engineering Research and Education at the Massachusetts Institute of Technology (MIT) and the Southern University of Science and Technology (SUSTech). *(Corresponding author: Chenglong Fu.)*

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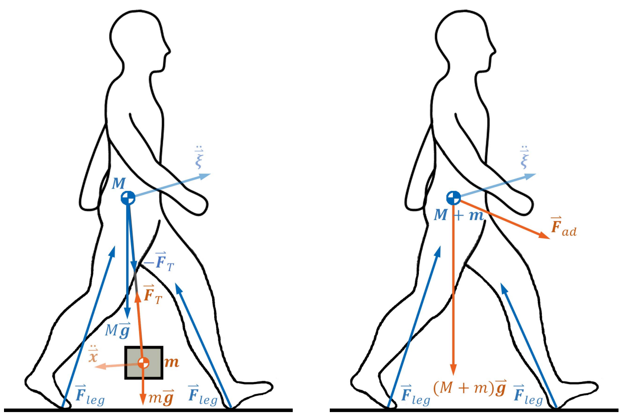
Digital Object Identifier 10.1109/TNSRE.2021.3055624



(a) (b)

Fig. 1. Model of human walking with the swinging load. (a) Force analysis of humans and loads. (b) Equivalent system of walking with the rigidly-attached load and an additional force caused by the relative movement of the load.

of the load in the spherical pendulum is unaffected by the load, indicating better applicability.

The structure has been applied in the shoulder pole, a tool developed for load carriage assistance and widely used in Asia. However, the present research about the shoulder pole [36]–[38] mainly focused on the pole's compliance and its effects on the vertical forces of walking and neglected the load swing motion. Moreover, some recent research considered the swing motion in the shoulder pole. Li *et al.* [39] discussed the influences of the structural parameters and predicted the interaction between the pole and the carrier via simulation. The work was enlightening. However, they did not directly analyze the influence on gait energetics. On the other hand, Schroeder *et al.* [40] proposed a two-dimensional trajectory optimization model to determine the energetic consequences and predict the gait adaptation. The work interpreted the reduction in the energetic cost and predicted the changes in step frequency. However, they did not discuss influences on lateral stability nor the effects of the swing motion coupled with elasticity. Inspired by these researches, we aim to figure out the role of the 3-dimensional load swing motion, including its influence on the GRFs, energetics and lateral stability of human walking.

Our study theoretically analyzes the effects of the swinging loads on the biomechanics and energetics of loaded walking in the fore-aft direction and medio-lateral direction. Experiments are conducted to validate the predictions that allowing the load to swing reduces the horizontal GRFs and the mechanical energy of the stance legs, and improves the lateral stability of human walking.

II. Hypothesis

A. Dynamics of Load Swing Motion

The 3-dimensional spherical pendulum model in Fig. 1a is adopted to estimate the load swing motion, where the load is connected to the human body with a weightless inextensible cable. The load movement $\boldsymbol{x}$ is thus stimulated by the periodic movement of the human body CoM $\boldsymbol{\xi}$.

Analyzing the forces exerted on the load, we have

$$m\ddot{\boldsymbol{x}} = \boldsymbol{F_T} + m\boldsymbol{g} \tag{1}$$

where m is the load mass, $\boldsymbol{F_T}$ is the force of cable exerted on the load, and $\boldsymbol{g}$ is the gravitational acceleration.

Defining the relative load movement $\boldsymbol{x_r} = \boldsymbol{x} - \boldsymbol{\xi}$, the cable force can be expressed as

$$\boldsymbol{F_T} = -\frac{1}{R} F_T(t)\boldsymbol{x_r} \tag{2}$$

where R is the cable length, and $F_T(t)$ is the value of the time-dependent cable tension.

Therefore (1) can be written as

$$\ddot{\boldsymbol{x}}_{\boldsymbol{r}} = -\frac{1}{mR} F_T(t)\boldsymbol{x_r} - \ddot{\boldsymbol{\xi}} + \boldsymbol{g}. \tag{3}$$

Assuming the cable is always under tension and never goes slack, i.e., $F_T(t) > 0$, the relative velocity is always perpendicular to the cable, that is, $\dot{\boldsymbol{x}}_{\boldsymbol{r}} \cdot \boldsymbol{x_r} = 0$. Then we have

$$\ddot{\boldsymbol{x}}_{\boldsymbol{r}} \cdot \boldsymbol{x_r} + \dot{\boldsymbol{x}}_{\boldsymbol{r}} \cdot \dot{\boldsymbol{x}}_{\boldsymbol{r}} = 0. \tag{4}$$

Combining (3) and (4), the cable tension can be calculated as

$$F_T(t) = \frac{m}{R}[(\boldsymbol{g} - \ddot{\boldsymbol{\xi}}) \cdot \boldsymbol{x_r} + \dot{\boldsymbol{x}}_{\boldsymbol{r}} \cdot \dot{\boldsymbol{x}}_{\boldsymbol{r}}]. \tag{5}$$

Thus the equation of the relative motion can be written as

$$\ddot{\boldsymbol{x}}_{\boldsymbol{r}} = -\frac{1}{R^2}[(\boldsymbol{g} - \ddot{\boldsymbol{\xi}}) \cdot \boldsymbol{x_r} + \dot{\boldsymbol{x}}_{\boldsymbol{r}} \cdot \dot{\boldsymbol{x}}_{\boldsymbol{r}}]\boldsymbol{x_r} + (\boldsymbol{g} - \ddot{\boldsymbol{\xi}}) \tag{6}$$

from which we can infer that the load movement is unaffected by the load mass.

With the acceleration of the given periodic base excitation $\ddot{\boldsymbol{\xi}}$ and the initial state $(\boldsymbol{x_0}, \dot{\boldsymbol{x}}_{\boldsymbol{0}})$, we can derive the numerical solution to (6) and therefore predict the load swing motion.

B. Human Body CoM Acceleration

Assuming the movement of human body CoM when walking with the swinging load and with the rigidly-attached load are the same as normal walking, we can derive the human body CoM acceleration from the GRFs of normal walking.

Considering a foot that is on the ground from time $t = -\tau/2$ to $t = \tau/2$, where τ is the duration of the ground contact of each foot, the pattern of the three-dimensional GRFs $\boldsymbol{F}_{\mathbf{leg}}$ during this interval can be approximated by the major components of the Fourier series [41], [42]:

$$F_{\mathrm{x}} = p_{\mathrm{x}}[\sin(2\pi t/\tau) - q_{\mathrm{x}}\sin(4\pi t/\tau)]Mg \tag{7}$$

$$F_{\mathrm{y}} = p_{\mathrm{y}}[\cos(\pi t/\tau) - q_{\mathrm{y}}\cos(3\pi t/\tau)]Mg \tag{8}$$

$$F_{\mathrm{z}} = p_{\mathrm{z}}[\cos(\pi t/\tau) - q_{\mathrm{z}}\cos(3\pi t/\tau)]Mg \tag{9}$$

where $F_{\mathrm{x}}, F_{\mathrm{y}}, F_z$ are the GRFs in the fore-aft, medio-lateral and vertical direction, respectively, and M is the body mass. The parameters are taken as $p_{\mathrm{x}} = 0.2$, $p_{\mathrm{y}} = 0.08$, $p_{\mathrm{z}} = 1.15$, $q_{\mathrm{x}} = 0.3$, $q_{\mathrm{y}} = 0.35$, $q_{\mathrm{z}} = 0.4$, to fit the profile of GRFs in normal walking.

Moreover, as the duration of the double-support phase is around $0.2T$, where T is the period of step cycle, we have $\tau = T + 0.2T = 1.2T$. Thereby the approximated GRFs of the two sides when $T = 0.6$ s can be shown as in Fig. 2.

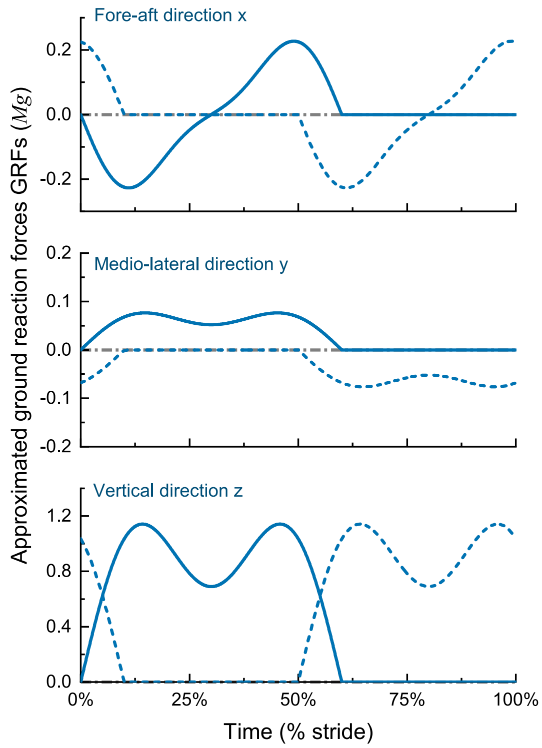


Fig. 2. Approximated GRFs of three directions versus time in a stride cycle. The GRFs are approximated with the major components of the Fourier series.

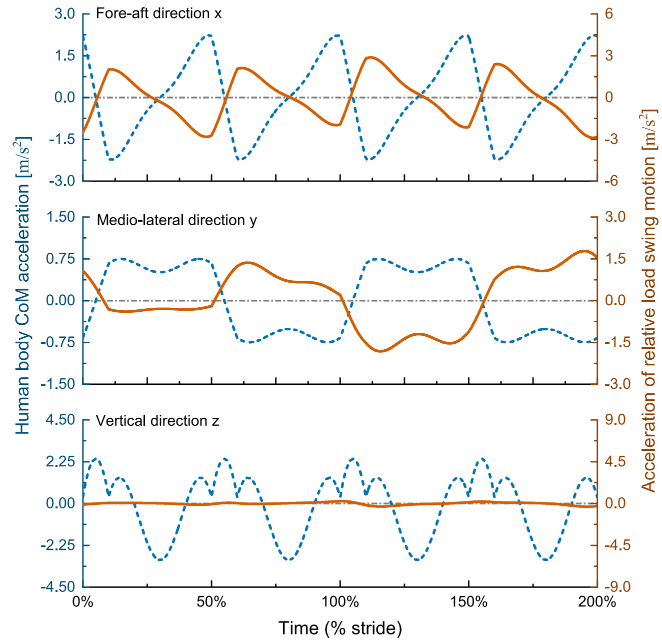


Fig. 3. Approximated human body CoM acceleration $\ddot{\xi}$ and predicted acceleration of the relative load swing motion $\ddot{x}_r$ versus time. $\ddot{x}_r$ is out of phase to $\ddot{\xi}$ in the fore-aft and medio-lateral directions, and keeps at a low level in the vertical direction.

The human body CoM acceleration $\ddot{\boldsymbol{\xi}}$ in normal walking can be derived from GRFs of both legs and gravity:

$$\ddot{\boldsymbol{\xi}} = \Sigma \boldsymbol{F}_{\mathbf{leg}}/M + \boldsymbol{g} \tag{10}$$

Thereby, the acceleration of the relative load swing motion $\ddot{\boldsymbol{x}}_{\boldsymbol{r}}$ with a cable length of $R = 1$ m and initial state of $\boldsymbol{x_0} = \mathbf{0}, \dot{\boldsymbol{x}}_{\mathbf{0}} = \mathbf{0}$ could be further predicted numerically with (6) and (10) using ode45, as shown in Fig. 3. The acceleration of the relative load swing motion $\ddot{\boldsymbol{x}}_{\boldsymbol{r}}$ is out of phase to the human body CoM acceleration $\ddot{\boldsymbol{\xi}}$ in the fore-aft and medio-lateral directions, and keeps at a low level in the vertical direction.

C. Effects of the Swinging Load

Analyzing the forces exerted on human CoM as shown in Fig. 1a, we have

$$M\ddot{\boldsymbol{\xi}} = \Sigma \boldsymbol{F_{leg}} - \boldsymbol{F_T} + M\boldsymbol{g} \tag{11}$$

where M is the human body mass, and $\boldsymbol{F_{leg}}$ are the leg forces.

Adding up (1) and (11), we have

$$(M+m)\ddot{\boldsymbol{\xi}} = \Sigma \boldsymbol{F_{leg}} + \boldsymbol{F_{ad}} + (M+m)\boldsymbol{g} \tag{12}$$

where

$$\boldsymbol{F_{ad}} = -m\ddot{\boldsymbol{x}}_{\boldsymbol{r}} = -m(\ddot{\boldsymbol{x}} - \ddot{\boldsymbol{\xi}}) \tag{13}$$

indicating an equivalent additional force caused by the relative movement of loads compared to the rigidly-attached backpack.

It can be inferred from (12) that the system of human walking with the swinging load is equivalent to walking with the rigidly-attached load and an additional force $\boldsymbol{F_{ad}}$, as shown in Fig. 1, whereas $\boldsymbol{F_{ad}} = \mathbf{0}$ for the zero relative load movement in the typical rigid backpack. Therefore the biomechanical and energetic effects of the swinging load are all caused by this additional force $\boldsymbol{F_{ad}}$ compared to the typical rigid backpack.

We have claimed that the acceleration of the relative load swing motion $\ddot{\boldsymbol{x}}_{\boldsymbol{r}}$ is out of phase to the human body CoM acceleration $\ddot{\boldsymbol{\xi}}$ in the fore-aft and medio-lateral directions. Therefore, with the definition of the additional force $\boldsymbol{F_{ad}}$ in (13), $\boldsymbol{F_{ad}}$ are in phase with the summed GRFs in the fore-aft and medio-lateral directions, and further reduce the required summed leg forces to induce the unchanged human CoM trajectory in this two directions. Assuming the changes in GRFs of two legs caused by $\boldsymbol{F_{ad}}$ have the same ratio between the vertical GRFs of two sides, we can predict the GRFs of walking with the swinging load, as shown in Fig. 4, giving an example of the predicted GRFs when mass ratio $m/M = 0.25$ and step frequency $f = 1.7$ Hz. Simulation results show that allowing the load to swing would reduce the leg impulses and peak leg forces in the fore-aft and medio-lateral direction, and have little effect on the vertical GRFs. The reduction rates of the fore-aft and lateral leg impulses are positively correlated to the mass ratio of load to human, as shown in Fig. 5.

Moreover, the reduction in the fore-aft and lateral GRFs would further affect the mechanical energy of the stance leg, which could be estimated by the CoM work of individual legs. The instantaneous mechanical power of leg forces was

$$P_{mech} = \boldsymbol{F_{leg}} \cdot \dot{\boldsymbol{\xi}}. \tag{14}$$

where $\boldsymbol{F_{leg}}$ is the GRF of the individual leg, and $\dot{\boldsymbol{\xi}}$ is the velocity of human body CoM, obtained by the integral of the acceleration $\ddot{\boldsymbol{\xi}}$ and the walking speed.

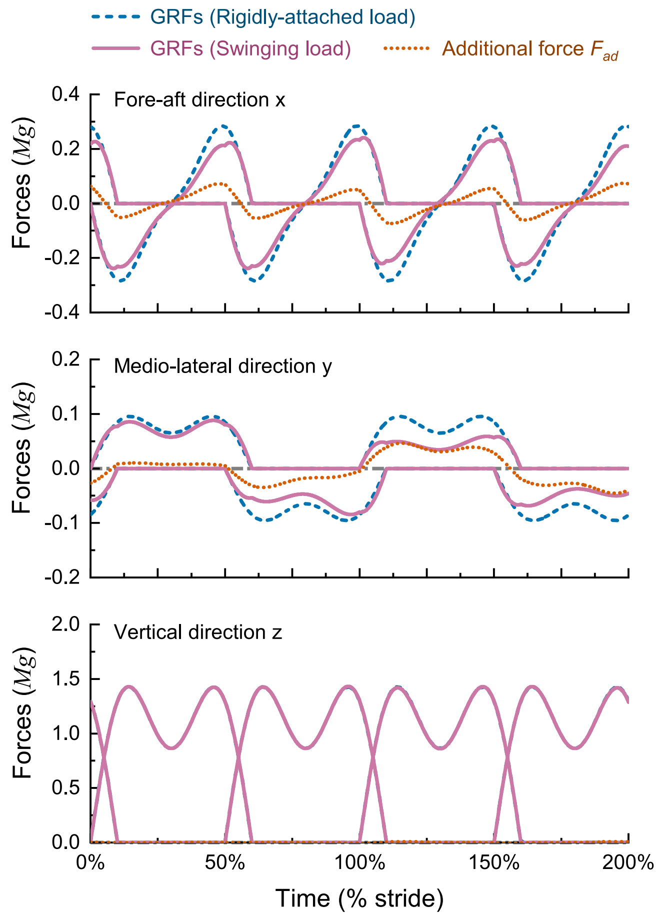


Fig. 4. Additional forces $\boldsymbol{F_{ad}}$ induced by swing motion and GRFs of two legs with rigidly-attached loads and swinging loads versus time. Compared to the rigidly-attached load, allowing the load to swing is predicted to reduce the leg impulses and peak leg forces in the fore-aft and medio-lateral direction. There is little change in the vertical GRFs.

Positive and negative work were performed alternately on human CoM corresponding to different efficiency of muscles η, where $\eta = 25\%$ when muscles perform positive work and $\eta = -120\%$ for negative work [43]. Considering the efficiency of muscles performing mechanical work, the mechanical energy of the stance legs over a step was estimated as

$$W = \Sigma \int_0^T \boldsymbol{F_{leg}} \cdot \dot{\boldsymbol{\xi}}/\eta dt \tag{15}$$

where muscle efficiency

$$\eta = \begin{cases} 25\% & \boldsymbol{F_{leg}} \cdot \dot{\boldsymbol{\xi}} > 0 \\ -120\% & \boldsymbol{F_{leg}} \cdot \dot{\boldsymbol{\xi}} < 0 \end{cases} \tag{16}$$

and T was the time interval of a step.

The stance legs cost of transport (CoT) is defined as (energy cost)/(body weight × distance traveled) and can be calculated as

$$\mathrm{CoT}_{leg} = \frac{W}{MgvT} \tag{17}$$

where v is the walking speed.

The change in the mechanical energy of the stance legs induced by the swinging load could then be predicted, as shown in Fig. 6. The mechanical energy of stance legs is predicted to be reduced by the swinging load, indicating an improvement in the energy efficiency of walking. Moreover, the CoT decreases with added mass until the load is about 50% of body mass, implying that allowing the load to swing is more effective when carrying heavy loads. The results justify the use of very large loads in previous studies [40] (30% and 50% of body mass) based on reports of farmworkers carrying incredibly heavy loads in the fields.

In addition to the biomechanics and energetics, lateral stability is another critical index evaluating the performance of human walking, which could be indicated by the dynamic lateral margins of stability (MoS). The lateral MoS [44], [45] is defined as the minimum lateral distance between the boundary of the base of support (BoS) and the system's extrapolated CoM (xCoM), which is calculated as the position of the vertical projection of the CoM plus its velocity times a factor of the ratio of L/g [46]:

$$x_{\mathrm{y}}^{(\mathrm{ex})} = x_{\mathrm{y}}^{\mathrm{sys}} + \dot{x}_{y}^{\mathrm{sys}} * L/g \tag{18}$$

where L is the human leg length and $x_{\mathrm{y}}^{\mathrm{sys}}$ is the lateral displacement of system CoM:

$$x_{\mathrm{y}}^{\mathrm{sys}} = (m x_{\mathrm{y}} + M\xi_{\mathrm{y}})/(m + M). \tag{19}$$

The minimum lateral distance is achieved during the single-support phase approximately when the lateral displacement of xCoM reaches its peaks/valleys in normal walking. Therefore, the reduction in the amplitude of the lateral xCoM induced by the swinging load, as shown in Fig. 7, indicates the increase of the lateral MoS, implying an improvement in lateral stability.

Based on the analysis above, we propose the hypothesis of the effects of allowing the load to swing as:

1) Reduction in the fore-aft and medio-lateral leg impulses.
2) Reduction in the mechanical work of the stance legs scaled by the efficiency of muscle.
3) Increase of lateral margin of stability (MoS), implying an improvement in lateral stability.

D. Selection of the Pendulum Length

In the theoretical analysis above, we take the value of the pendulum length as $R = 1$ m, giving a resonant pendulum frequency of $f_{\mathrm{n}} = \frac{1}{2\pi}\sqrt{\frac{g}{R}} \approx 0.5$ Hz, which is around half of a typical stride frequency at moderate walking speeds. If we reduce the pendulum length R, the resonant pendulum frequency will get closer to the stride frequency, and the vertical relative load acceleration will have the same sign as human CoM acceleration during half of the stride period, as shown in Fig. 8. This relative load movement pattern would result in the increase of vertical leg impulses and mechanical power of the stance leg at these moments. Besides, the lateral relative load acceleration has an increased amplitude with the smaller pendulum length while remaining out of phase to the human CoM acceleration. Therefore, reducing the pendulum length would help to increase the lateral MoS, implying an improvement in lateral stability. Summarizing the analysis above, we select the pendulum length $R = 1$ m in the following experiments based on the trade-off between the

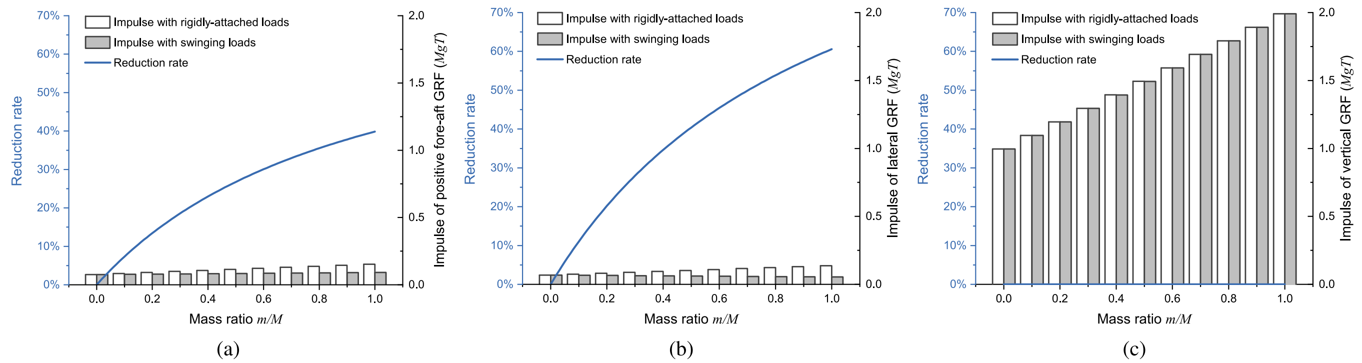


Fig. 5. Impulses of leg forces and its reduction rate versus mass ratio. (a) Fore-aft direction. The impulse of positive fore-aft GRF decreases significantly with the swinging load. The reduction rate is positively correlated to the mass ratio. The impulse of negative fore-aft GRF has the same value as that of positive fore-aft GRF. The positive and negative means forward and backward, respectively, relative to the direction of travel. (b) Medio-lateral direction. The impulse of lateral GRF decreases significantly with the swinging load. The reduction rate is positively correlated to the mass ratio. (c) Vertical direction. Allowing the load to swing has little effect on the vertical leg impulses.

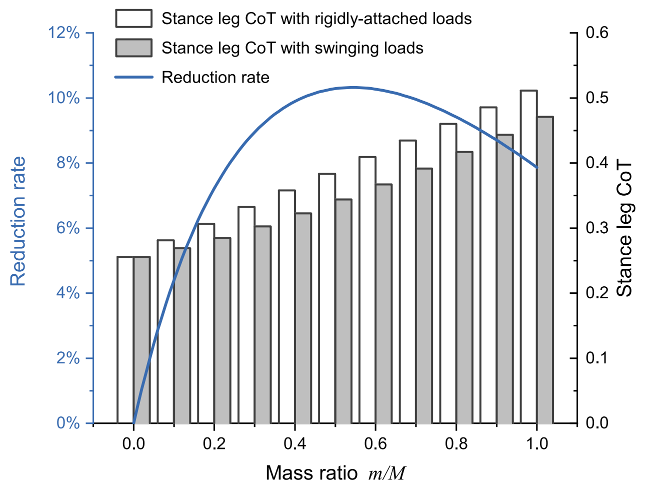


Fig. 6. Cost of transport (CoT) of the stance legs and corresponding reduction rate versus mass ratio. The walking speed in the simulation is 1.5 m/s. Allowing the load to swing results in a reduction in the mechanical energy of the stance legs.

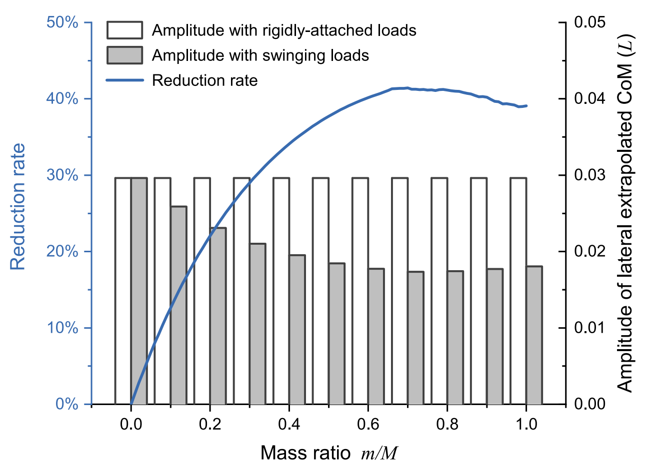


Fig. 7. Amplitude of lateral extrapolated CoM $x_y^{(ex)}$ normalized by leg length L and corresponding reduction rate versus mass ratio. Allowing the load to swing results in a reduction in the amplitude of the lateral extrapolated CoM movement.

horizontal and vertical direction effects under the moderate walking speeds.

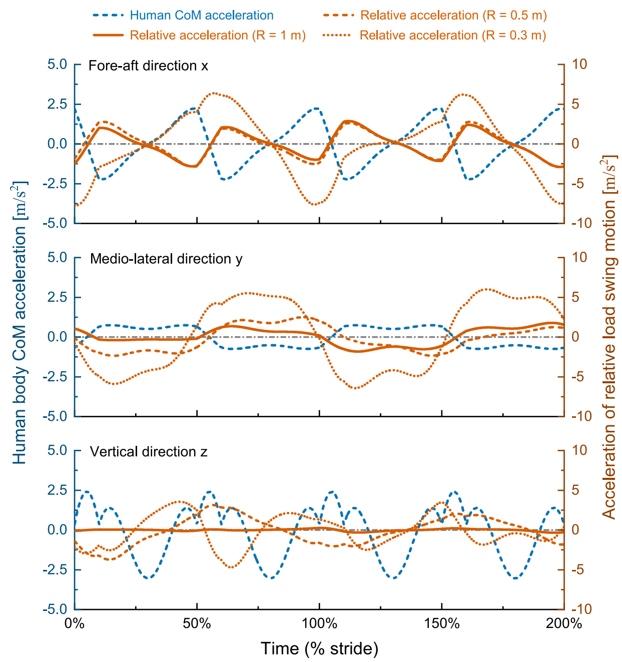


Fig. 8. Human body CoM acceleration $\ddot{\xi}$ and predicted acceleration of the relative load swing motion $\ddot{x}_r$ versus time with different pendulum length. The pendulum length $R = 1, 0.5, 0.3$ m in the figure corresponds to the resonant pendulum frequency of $f_n = 0.5, 0.7, 0.9$ Hz, while the stride frequency is 1.0 Hz in the simulation. Reducing the pendulum length R would increase the amplitude and change the pattern of the relative load acceleration $\ddot{x}_r$.

III. Experimental Validation

A. Experimental Protocol

To validate the hypothesis we proposed above, we compared the biomechanical and energetic responses of 8 human adult subjects walking with rigidly-attached loads and with the swinging loads. The subjects (male, age 42±6 years; body mass 63.0±10.4 kg; leg length 0.94±0.03 m; mean±s.d.) had the past experience with pole carrying to get accustomed to the experimental protocol more quickly. They reported no history of balance or gait disorders and signed the informed consent form in accordance with Tsinghua University policy. Subjects were equipped with the respirometry system and walked along

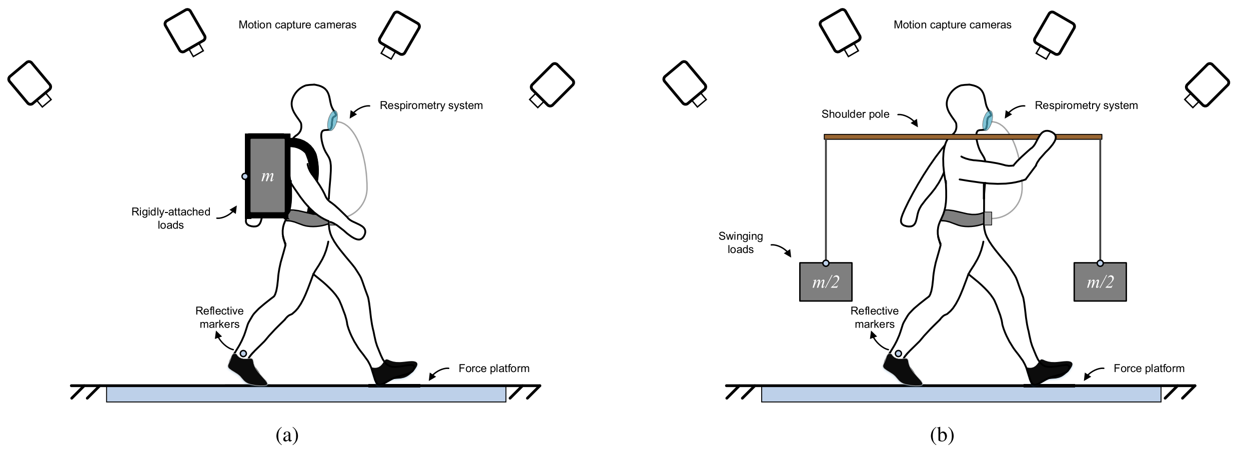


Fig. 9. Illustration of the experimental environment. (a) Walking with the typical rigid backpack. (b) Walking with the swinging load.

a 50 m long and 1 m wide circular walkway with 10 force platforms and 8 motion capture cameras instrumented in 5 m of the straight section. They walked with their self-selected natural speeds, carrying a 30 kg load representing around 47.6% of their body weight.

Two experimental conditions were conducted as shown in Fig. 9:

1) Rigid: walking with the typical rigid backpack supported by the hip. The mass of the backpack was 1.3 kg.
2) Swing: walking with the loads connected to the ends of a rigid wooden pole (length 1.3 m; width 0.08 m; stiffness 16,333 N/m) with two Bowden cables (length 1.0 m). Subjects supported the middle point of the pole with shoulder and controlled the balance of the system with the arm. The loads were free to swing with the suspension point placed at the ends of the pole. The mass of the system was 1.5 kg.

The experimental protocol contained two sessions: training and testing. Firstly, a subject conducted the 15-minute training session to get accustomed to walking with the backpack and the swinging loads. Then the subject completed the two 5-minute walking tests carrying loads with the backpack and the swinging loads, the order of which were arranged randomly. Subjects took a 10 min break between each set of trials.

B. Measurement and Data Processing

The ground reaction forces (GRFs) $\boldsymbol{F_{leg}}$ were measured using force platforms (Bertec, OH, USA) at a 1000 Hz sampling frequency and low-pass filtered with a 25 Hz cut-off frequency (fourth-order, zero-phase-shift Butterworth digital filter). A step was defined as the interval from initial contact of one foot to the initial contact of the opposite foot identified by the GRFs, whereas a stride was defined as the interval between the initial contact of the same foot.

The load movement, step lengths, and step widths were measured using an optical marker system (Motion Analysis, CA, USA) at a 120 Hz sampling frequency with markers placed on the loads and human ankles, and low-pass filtered with a 10 Hz cut-off frequency (fourth-order, zero-phase-shift Butterworth digital filter). The step (stride) lengths and step (stride) widths were defined as the fore-aft and lateral distance between the ankle markers of the heel-strike side over a step (stride), respectively.

The load forces $\boldsymbol{F_{load}}$ exerted on the carrier were calculated from the 2nd order differential of the load movement $\boldsymbol{x}$ times the load mass m as

$$\begin{aligned}\boldsymbol{F_{load}}(t) &= m\boldsymbol{a}(t) - m\boldsymbol{g}\\ &= m\frac{\boldsymbol{x}(t+dt)+\boldsymbol{x}(t-dt)-2\boldsymbol{x}(t)}{(dt)^2} - m\boldsymbol{g}\end{aligned} \tag{20}$$

where dt was the sampling period.

The movement of human CoM was determined through integration over a complete step of the vector sum of ground reaction forces and load forces [47]. Firstly, the acceleration of human CoM $\boldsymbol{a_b}$ was calculated from the summed forces exerted on the carrier divided by body mass M:

$$\boldsymbol{a_b} = \frac{\Sigma\boldsymbol{F_{leg}} + \boldsymbol{F_{load}} + M\boldsymbol{g}}{M}. \tag{21}$$

Secondly, the velocity of human CoM $\boldsymbol{v_b}$ was calculated by integrating the human CoM acceleration over a stride:

$$\boldsymbol{v_b}(t) = \boldsymbol{v_0} + \int_0^t \boldsymbol{a_b}(\tau)d\tau \tag{22}$$

where $\boldsymbol{v_0}$ was the integration constant representing the human CoM velocity at the beginning of the stride. $\boldsymbol{v_0}$ was determined by requiring the average lateral and vertical human CoM velocity to be zero and the average fore-aft human CoM velocity to be the walking speed calculated as the ratio of stride length to stride period [48].

Then the position of human CoM $\boldsymbol{x_b}$ was calculated by integrating the human CoM velocity over a stride:

$$\boldsymbol{x_b}(t) = \boldsymbol{x_0} + \int_0^t \boldsymbol{v_b}(\tau)d\tau \tag{23}$$

where $\boldsymbol{x_0}$ was the integration constant representing the human CoM position at the beginning of the stride. $\boldsymbol{x_0}$ was determined by requiring the average lateral human CoM position to be zero for symmetry. The vertical and fore-aft component

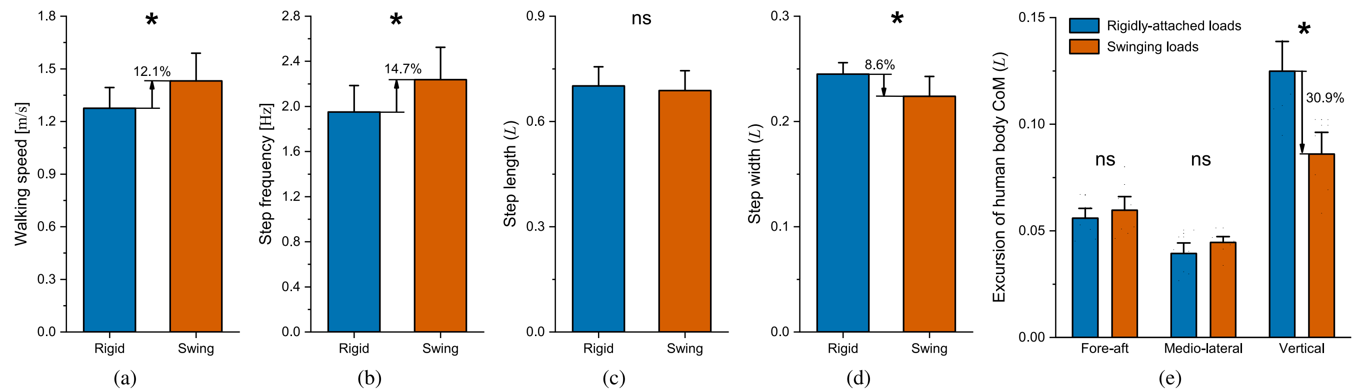


Fig. 10. Comparison of gait parameters between walking with rigidly-attached loads and swinging loads. The asterisk (*) indicates a significant difference ($p < 0.05$). 'ns' indicates no significant difference. (a) Walking speed. Allowing the load to swing increased the self-selected natural walking speed by 12.1%. (b) Step frequency. Allowing the load to swing increased the step frequency by 14.7%. (c) Step length. There was no significant difference in step length between the two loading conditions. (d) Step width. Allowing the load to swing reduced the step width by 8.6%. (e) Excursion of human body CoM. There were no significant difference in the fore-aft and medio-lateral human body CoM excursion. The vertical excursion was reduced by 30.9% with the swinging loads.

of $\boldsymbol{x_0}$ was set to be zero, as only the displacement matters, rather than the absolute position.

The mechanical energy of the stance legs per step was calculated with the mechanical work of the stance legs scaled by muscle efficiency, consistent with the definition in the theoretical analysis with (15), where $\boldsymbol{F_{leg}}$ was measured GRFs and $\dot{\boldsymbol{\xi}}$ was the velocity of human CoM $\boldsymbol{v_b}$ in (22). The cost of transport (CoT) of stance legs was then obtained by dividing the cost with body weight, walking speed, and step period as in (17).

Moreover, we measured the rates of oxygen consumption ($\dot{V}_{O_2}$) and carbon dioxide production ($\dot{V}_{CO_2}$) with a respirometry system (K4b2, Cosmed, Italy) and evaluated the whole-body metabolic cost with indirect calorimetry [49]. We took the 4th minute out of the 5-minute test for collection of average $\overline{\dot{V}}_{O_2}$ (mlO_2s^{-1}) and $\overline{\dot{V}}_{CO_2}$ $(mlCO_2s^{-1})$ data to guarantee that the subject had reached the steady state. The average metabolic power for each trial was calculated with the standard equation [50]:

$$\overline{P}_{met,gross} = 16.58\frac{Ws}{mlO_2}\overline{\dot{V}}_{O_2} + 4.51\frac{Ws}{mlCO_2}\overline{\dot{V}}_{CO_2}. \quad (24)$$

The dimensionless net whole-body metabolic cost of transport (CoT) was calculated as (average gross metabolic power - average standing metabolic power) / (body weight × walking speed):

$$\mathrm{CoT}_{met} = \frac{\overline{P}_{met,gross} - \overline{P}_{met,standing}}{Mgv} \quad (25)$$

where v is the walking speed.

C. Statistical Analysis

For each experimental condition, we calculated the mean values of each variable for 20 strides per subject for statistical analysis. We performed multiple comparisons across the data of 8 subjects using two-sided paired t-tests ($n = 8$) with Bonferroni correction. The significance level was set to be $\alpha < 0.05$ for all analysis. All statistical analysis was conducted in MATLAB (MathWorks Inc., USA).

IV. Results

A. Gait Parameters

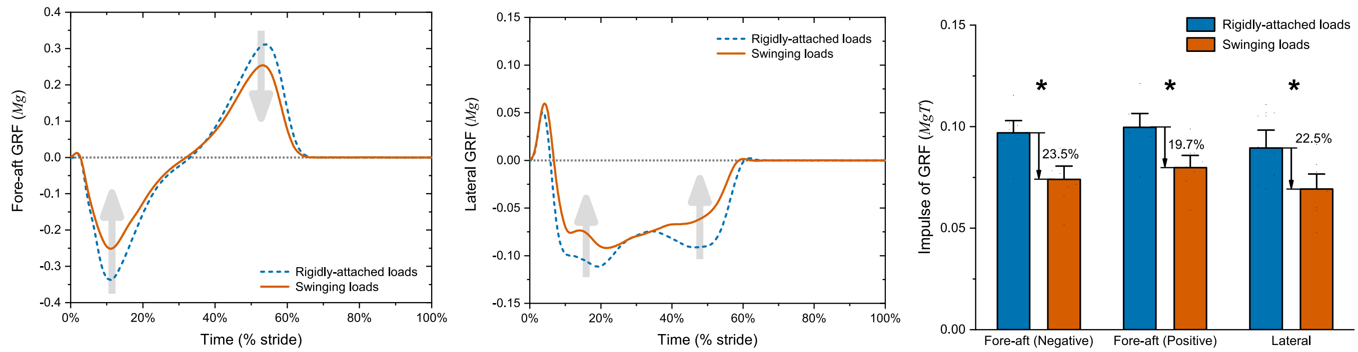
Experiments revealed that the load-carrying device would affect the self-selected gait parameters, as shown in Fig. 10. Compared to the typical rigid backpack, allowing the load to swing increased the preferred walking speed and step frequency by 12.1% and 14.7%, respectively. The preferred step width was reduced by 8.6%, while the step length was not changed. Besides, walking with the swinging loads reduced the vertical excursion of the human body CoM by 30.9%, while there was no significant difference in the fore-aft and medio-lateral human body excursion.

B. Ground Reaction Forces (GRF)

Consistent with the hypothesis, allowing the load to swing reduced the horizontal GRFs, as shown in Fig. 11. The negative and positive fore-aft leg impulses are with the directions pointing against and towards the walking direction, respectively. We calculated the impulses of GRFs with trapezoidal integral to evaluate the reduction in GRFs quantitatively. Compared with walking with rigidly-attached loads, walking with swinging loads resulted in a significant reduction in the impulses of negative and positive fore-aft and lateral GRFs by 23.5%, 19.7%, and 22.5%, respectively, indicating a significant reduction in the fore-aft and lateral leg impulses.

C. Mechanical Energy of the Stance Leg

The experimental stance leg CoT and the whole-body metabolic CoT are presented in Fig. 12. Compared to walking with rigidly-attached loads, allowing the load to swing resulted in a significant reduction in the stance leg CoT by 12.9%, consistent with our hypothesis. However, the whole-body metabolic cost calculated with indirect calorimetry showed no significant difference between the two loading conditions.



(a) (b) (c)

Fig. 11. Comparison of horizontal GRF when walking with rigidly-attached loads and swinging loads. (a) Fore-aft GRF of an individual leg over a stride versus time. (b) Lateral GRF of an individual leg over a stride versus time. (c) Comparison of the GRF impulses between two loading conditions. The negative and positive means backward and forward, respectively, relative to the direction of travel. The asterisk (*) indicates a significant difference ($p < 0.05$).

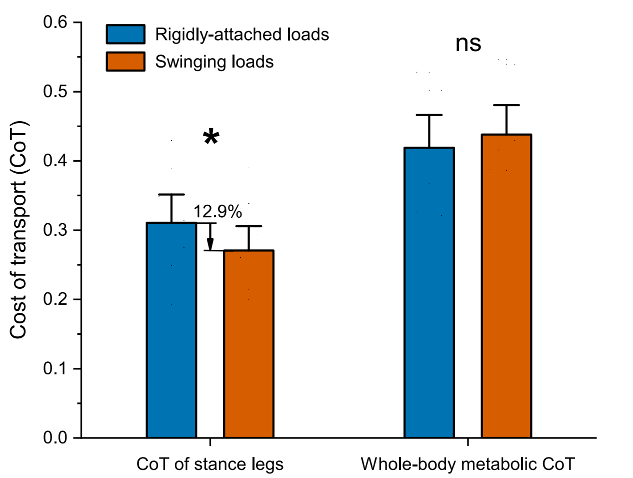
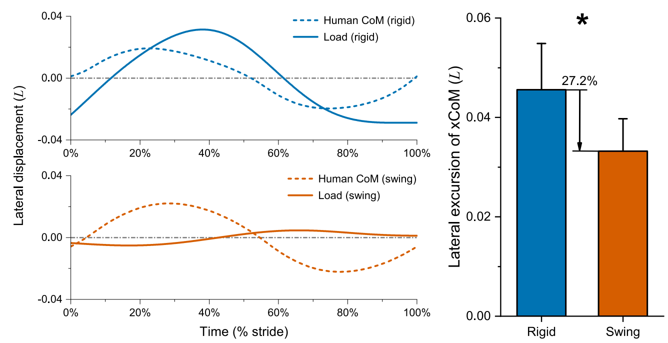


Fig. 12. Comparison of the cost of transport (CoT) between the two loading conditions. Walking with swinging loads resulted in a reduction in the CoT of stance legs compared to walking with rigidly-attached loads. The whole-body metabolic cost of transport had no significant difference. The asterisk (*) indicates a significant difference ($p < 0.05$). 'ns' indicates no significant difference.



(a) (b)

Fig. 13. Effects of the swinging loads on lateral xCoM displacement. (a) Lateral displacement of humans and loads under the two load-carrying conditions versus time. The lateral displacement of the swinging load was out of phase with that of the human body, whereas the lateral displacement of the rigidly-attached load was in phase with that of the human body. (b) Comparison of the lateral excursion of xCoM between walking with rigidly-attached loads and with swinging loads. Allowing the load to swing reduced the lateral excursion of xCoM significantly. The asterisk (*) indicates a significant difference ($p < 0.05$).

D. Lateral Margin of Stability (MoS)

Compared to the rigidly-attached loads whose CoM movement was approximately in-phase with the human CoM, the lateral displacement of the swinging load was out-of-phase to the human CoM movement, as shown in Fig. 13a, which further reduced the lateral excursion of xCoM by 27.2%, as shown in Fig. 13b. The reduction in the excursion of xCoM indicated that the lateral dynamic MoS was increased by allowing the load to swing, implying the improvement in the lateral stability.

V. Discussion

A. Changes in the Gait Parameters

In the theoretical analysis, we assume the human body CoM motion is a fixed input. However, the empirical data showed some changes in the gait parameters walking with the swinging loads, including the increase of step frequency and walking speed, and the reduction in step width and vertical excursion of human body CoM.

Previous studies have shown that the carrier prefers a higher step frequency with the use of a rigid pole in response to load [40], which may be caused by the sharp vertical interaction forces from a rigid pole as opposed to relatively cushiony backpack straps. The increase of the preferred step frequency when walking with the swinging loads may also contribute to the rigid pole, which needs further investigation by designing a comfortable interface between the carrier and the pole. Besides, the introduction of the spherical pendulum structure may also affect the optimal step frequency for energetic benefit, which should be considered with the gait adaptation in future study. As the step length keeps the same, the increase of preferred step frequency leads to increased walking speed. The increased step frequency and walking speed would reduce the fore-aft leg impulses [51], and increase the mechanical energy of stance legs [52] and the costs for swinging the legs [53], [54].

Besides, we have shown that the lateral excursion of xCoM is reduced by the swinging loads, implying an improvement in lateral stability. Previous studies have shown that individuals tend to take a narrower step when lateral stability is improved [28], [55]. Therefore the reduction of the preferred step width observed in the experiments may be attributed to the improvement of lateral stability. Moreover, the reduction of the step width may reduce the medio-lateral leg impulses [55].

It is also observed in the experiments that the vertical excursion of human body CoM decreases when walking with the swinging load. Simulation reveals that the change in the vertical body CoM excursion has little effect on the relative load movement, as shown in Fig. 14. As the vertical acceleration of the load is approximately equal to that of the carrier, the reduced vertical excursion would reduce the vertical component of the interaction force. Therefore, the reduced vertical excursion may be a strategy to avoid sharp and painful vertical interaction forces at the shoulder due to the load's relative rigidity in the vertical direction, as opposed to the cushiony backpack straps. Similar results were also observed in another study [56] showing that metabolic increased when subjects carried a compliant backpack but lowered the vertical excursion of their CoM. The results were explained in [40] to be a strategy to avoid large spikes in vertical interaction forces occurring near resonance. Given simple inverted pendulum kinematics, the fact that step length does not change indicates that vertical excursion of the body should also not change. Since individuals did reduce their vertical excursion, this implies increased leg compliance, i.e., deviation from an inverted pendulum, which would increase the metabolic cost of walking [57], [58].

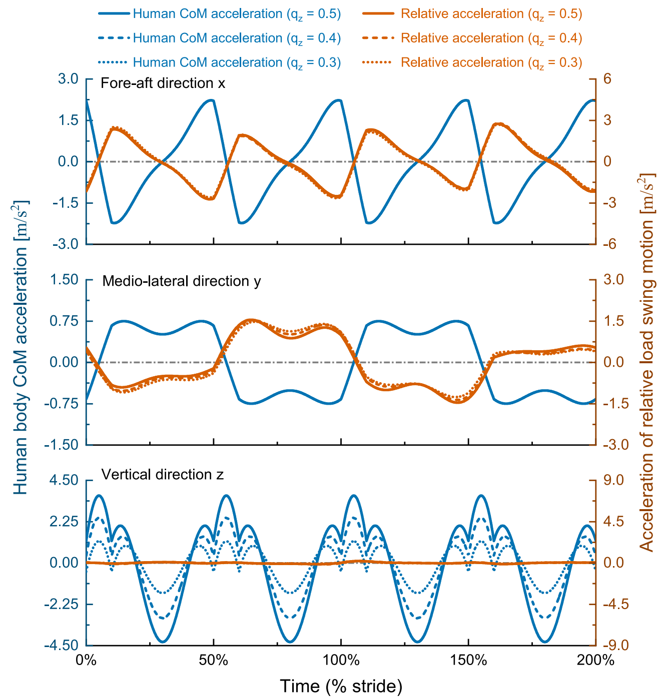


Fig. 14. Human body CoM acceleration $\ddot{\xi}$ and predicted acceleration of the relative load swing motion $\ddot{x}_r$ versus time with different vertical human CoM excursion. The shape factor q_z in (9) determines the vertical human CoM excursion, the change in which has little effect on the relative load acceleration $\ddot{x}_r$.

B. Reduced Fore-Aft and Medio-Lateral Leg Impulses

We have observed that the fore-aft and medio-lateral impulses were reduced when carrying the swinging load versus the rigid load. The theoretical analysis explains the phenomenon with the equivalent additional forces induced by the swinging load with the same direction as the summed GRFs. Considering that the increase in the preferred step frequency would also reduce the fore-aft leg impulses, we present the empirical results of the fore-aft leg impulses normalized by MgT in Fig. 11. The gravitational impulse model [51] has predicted the push-off impulse as $P^* = Mv^- \tan\theta + Mg\Delta t/(2\cos\theta)$, with v^- indicating the pre-collision velocity of human body CoM, θ indicating the half step angle, and Δt indicating the duration of double-support phase. The normalized push-off impulse can then be expressed as $P_n^* = v^- f \tan\theta/g + r/(2\cos\theta)$, where f is the step frequency and r is the proportion of the double-support phase in a cycle. Assuming the proportion r is not affected by the step frequency, the normalized push-off impulse should increase with the step frequency, as the pre-collision velocity v^- should increase with higher step frequency, and the half step angle θ should remain the same for the unchanged step length observed in experiments. As the fore-aft leg impulse is positively correlated to the push-off impulse, the analysis above predicts it to increase with higher step frequency. Therefore, the observed reduction in the normalized fore-aft leg impulses should be attributed to the change in the interaction forces induced by the swinging load rather than the increased step frequency. To figure out the effects of the increased step frequency and the change in the GRFs, future studies collecting the data of walking with constant step length and different step frequency would help to make it clearer and more convincing.

Besides, with the medio-lateral leg impulses, in addition to the interaction force caused by the swinging load as demonstrated in the theoretical analysis, the reduction in the preferred step width induced by the swinging load could also lead to the reduction in medio-lateral impulses.

C. Mechanical Energy of the Stance Leg and Whole-Body Metabolic Cost

One of the significant results in the theoretical analysis and experiments is that the mechanical energy of the stance legs is reduced by allowing the load to swing. Although the increase in the walking speed might be expected to increase the CoT of the stance legs [2], [51], the changes in the interaction forces induced by the swinging loads offset the effect and further reduce the CoT of stance legs.

Besides, it is worth noting that although allowing the load to swing reduces the mechanical energy of the stance leg, there is no significant difference in the whole-body metabolic cost. There are several possible interpretations. The increase of step frequency may induce a higher cost of swinging the legs [53], [54]. The reduced vertical excursion of body CoM may increase the metabolic cost [57], [58]. The additional effort made by arms to control the pose of the pole and prevent it from slipping off the shoulder may increase the energetic cost of the upper body. Holding the arms while walking may also result in a higher metabolic cost [33]. We have discussed that the increased step frequency and the reduced vertical body CoM excursion may be caused by the sharp vertical interaction forces of the rigid pole at the shoulder. Therefore, designing the interactive interface between the pole and the carrier for higher comfort may help to reduce the metabolic cost. The additional effort of arms may be solved with a more sophisticated design of the load carriage assistance device in the future, such as fixating the pole to a frame rigidly attached to the carrier and supported by the hip to free the hands and reduce the required effort of the upper body.

Moreover, some earlier research measuring the whole-body metabolic cost of walking with compliant poles has similar results to our study of the swing motion. The loads are placed at the end of the poles in these experiments. The compliant pole results in a 5% reduction in the whole-body metabolic cost compared to carrying loads with the rigid pole [38]. However, there is no significant difference in the whole-body metabolic cost between carrying loads with compliant poles and with typical backpacks [36], [38], which may be induced by an extra energetic cost of controlling the balance of the pole [39]. Our work compared the difference in the mechanical energy of the stance leg derived from the mechanical work besides the whole-body metabolic cost induced by the muscle work of the entire body. The results implied that the energetic cost distribution between lower extremities and the upper body should be noted in the future study of the load carriage mechanism.

Thereby we can conclude that the vertical elastic suspension and the load swing motion both can reduce the mechanical energy of the stance leg, whereas the introduction of the pole requires additional cost of leg swing or the upper body. These three factors influence the whole-body metabolic cost together, and the effects of elastic suspension or swing motion can approximately cancel out the adverse side effects introduced by the pole. If the combination of elastic suspension and swing motion could gain synergy, the system would have better performance than the elastically suspended backpack. Therefore the combination form and the structural parameters should be carefully designed and need further investigation in the future.

D. Improvement in Lateral Stability

We have discussed the improvement in lateral stability induced by the swinging load from the view of the lateral margin of stability (MoS). The reduction in the lateral excursion of xCoM induced by the lateral load oscillation out of phase to the human CoM indicates the increase of MoS, and further implies the improvement in lateral stability. The backpack with load compliance in the lateral direction found similar results, including the load oscillation out of phase to the subjects, the reduction in horizontal impulses, and the reduction in the preferred step width [30].

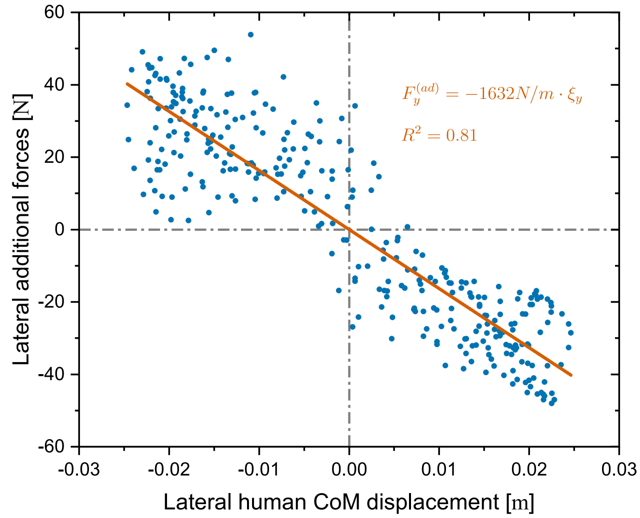


Fig. 15. Empirical data of the additional lateral forces versus lateral human CoM displacement when walking with the swinging loads (marked as circles). The data points are sampled from different moments of a stride cycle. Linear regression through origin fits (solid lines) are performed ($R^2 = 0.81$). The additional lateral forces induced by the swinging load are equivalent to a spring with stiffness $k = 1631$ N/m.

Moreover, the effect of the swinging load in the medio-lateral direction could also be viewed as an equivalent external lateral stabilizer [28] pulling bilaterally from the waist with springs. The additional lateral forces induced by the swinging loads are opposite to the lateral displacement of human CoM, and acts equivalent to a spring of 1631 N/m when the load mass is 30 kg, taking the effect of the external stabilizer, as shown in Fig. 15. The additional forces in the figure are calculated as $F^{(ad)} = -m(\ddot{x} - \ddot{\xi})$, where $\ddot{x}$ and $\ddot{\xi}$ are the acceleration of load and human CoM in the experiments of walking with the swinging load. The study about the external lateral stabilizer [28] has shown its improvement of the lateral stability and found similar results of the reduction in preferred step width.

E. Advantages Over Elastically Suspended Backpacks

Allowing the load to swing has similar effects to the elastically suspended backpacks in the reduction of the mechanical energy of the stance leg and improvement of stability [59]. We will discuss some advantages of swinging loads over the suspended backpacks.

Dynamics of walking with existing elastically suspended backpacks could be characterized by the forced vibration of the spring-mass-damper system [60]. The load movement and corresponding energetic effects of the elastically suspended backpack are significantly affected by the relationship between the walking frequency $\omega = 2\pi f$ and the natural frequency of the system $\omega_n = \sqrt{k/m}$ [24], where k is the stiffness and m is the load mass. When $\omega > \omega_n$, the load movement is out-of-phase to that of human CoM, resulting in a reduction in the energetic cost. However, when $\omega < \omega_n$, the oscillating loads could even increase the energetic cost compared to the typical rigid backpack [56]. It means that carriers should carry heavy loads or walk faster than the preferred walking speed to make the suspended backpack function effectively, or else it may even have adverse side effects. The application of the elastically suspended backpack is thus limited. In contrast, we have shown that the swinging load movement actuated by the human CoM movement is independent of the load mass. Therefore, although the swing motion is also affected by the step frequency as discussed in Sec. II-D, allowing the load to swing with the pendulum length of 1 m would always gain biomechanical benefits regardless of the load mass under the moderate or higher walking speeds, implying a higher generality than the elastically suspended backpack.

Moreover, most of the elastically suspended backpacks adopt the linear guide rail and slider structure to restrict the load movement to keep in the vertical direction [8], [12], which means an additional mass in the structure of the device. On the contrary, the spherical pendulum structure only adopts cables and a rigid pole which is as light as the typical backpack. Therefore the device allowing the load to swing has an additional advantage of lightweight over the elastically suspended backpacks.

F. Limitations and Recommendation

Although allowing the load to swing performs well in reducing the horizontal leg impulses and the mechanical energy of the stance leg, and improving the lateral stability, there are some limitations and could be improved in the future study and design.

Firstly, we assume the trajectory of human body CoM to be fixed through the different load carrying conditions. However, the control of gait is often responsive to dynamic interactions with the environment. The empirical results also reveal that subjects tend to take larger walking speed and step frequency when carrying the swinging load. Future investigation may consider human adaptation and predict the preferred walking speed with an optimization model of human walking. It is also worth conducting experiments controlling the walking speeds and compare the walking performance of two load-carrying conditions.

Besides, as we have discussed, although the mechanical energy of the stance leg decreases, there is no significant difference in the whole-body metabolic cost, which may be attributed to the increase in the costs of swinging the legs or the effort of the upper body to control the balance of the pole. However, with the simplification in the theoretical model omitting the upper body and the leg swing, the effects of these two factors remain undetermined. Moreover, no measurements were taken of the upper body and leg swing motion regarding work and relevant muscle activity in the experiments. The cost of the upper body and the leg swing should be considered in the future theoretical and empirical studies to figure out their roles. After learning about the reason for failing to reduce the whole-body metabolic cost, the issue may be solved by the novel design of the tool freeing the hands and positioning the loads symmetrically, or by combining the swing motion with the elasticity to further reduce the mechanical energy of the stance leg canceling out the extra cost introduced by the pole.

Moreover, the horizontal movement of loads requires a larger room to avoid interference to human limbs. Therefore we use a pole and place the suspension points of the spherical pendulum at the ends of the pole to increase the horizontal distance between the human and loads. The greater space occupied by the device restricts the application scenario to be the outdoor open space and limits its generality.

In future work, the swing motion may be combined with the vertical suspended system and obtain better performance in load carriage assistance. The selection of the parameters, including the stiffness, damping, and pendulum length, needs further investigation and generates the optimal design for the device similar to the compliant shoulder pole widely used in Asia.

VI. Conclusion

In this paper, we theoretically analyze and experimentally validate the biomechanical and energetic benefits of allowing the load to swing rather than the typical rigid backpack. The benefits include the reduction in horizontal GRFs and the mechanical energy of the stance leg, and the improvement of lateral stability. Allowing the load to swing results in an additional force with the same sign as the horizontal GRFs, reducing the backward, forward, lateral impulse of GRFs by 23.5%, 19.7%, and 22.5%, respectively, compared to carrying a 30 kg load with typical rigid backpacks. The reduction in horizontal GRFs further leads to a reduction of 12.9% in the mechanical energy of the stance leg. Besides, the load movement out of phase to the human body CoM in the lateral direction reduces the lateral excursion of xCoM by 27.2%, indicating an increase in the MoS and implying an improvement of lateral stability. Although the whole-body metabolic cost has no significant difference based on the current design of the tool, it may be improved by a novel design such as combining the elastic suspension, which needs further investigation in the future.

Appendix
Symmetry Between the Two Loads and the Bilateral Sides

The walking model in our study includes a single load attached to the body's CoM, which is a simplified version of the real-world scenario where subjects were asked to carry two loads symmetrically using a fixed pole. This appendix presents the symmetry between the two loads and the bilateral sides from the empirical data.

Comparing the movement of the two loads measured with experiments, the front and the rear loads were always swinging in phase with each other, with similar amplitude in the three directions, as shown in Fig. 16. Therefore the effect of the two loads could be equivalent to the simplified single load in the model.

Besides, the load was supposed to be positioned along the middle line of the human body in the walking model, whereas the rigid pole carrying the loads was placed on the right shoulder of the subjects in the experiments, which may induce some asymmetry. We compare the leg impulses for the bilateral sides

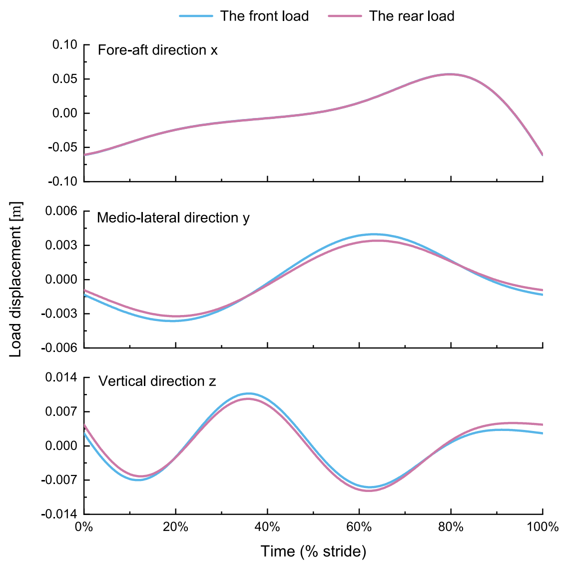


Fig. 16. Displacement of the front and the rear load. The two loads were always swinging in phase with each other, with similar amplitudes in the three directions.

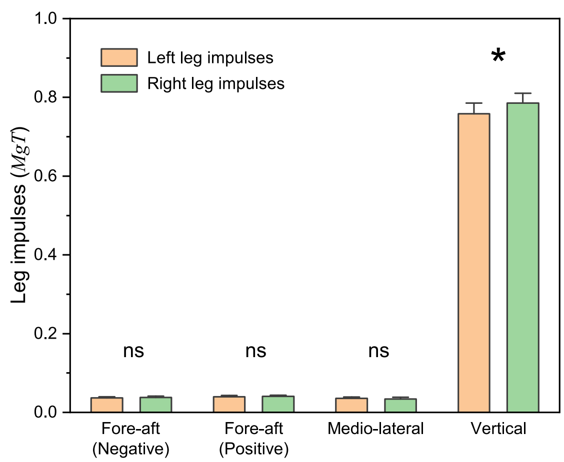


Fig. 17. Comparison between the leg impulses of the left and the right side. The vertical leg impulse per stride of the right side was larger than the left side. There was no significant difference in the fore-aft and medio-lateral leg impulses between the two sides. The asterisk (*) indicates a significant difference ($p < 0.05$). 'ns' indicates no significant difference.

in Fig. 17. There is no significant difference in the fore-aft and medio-lateral leg impulses. However, the vertical leg impulses of the right side are larger than that of the left side, which may be caused by a lateral torque on the body due to the pole positioned on the right shoulder. Future analysis is needed to study the possible effects of the way the pole is positioned on the body.

Acknowledgment

The authors would like to thank anonymous reviewers for their constructive comments and suggestions on the paper.

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