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Human Walking Centre of Mass Trajectory and Dual-Degree-of-Freedom Power Suspension Backpack System

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ABSTRACT

The human body generates a sophisticated three-dimensional complex motion of the centre of mass when walking, with the majority of this vibration occurring in the vertical and coronal axes. This paper presents an analysis of the human walking motion, accompanied by the establishment of an equivalent model of the human walking COM motion. This law is integrated into the coronal plane to generate the Lissajous curve expression of the human walking COM motion trajectory. Finally, the reliability of the results was verified by comparing the COM motion equivalent model with several sets of walking experimental data. The model is capable of accurately predicting and analysing the walking amplitude and frequency of the COM at varying heights and speeds. Furthermore, a dual-degree-of-freedom power suspension backpack system has been designed based on the model, which has the potential to significantly reduce the low-frequency vibration and inertial force impacts of heavy loads on the human body's vertical and coronal axes during load-bearing walking. This could help to alleviate the resulting mechanical injuries to the human body and reduce the additional metabolic energy consumption required to resist these impacts. The device is capable of effectively reducing the low-frequency vibration and inertial force impacts in the vertical and coronal axes of the human body when walking with a load.

KEYWORDS

Centre of mass, Walking exercise, Suspension system

1. INTRODUCTION

Human gait is a complex motion, but most studies only need to determine the laws of motion of the centre of mass (CoM) during human gait. A number of studies have previously been conducted on the modelling of COM motion, including those by Dingwell and Cusumano [1, 2], Hof [3, 4], and Srinivasan and Ruina [5], among others. However, the majority of these studies have concentrated on the direction of the vertical axis of the human body, with comparatively fewer investigations into the COM motion of the coronal axis of the human body. Examples of such studies include those by Hof [3, 4], Kuo [6, 7], and McGeer [8], and there is a paucity of articles that address the correlation between the two. Based on the law of motion of the COM of the human body, this paper calculates and fits the law of motion of the COM of the human body in two directions: vertical axis and coronary axis, where the vibration of the COM of the human body is largest. Through the equivalent model of human body walking, the Lissajous curve expression of the COM motion on the coronary plane is established.

The objective of this study is to investigate the impact of inertial forces generated by the human body when carrying weight and additional bioenergy consumption, as observed in previous studies [9-11], and to develop a carrying system that can reduce these forces. The proposed system employs the principle of elastic suspension, as seen in other forms of elastic support, such as the elastic riding posture [12] and shoulder load [13]. It is important to note that similar studies have been conducted by other researchers in the past. For example, He and Zhang [14] designed an active-passive powered backpack with the objective of reducing vertical inertial force injuries and metabolic energy consumption. Furthermore, there is a correlation between horizontal stability and the energy expended during walking. It has been demonstrated in experimental studies that the use of elastic elements to provide lateral stability-assisted restraints can reduce the metabolic cost of walking and stabilise gait [15-17]. Furthermore, Martin and Li et al. [18, 19] designed an oscillating energy-trapping backpack for vibrations in the direction of the body's coronal axis, with the aim of reducing peak load horizontal vibration stresses and capturing electrical energy.

The human walking equivalent model studied in this paper allows for the clear demonstration of the motion relationship between the human COM and the load on the back. Based on this model, a dual-degree-of-freedom power suspension backpack system has been designed for this specific motion mode. The system simultaneously mitigates the vibration of the load on the human body in both vertical and horizontal directions during walking and reduces the inertial force impact of the load and the unnecessary energy consumption of the human body.

2. HUMAN WALKING MOTION ANALYSIS AND EQUIVALENT MODELLING

2.1. Analysis of Human Walking Motion Characteristics

During walking, the human body's COM generates a substantial amount of movement information in three-dimensional space simultaneously, which would be challenging to analyse and quantify in the absence of a unified method of description. In essence, organismal motion can be regarded as the translation of the COM of each part in space along the path of least energy expenditure. This concept provides a qualitative analysis and a unifying principle for the fundamental determinants of gait. In order to facilitate analysis, this paper will assume that the body weight of the human body during the walking cycle is concentrated at a single point, namely the body's COM. Furthermore, the limbs will be regarded as weightless with respect to the body, thereby avoiding a discussion of the forces acting on the individual joints and body parts [20].

This paper presents an analysis of several publicly available datasets on human walking, collected by the research teams of Fukuchi [21], Camargo [22] and Van [23]. The objective is to investigate and validate the model's universal laws. The analysis of these datasets allows the retrieval of multiple sets of human COM 3D motion trajectories comparable to the one depicted in Figure 1 (a). Following the spreading and filtering of the aforementioned trajectories, the motion trajectories in the z-direction of the vertical axis and the y-direction of the coronal axis can be obtained, as illustrated in Figure 1 (b). Given that the trajectory of the COM motion when the human body walks approximate a sinusoidal curve in both directions [20], this paper employs a sinusoidal curve to fit the motion in these two directions and analyse the change of the motion in the vertical z-direction and the horizontal y-direction in terms of the period T against it, as illustrated in Figure 1 (c). The results of the analysis demonstrate that at the instant of simultaneous foot contact, the COM of the human body is situated at the vertical nadir, with the horizontal position approaching the midpoint. Conversely, at the highest point of the COM in the vertical direction, the horizontal displacement reaches its limit. Concurrently, the period of vibration in the vertical direction is equivalent to the half-step period of walking, which is defined as half the whole gait cycle.

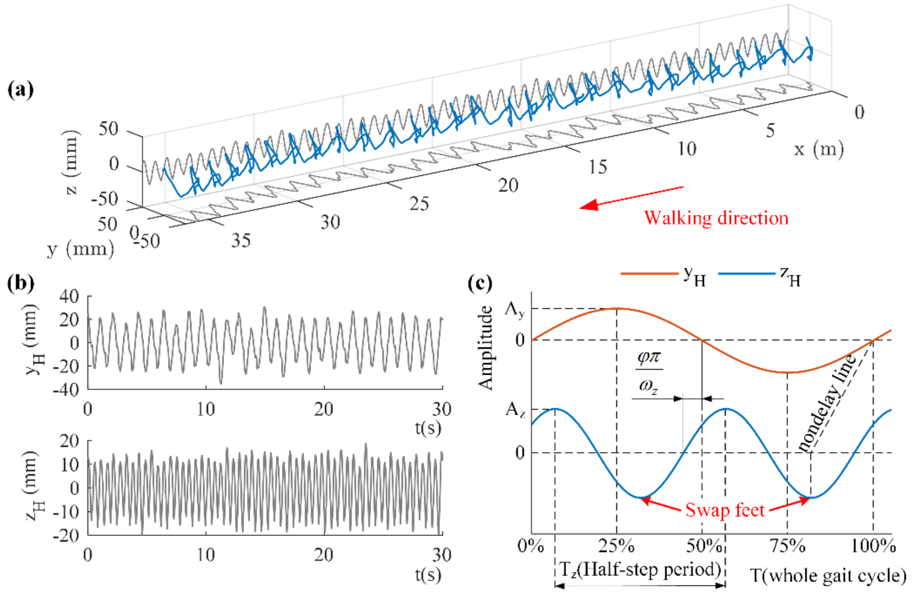

Figure 1. The motion law of human walking COM. (a) Three-dimensional trajectory of the human COM. (b) Filtered planar expansion of the human COM trajectory in the y-direction of the coronal axis and the z-direction of the vertical axis. The figure also indicates the phase delay width $\varphi\pi/\omega_z$ and the moment of feet swap. The inclination of the nondelay line in the graph is the primary factor contributing to the formation of the W-shaped walking COM motion in the coronal plane projection. Conversely, when the line is vertical, the COM motion assumes a V-shaped configuration in the coronal plane projection.

2.2. Equivalent Model of Human Walking

In order to quantify and statistically analyse the state of CoM motion of different individuals in different situations over time, an equivalent model of human walking can be established to facilitate a deeper understanding of the relationship between the CoM motion law and individual differences and different working conditions. The primary coefficients can be identified through sinusoidal fitting, whereby the amplitude A of vibration and the frequency ω of walking are determined. In the equivalent model, the corresponding variables are the height of the individual h_H and the speed of walking v.

The relationship between height and leg length is $l_0 = 0.53h_H$, the relationship between pendulum angle α and step size is $l_p = 2l_0 \sin\alpha$, the amplitude of COM is $A_z = \frac{1}{2}l_0(1-\cos\alpha)$, walking speed $v = f_z l_p$, The step frequency f_z can be converted to $\omega_z = 2\pi f_z$. The vertical COM vibration frequency ω_z and amplitude A_z can be obtained by deriving and correcting the equations for the knee and ankle joints, as well as incorporating the effect of pelvic rotation on the human body's COM motion during walking, as demonstrated in equations (1) and (2) [24].

$$\omega_z = 4\pi \times 64.8(\frac{v}{h_H})^{0.57} / 60 \tag{1}$$

$$A_z = 0.265h_H\left(1-\sqrt{1-\left(\frac{0.963v}{2.2896h_H\left(v/h_H\right)^{0.57}}\right)^2}\right)-0.008321h_H \tag{2}$$

The unit of step frequency ω_z in the formula is rad/s, amplitude A_z and height h_H units are m, walking speed v in units of m/s.

Similarly, the horizontal direction is modelled. From Figure 1 (c), the vibration frequency of the COM in the z-direction is related to the vibration frequency in the y-direction by a factor of two, i.e., $\omega_z = 2\omega_y$. In a gait cycle, during the process of a pedestrian touching the ground from the heel of one foot to the tip of that foot leaving the ground, the COM of the human body will move from the inside out and then inward in the y-direction [20], and this displacement will decrease with the increase of walking speed [25]. Further research has found that the lateral displacement of the COM is in a U-shaped relationship with walking speed [26, 27].

Since the horizontal COM motion is affected by various aspects of human kinematics and coordination between muscle mobilisation and various joints, it is difficult to calculate the relationship between its specific motion parameters and variables well with a purely mechanical model, so the data fitting method was chosen to predict the horizontal motion amplitude. The horizontal amplitude can be obtained by fitting, such as $A_y = \frac{1}{2}\left(-68.1Fr+165.3Fr^2+10.84\right)/100$, the unit is m [26], wherein, the dimensionless Froude number of walking force is $Fr = v^2/gl_0$ [28-30]. Finally, the relationship between the magnitude of the horizontal motion amplitude of the COM A_y and the walking speed v and height h_H can be obtained, as shown in Equation (3).

$$A_y = 0.8265\frac{v^4}{0.53^2h_H^2g^2}-0.3405\frac{v^2}{0.53h_Hg}+0.0542 \tag{3}$$

The height and walking speed are calculated by the equivalent model of human walking and compared with the parameters obtained by fitting the actual measured human body centroid data. The raster point cloud map shows the comparison of the two on the independent variables, and then the consistency evaluation is carried out using the Bland-Altman plot. After eliminating the invalid data, as shown in Figure 2. It can be seen that most of the points fall within the 95% confidence interval, which indicates that the amplitude consistency between the model calculation and the actual human body's COM is good, and the equivalent model is reliable.

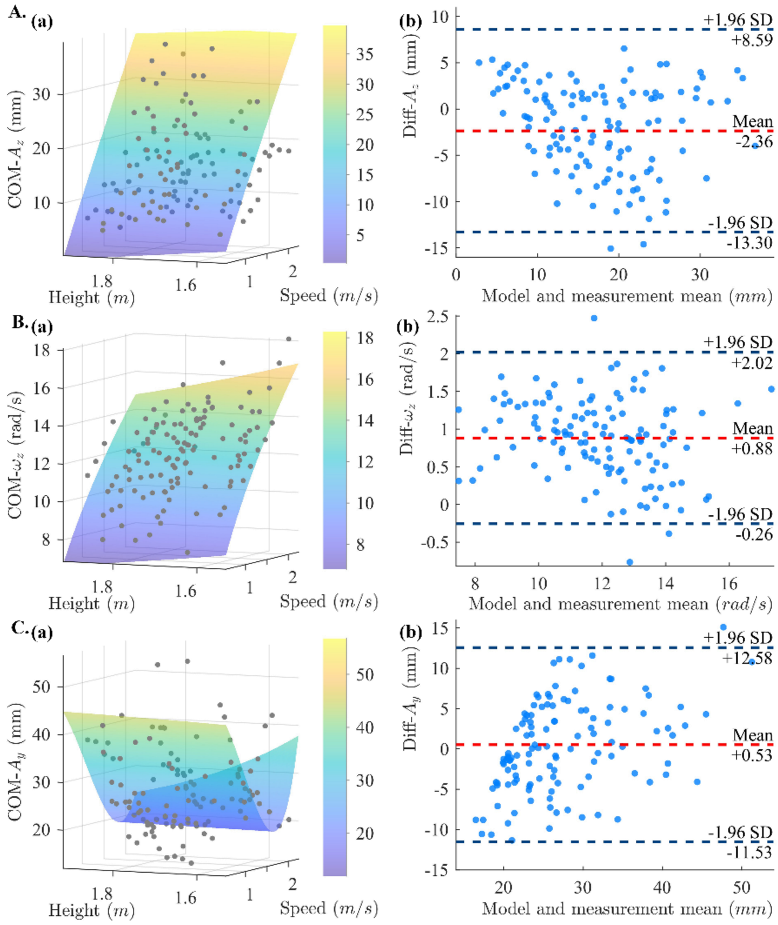

Figure 2. Data analysis of human walking equivalent model. A and B represent the relationship between amplitude and frequency in the vertical direction of the COM and the height and walking speed of the walker, respectively, where (a) represents the variation and relationship between the measured data and the model-predicted data in the independent variables of height and speed (the points in the graphs are the actual measurements and the planes are the model-predicted values), and (b) represents the Bland-Altman analytical plots of the actual measurements and the model-predicted values. Since ω_z and $2\omega_y$ are strictly equal in the law of motion, the verification of ω_y is omitted.

Due to the transfer of the COM and the pelvic rotation during walking, there is an additional phase difference between the horizontal and vertical directions of the COM motion function, which makes the projection of the human COM trajectory on the coronal plane in the walking state appear as a W-shape instead of a V-shape, while the phase lag φ has a significant dependence on the walking speed [25], which can be derived from the fitted equation (4).

$$\varphi = -0.3517v + 0.747 \tag{4}$$

Where the unit of phase φ is $\pi\ rad/s$. When $\varphi > 0$, the y-direction function is phase-delayed at z; when $\varphi < 0$, the z-direction function is phase-delayed at y. The value of φ at the time of walking should be positive.

With all parameters of the equivalent model of human walking, the motion components of the COM in horizontal and vertical directions can be obtained by substituting the height and walking speed into the two parameters, and then the motion trajectory of the COM of walking with time as a variable can be obtained by substituting it into the sine curve. After the sine trajectory of the two vertical directions is combined, the Lissajous curve [25] on the centroid motion coronal plane can be obtained, which is driven by Equation (5).

$$\begin{cases} y_H = \pm A_y \sin(\frac{1}{2}\omega_z t) \\ z_H = A_z \sin(\omega_z t + \varphi\pi) \end{cases} \tag{5}$$

Where, y_H and z_H are the displacement components of human body's COM in two directions of the coronary axis and vertical axis, in mm. The equation expresses only the stabilisation phase of the gait cycle and does not include the transition phase from standing to walking. The equation is expressed from the beginning of the first stable walking cycle, i.e., at the moment of the first foot change. y is positive or negative depending on whether the starting leg is the left or the right leg; if the starting leg is the left leg (i.e., the left foot is lifted first at the start), the left leg is the supporting leg after the first foot change. If the starting leg is the left leg (i.e., the left foot is lifted first for the start), the left leg is the supporting leg after the first foot change, and the right leg is the swing back, and at this point the COM tends to be far away from the sagittal plane of the body to reach the furthest end of the left-side displacement, and y takes a positive number, on the contrary, y shall be negative.

3. DUAL-DEGREE-OF-FREEDOM POWER SUSPENSION BACKPACK SYSTEM DESIGN

3.1. Mechanical Principle of the Backpack System

The fundamental objective of the backpack system design is to decouple the motion relationship between the load and the human body. This is achieved by reducing the vibration of the load in both vertical and horizontal directions due to the human body's motion. Consequently, the wearer's expenditure of extra metabolic energy is also reduced. From the standpoint of comprehensive system analysis, a dual-degree-of-freedom zero-acceleration levitation backpack system is devised by transferring and postponing the phase of the backpack system to attain zero system stiffness and damping in the two degrees of freedom.

As illustrated in Figure 3, the configuration of the dual DOF dynamic suspension backpack system encompasses the following principal components: the primary structure is partitioned into two sections, namely the back plate, which is affixed to the posterior aspect of the human body, and the storage plate, which is secured to the load. The upper portion of the system is constrained by double guide rails, which serve to restrict the reciprocating motion of the load in the vertical direction and to bear the overturning moment resulting from the outward movement of the load COM. The direction is determined by the right-hand rule, which points in the negative direction of the coronary axis. The lower part of the system is connected to the guide rail fixed point through the large lead high-speed lead screw, forming an inverted triangle structure. This structure actively controls the vertical lifting movement of the load through the servo motor. The load board, which bears the weight, is connected with the lead screw slider through the rotating shaft. An arc notch is set at the connection with the guide rail slider, and a grooved bearing is assembled. This configuration allows the weight to swing freely to the left and right in the horizontal direction. The storage plate and the back plate are connected with an elastic belt through multiple pulleys, thereby providing a vertical supporting force for the load. This allows the weight to be balanced in the middle of the guide rail, thus reducing the motor load. Additionally, horizontal springs are connected between both sides of the parcel shelf and the guide rail slider. These springs provide a centring restoring force when the load swings left and right.

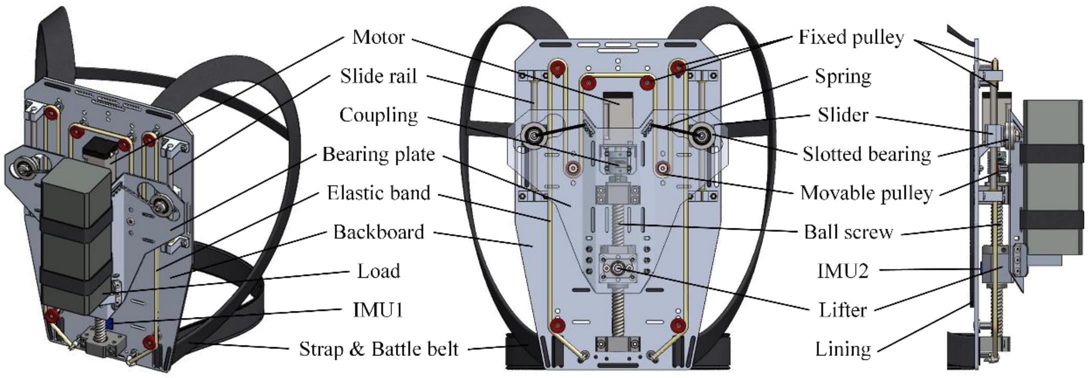

Figure 3. Schematic diagram of the Dual-degree-of-freedom power suspension backpack system

The human-machine interface design of the backpack incorporates a waist belt with a waist seal, which concentrates the mass of the backpack more on the waist than on the shoulders. While this does not significantly reduce the metabolic exertion of the wearer [31], the multiple points of support can effectively reduce the pressure on the shoulders and enhance the wearer's wearing comfort. Furthermore, this configuration minimises the impact on body manoeuvrability when concentrating the load mass on the waist [32]. The physical demonstration of this system is presented in Figure 4.

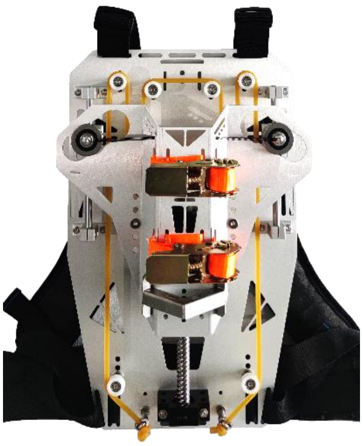

Figure 4. Picture of the dual-degree-of-freedom power suspension backpack system

3.2. Design Parameters of the Backpack System

The mean walking speed of the human body is approximately 5 to 7 kilometres per hour. In order to guarantee the applicability of the design parameters, the maximum value of 7 km/h has been selected as the basis for model selection in this study. By fitting the walking data, the centroid displacement trajectory function can be obtained. By differentiating this function, it is possible to calculate the instantaneous velocity and acceleration during the act of walking. Furthermore, the maximum tensile force required for the load at the extreme position can be determined, which provides a crucial basis for the design parameters of the backpack, as illustrated in Table 1.

In the vertical direction, the velocity is at a maximum when the acceleration is zero, and the acceleration is at a maximum when the velocity is zero. This is due to the inherent properties of the sinusoidal function and the design of the backpack. Concurrently, the coefficient of friction is at its lowest value at the maximum velocity, and at the minimum velocity, it is at its highest value. In the context of prolonged weight-bearing exercise, it is recommended that the overall weight of the backpack should be maintained at a level that represents no more than 30% of the wearer's body weight [33]. Accordingly, following an assessment of the backpack's own mass, a standard load capacity of 15 kg was established.

In accordance with the findings of Martin and Li, the mean horizontal displacement of the human body during walking does not exceed 2 cm [18]. Accordingly, the horizontal swing angle has been designed to be ±5°, and the height from the COM of the storage plate to the rotating axis is more than 12 cm. This ensures that the horizontally permissible travelling distance of the load's COM meets the horizontal displacement requirements of the human body. Additionally, at the limit position, the centring restoring force is sufficient to return the swing angle to the central position.

Table 1. Design parameters of the system

vertical/horizontal stroke	maximum vertical speed	maximum vertical acceleration	vertical/horizontal maximum traction force
120/20 mm	600 mm/s	6 m/s^2	37/1.51 N

3.3. Mechanical Design of the Backpack System

The selection of elastic bands and springs present in the design of the backpack exerts a significant influence on the backpack's performance, as well as the varying results of their mounting positions. Therefore, it is essential to adjust the modulus of elasticity of the elastic bands and the stiffness selection of the springs, as well as the suspension positions of both, in order to achieve the desired outcome.

The conversion of the elastic band model to a system stiffness facilitates comprehensive system analysis. The vertical stiffness of the system k_{sysZ} can be derived from Eq. (6) in N/m.

$$k_{\text{sysZ}} = 4^2 E \frac{S}{L_{e0}} \tag{6}$$

In the formula, S represents the cross-sectional area of the elastic band, measured in m^2, while L_{e0} denotes the effective length of the elastic band when it is not subjected to external forces, measured in m. As can be observed from the formula, an increase in the length of L_{e0} (the effective length of the elastic band installed into the system when it is not stretched by force) will result in a reduction in system stiffness. This is beneficial for the performance and control of the backpack, as system stiffness can impede these functions. Therefore, in the design of the backpack, the elastic band will be connected by multiple sets of pulleys with the objective of increasing the initial length, thereby reducing system stiffness.

In order to prevent the system from resonating, the excitation frequency ω_E, i.e., the vibration frequency of the COM of the human body, needs to be greater than two times the ω_N, i.e., $\omega_E > 2\omega_N$, of the intrinsic frequency of the backpack in this direction, so that the system is able to avoid resonance. The intrinsic frequency ω_{Nz} of the system in the vertical direction can be derived from equation (7), the excitation frequency in the vertical direction is ω_z in Eq. (5).

$$\omega_{\text{Nz}} = \sqrt{k_{sysZ} / m_l} \tag{7}$$

Where m_l is the mass of the load, which also contains components such as the shelf plate that remain rigidly connected to the load.

Similarly, in the horizontal motion, the system presents an inverted pendulum model with spring centered restoring force. Because the horizontal component of elastic belt tension is coupled with the vertical motion, the stress is relatively complicated. In order to verify the resonance characteristics, a dynamic equation shall be established, as shown in formula (8). Through motion simulation and fast Fourier transform (FFT), the swing angle θ frequency is the undamped natural frequency of the swing part. When it is 2 times less than the horizontal vibration frequency of human walking, the resonance of the system in the horizontal direction can be prevented.

$$R_s \sum F_{sp\tau}(t) + R_e \sum F_{e\tau}(t) + R_l m_l g \sin\theta(t) = R_l^2 m_l \ddot{\theta}(t) \tag{8}$$

Where $F_{e\tau}$ represents the component of the elastic band tension F_e in the tangential direction of the swing of the movable pulley of the shelf plate. Similarly, R_l, R_s, R_e denotes the radius of swing of the load, movable pulley and spring at the hanging point on the shelf plate.

Different from the vertical direction, a reduction in stiffness and damping in the horizontal direction has a beneficial impact on the vibration damping of the backpack. However, due to the limitations of spatial displacement in the horizontal direction when the human body walks, a reduction in stiffness and damping can also result in the backpack moving beyond the limit position, leading to additional structural collisions. The degrees of freedom in the horizontal direction of the backpack are primarily employed to mitigate the impact of inertial forces, facilitate the half-step vibration of the horizontal distance, and prevent the generation of resonance. A sufficient collision buffer at the limit position is also essential.

4. CONCLUSIONS

This paper presents a study of the motion law of the COM, based on the vibration phenomenon observed in the COM of a human walking. On the basis of the equivalent model of COM motion during walking, the Lissajous curve of COM motion is plotted for the most prominent coronal surface of COM vibration. The reliability of the equivalent model of walking COM motion is verified by comparing the model data with more than 100 groups of human walking experimental data.

This paper presents an investigation into the impact of inertial force and the phenomenon of extra bioenergy consumption in the context of a human back-loaded weight as a result of the vibration of the COM. This is achieved through the utilisation of an equivalent model of the COM motion. Furthermore, a dual-degree-of-freedom power suspension backpack system has been designed with the objective of reducing the vibration of the load in both the vertical and horizontal directions. In the simulation environment, a number of sets of measured walking data were loaded in order to mimic the human walking motion, with the aim of verifying the actual effect of the back-loading system under different individuals and different working conditions. The statistical results demonstrate that the backpack system effectively reduces the integrated inertia force in both the horizontal and vertical directions by 89.8% and 91.6%, respectively. Additionally, it markedly reduces the displacement in the vertical direction by 92.5% and in the horizontal direction by 42.5%. The damage to the body's inertial forces and the extra bioenergy consumption caused by weight-bearing walking were significantly reduced. The dual-degree-of-freedom power suspension backpack system, designed in this study, demonstrated a superior suspension damping effect in comparison to the single-degree-of-freedom vibration damping backpack.

The dual-degree-of-freedom power suspension backpack system, the subject of this paper, is designed with the primary objective of reducing the risk of sports injury and minimising bioenergy consumption when carrying heavy loads over extended periods. This is achieved by decoupling the movement of the backpack from that of the human body. However, due to weight and cost limitations, the backpack system's carrying cost is relatively high, and the control effect requires further optimisation. In the future, through more advanced ergonomics research, we anticipate being able to better address these issues.

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