A rehabilitation gait for the balance of human and lower extremity exoskeleton system based on the transfer of gravity center

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Abstract

Purpose – Most current lower extremity exoskeletons emphasize assistance for walking rather than stability. The purpose of this paper is to propose a rehabilitation gait based on the transfer of gravity center to improve the balance of exoskeleton rehabilitation training of the hemiplegic patients in the frontal plane, reducing the dependence on crutches/walking frames.

Design/methodology/approach – The real-time and predictable instability factors of human and exoskeleton system (HES) are analyzed. Inspired by the walking balance strategy of the blind, a rehabilitation gait based on the transfer of gravity center is proposed and studied by modeling and experimental test and is finally applied to the prototype – Zhejiang University lower extremity exoskeleton (ZJULEEX) – to verify its feasibility.

Findings – At least three real-time and predictable factors cause the instability of HES, and the factor of lateral tilt caused by gravity should be focused in the balance control of frontal plane. With the proposed gait, the hip height of stepping leg of HES does not reduce obviously even when the crutches do not work, which can improve the balance of HES.

Research limitations/implications – However, the rehabilitation gait control needs to be more complete and intelligent to response to other types of perturbations to further improve the balance of HES. In addition, more clinical trials should be conducted to evaluate the effect of the proposed gait.

Social implications – May bring happiness to the rehabilitation of patients with hemiplegia.

Originality/value – The rehabilitation gait based on the transfer of gravity center to improve the balance of HES is first proposed and applied to HES.

Keywords Balance, Center of pressure, Low limb exoskeleton, Transfer of gravity center

Paper type Research paper

1. Introduction

Robotic exoskeletons can be divided into three broad categories based on their intended use: human performance augmentation exoskeletons (Bogue, 2015; Brown *et al.*, 2003), assistive exoskeletons and therapeutic exoskeletons. With the development of science and medical technology, the research of therapeutic exoskeletons is maturing. Various prototype lower extremity exoskeletons (Yan *et al.*, 2015; Young and Ferris, 2017) have been developed, some of them have become commercially available, such as Ekso (Ekso Bionics, 2018), HAL (Cyberdyne, 2018), ReWalk (ARGO Medical Technology, Ltd., 2018) and Indego (Murray *et al.*, 2015). However, one limitation of the current exoskeletons is that they do not provide the function of maintaining the lateral stability of the coupled wearer exoskeleton system (Zhang *et al.*, 2018). The problem of how to design and control an exoskeleton that can effectively support the wearer-exoskeleton system's stability during upright locomotion remains an open question, which greatly constrains the development of lower extremity exoskeleton.

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For medical purposes, several solutions for the balance of human and exoskeleton system (HES) in the sagittal or frontal plane have recently been proposed. The most common one is that patients manipulate crutches/walking frames to partially support body weight and keep balance, such as Ekso, HAL, ReWalk and Indego, which is high energy costing, requiring patients with sufficient upper body strength. Many exoskeletons are placed in a support polygon with suspension system to support patients' body weight, such as the Active Pelvis Orthosis (Monaco *et al.*, 2017) and Lokomat (Hidler *et al.*, 2009). Although the balance of HES can be greatly guaranteed, the patient's movement is limited due to the large frame size. Li *et al.* (2015) designed an additional balance stabilizer mechanism and Munawar H *et al.* (2016) designed an balance trainer called AssistOn-Gait to enhance the stability. However, the balance stabilizer mechanism and the balance trainer limit the mobility of the trainer. Zhang T *et al.* (2018) designed a hip exoskeleton with powered hip abduction/adduction and hip flexion/extension joints and used a balance controller based on the extrapolated center of mass (XCoM) concept to maintain walking stability in both the sagittal and frontal planes. However, at present, it is effective mainly for the patients who have muscular weakness but retain voluntary lower-limb motor control. The design and control of exoskeleton joint with variable stiffness was studied, such as in XoR (Ugurlu *et al.*, 2016) and in a knee exoskeleton (Ajoudani, 2016). REX (REX Bionics, 2018) can perform balance control by itself with enough powered joints, which makes the exoskeleton large and heavy. In addition, the limb movements of the patient are completely driven by REX during rehabilitation training, thus reducing the patient's training initiative and enthusiasm significantly. In terms of control, to improve balance of HES, human movement intentions are identified in real time to control the exoskeleton, for example, HAL measures EMG signals and ReWalk measures upper body inclination. Balance control based on capture point concept and human balance strategies (Huynh *et al.*, 2016) and dynamic balance gait planning method (Chen *et al.*, 2018) were studied. These control methods can help to enhance the balance of HES in the sagittal plane, but cannot overcome the lateral tilt problem caused by gravity, and still require a frame or a crutch to achieve balance in the frontal plane.

When it comes to the hemiplegic patients' rehabilitation training with exoskeleton, it's more difficult to make HES balance in the frontal plane, because the hemiplegic patients are unable to use two crutches symmetrically to keep balance since their half body have physical movement disorders. An important feature of exoskeleton systems, which distinguishes them from bipedal robots, is the participation of the wearer in the locomotion, balance control, and decision-making processes (Ugurlu *et al.*, 2016). To make patients with hemiplegia can get better balance in exoskeleton rehabilitation training, the wearer's effort in balance recovery was considered in the paper.

A loss of balance can be induced by a number of perturbation events, such as slipping, tripping and stumbling (Zhang *et al.*, 2018). However, these disturbances are often sudden and unpredictable, and there is no systematic analysis of the real-time and predictable instability factors of HES during walking. We initially analyzed these instability factors and found that the lateral overturning caused by gravity during locomotion is an important cause of instability of the HES in the frontal plane. How to overcome the lateral instability caused by gravity with the teamwork of the wearer and the exoskeleton was studied in the paper.

In the clinical rehabilitation training of hemiplegic patients, we found that most of them had a certain ability to tilt their upper body, they could simply change the position of the center of gravity by tilting the upper body sideways or forward or backward. Inspired by the phenomenon that the blind tilts his upper body to the side of support leg to keep balance more easily when he uses swing leg to detect ground conditions during walking, a rehabilitation gait based on the transfer of gravity center with the participation of the wearer is proposed to solve the balance problem of HES in the frontal plane. The main idea of the rehabilitation gait is guiding the changes of the wearer's upper body posture to balance the HES in the frontal plane. In detail, before the patient's hind leg swings off the ground, the exoskeleton drives the patient's foreleg upright and the patient actively tilts the upper body to the side of the support leg, and patients are expected to keep the state until the heel of the swinging leg touch the ground to ensure that most of the weight of HES is supported by the HES upright leg during the swinging process, thereby overcoming the lateral tilt problem caused by gravity to improve the balance of the HES in the frontal plane.

The contributions of this paper are as follows:

- The real-time and predictable instability of HES was analyzed.
- A rehabilitation gait for the balance of HES in the frontal plane based on the transfer of gravity center was proposed and initially verified.

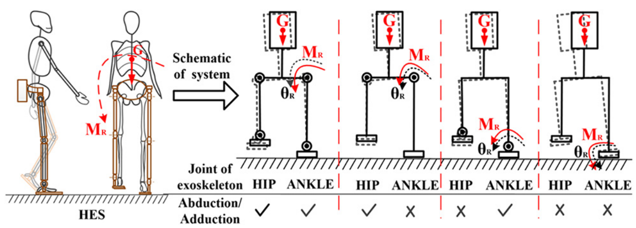
2. The analysis of instability of human exoskeleton system

We divided the relationship between HES and the environment into three levels: human and exoskeleton, human and environment, exoskeleton and environment. There are interactions between human and exoskeleton and predictable or unpredictable effects between human and the environment as well as exoskeleton and environment. Without considering the unpredictable perturbation effects, there are three main causes of HES instability: the lateral tilt caused by gravity, human-machine coordination, and the collision between the exoskeleton and the ground caused by the exoskeleton landing in advance.

2.1 The lateral tilt caused by gravity

In walking process, conventional people can achieve balance by controlling the CoM of the body (Inman and Eberhart, 1953; Perry and Davids, 1992). However, it is difficult to control the position of CoM independently to achieve balance for hemiplegic patients. Besides, for most existing low limb rehabilitation exoskeleton products, their hip's degree of adduction/abduction and ankle's degree of eversion/inversion are often passive or fixed to reduce the cost, quality, overall size and controlling difficulty, resulting that the exoskeleton cannot overcome the lateral tilt caused by gravity by itself. Figure 1 shows static analysis of the gravity-caused HES instability.

Figure 1 The static analysis of gravity – caused HES instability



The HES is subjected to a tilting moment M_R to the side of the swing leg caused by gravity:

$$\theta_R = \lim_{\Delta t \to 0} \sum_{i=1}^{n} \Delta\theta_{Ri} = \lim_{\Delta t \to 0} \sum_{i=1}^{n} \left(\omega_{Roi}\Delta t_i + \frac{1}{2}\beta_{Ri}\Delta t_i^2 \right) \tag{1}$$

$$T = \lim_{\Delta t \to 0} \sum_{i=1}^{n} \Delta t_i \tag{2}$$

T is the swing time of swing leg, θ_R is the lateral tilt angle of HES in time T, ω_{R0i}, β_{Ri} are the initial angular velocity and angular acceleration in period Δt_i respectively. According to formula (1), to reduce gravity impact, the swing speed is often increased to reduce the swing time. However, swing speed needs to be determined according to the specific conditions of the patient, because fast speed may easily lead to patient's discomfort and HES's instability.

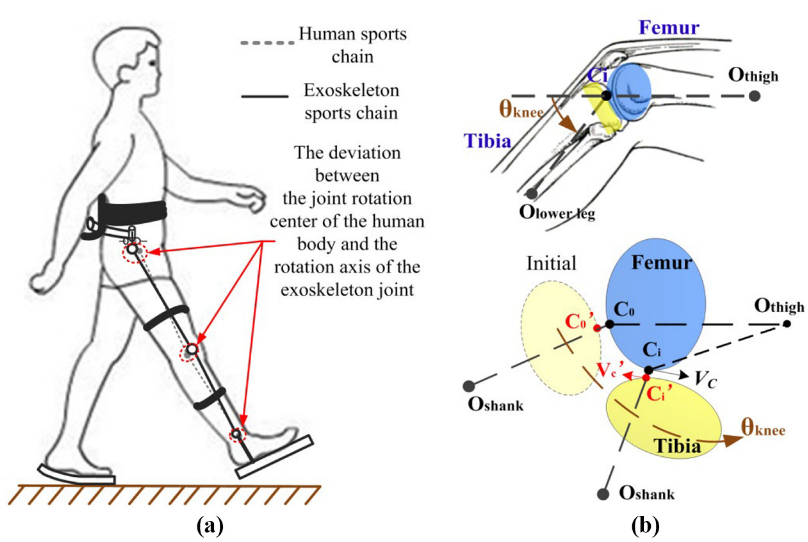
2.2 Human–machine incoordination

When doing rehabilitation training with exoskeleton, patient is not only the load but also a kind of disturbance for exoskeleton. The incoordination between patient and exoskeleton can lead to excessive human-exoskeleton interaction which will interfere with the movement of exoskeleton. Human-machine incoordination is mainly reflected in three aspects.

The first one is the incoordination in size, which means that the exoskeleton does not match the specific patient size. The second one is the incoordination in movement. In the process of rehabilitation, the exoskeletons drive the movement of patients' lower extremities, but many patients have more or less limb exercise ability. If the movement of the exoskeleton is very different from the movement of the patient, taking an exaggerated example, one of the patient's legs tries to stay upright, but the corresponding leg of the lower extremity exoskeleton tries to do flexion movement, then HES will easily lose stability. The third one is the incoordination in wear, specifically the misalignment of exoskeleton joints and human joints, as shown in Figure 2(a). Traditional exoskeleton robots generally use simplified single-degree-of-freedom hinge joints or three-degree-of-freedom ball joints to match human physiological joints. Such mechanical joints are simple, easy to implement and control, but it is difficult to completely and accurately match the human joints. When the center of joint of the exoskeleton robot mismatch the center of joint of the human body, the kinematic chains of exoskeleton and human limb cannot be completely consistent, and the uncontrollable interactive force between human and exoskeleton will emerge due to the rigid human-machine physical contact interface. It may exceed the range that stroke patients can tolerate. At the same time, the movement of exoskeleton robots and human bodies cannot be effectively controlled, and the requirements for safety and comfort cannot be met. Taking the knee joint as an example, as shown in Figure 2(b), there are two kinds of movements between femur and tibia in the knee joint: rolling and sliding (Wang *et al.*, 2014), however, many knee joint of lower extremity exoskeletons are often implemented with a single revolute joint. Therefore, the consistency of rotation centers of exoskeleton knee joint and the patient's knee joint are difficult to maintain.

There are many ways to solve this problem for reference. The exoskeleton with length-adjustable limbs can avoid the

Figure 2 The incoordination in wear of HES



Notes: (a) The mismatch between exoskeleton and human joints; (b) rolling/sliding model of knee joint incoordination in size. Optimize joint design, such as bionic joint design (Yang *et al.*, 2014) and flexible joint design (Lee and Wang, 2015), which improves similarity between exoskeleton and patient joint, was proposed to solve the incoordination in wear. To improve the coordination in movement, intelligent control with recognizing the patient's exercise intentions and joint torque flexibility control (Li *et al.*, 2017) and other control strategies were studied.

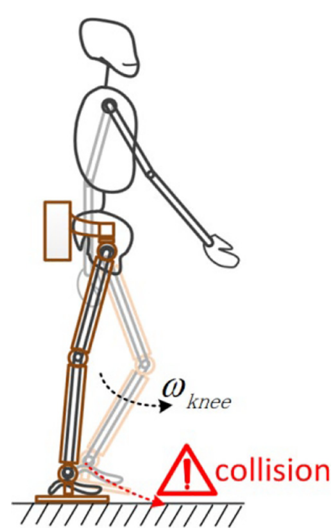
2.3 The collision between the exoskeleton and the ground caused by the exoskeleton landing in advance

For the exoskeleton whose joint motor moves mainly follow the preset angle curve, the hip height of stepping leg of HES will reduce during the swing process because of the effect of gravity, as shown in Figure 1, thus causing a phenomenon that the swing leg land in advance in the expected swing process as shown in Figure 3. In addition, the phenomenon will be caused by other factors such as uneven ground. Bringing the collision between the exoskeleton and the ground, this phenomenon will greatly affect the balance of HES.

3. The rehabilitation gait based on the transfer of gravity center

To improve the balance of the hemiplegic patients' rehabilitation training reducing walking frames use and without introducing additional complex or large balance stabilizer

Figure 3 The collision caused by landing in advance of HES



mechanism to exoskeleton, how to reduce the influences of the lateral overturning caused by gravity with the transfer of the patients' gravity center is the key, which is the main purpose of the proposed gait. The proposed rehabilitation gait is shown in Figure 4, and the right leg is represented by a solid line and the left leg is represented by a dotted line.

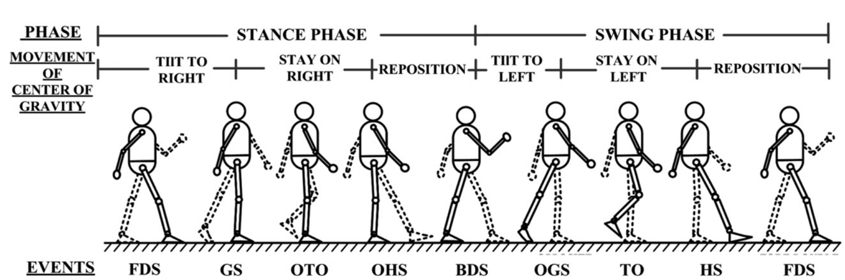
According to the changes of gait information, we divide the gait cycle into eight events:

1. *Fore double support (FDS*: the double support phase with right foot on the front and the body weight evenly supported by the two legs.
2. *Gravity shift (GS)*: the right leg is almost upright, the upper body of is tilted to the right side and the center of gravity is transferred to the right side. The left footpad gently support patient, and most of the body's weight is supported on the right leg.
3. *Opposite toe off (OTO)*: the right leg is almost upright, the upper body of the body stays on the right side, and the left foot toe swings off the ground, and the weight is completely supported by the right leg.
4. *Opposite heel strike (OHS)*: the right leg is still almost upright, the upper body of the body stays on the right side, and the left heel gently touches the ground, and most of the body's weight is supported by the right foot.
5. *Bake double support (BDS)*: the double support phase with left foot on the front and the body weight evenly supported by the two legs.
6. *Opposite gravity shift (OGS)*: the left leg is almost upright, the upper body of is tilted to the left side and the center of gravity is transferred to the left side. The right footpad gently support patient, and most of the body's weight is supported on the left leg.
7. *Toe off (TO)*: the left leg is almost upright, the upper body of the body stays on the left side, and the right foot toe swings off the ground, and the weight is completely supported by the left leg.
8. *Heel strike (HS)*: the left leg is still almost upright, the upper body of the body stays on the left side, and the right heel gently touches the ground, and most of the body's weight is supported by the left foot.

3.1 The study of the rehabilitation gait based on the transfer of gravity center

The dip angle of upper body in frontal plane (θ) of the proposed gait should not be large for the feasibility and comfort of the gait, and we analyzed it by mathematical modeling and actually measure the gait information of a number of healthy

Figure 4 The cycle of the rehabilitation gait based on the transfer of gravity center



subjects when they simulate the gait to verify the model, and apply the model to the gait analysis of the training exercise with exoskeleton.

3.1.1 The analysis of the gait

When a person tilted his upper body to the side of support leg, the degree of ankle eversion/inversion (AEI) of support leg also rotated. We analyzed the gait in the ground coordinate system as shown in Figure 5. To make the gait have the optimal stability margin, the projection of the human body's CoM on the ground should fall on the supporting foot midline:

$$X_C = \frac{\sum m_i X_i}{M} \tag{3}$$

$$X_{C-Optimum} = X_{sfm} \tag{4}$$

$$X_{sfm} = W_{fs}/2 + W_f/2 \tag{5}$$

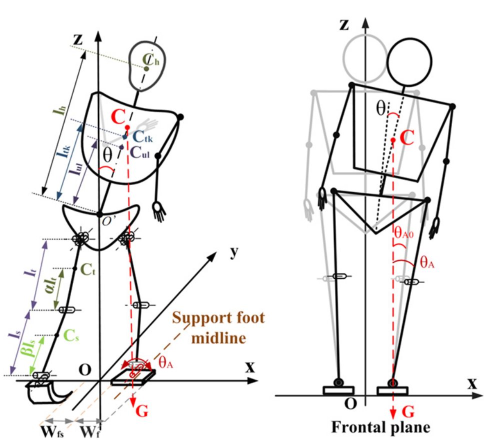
X_C is the abscissa value of the human body's CoM, m_i, X_i are the mass of each part of the human body and the abscissa value of its mass center respectively, M is the total body mass, X_{sfm} is the x-coordinate value of the supporting foot midline, W_{fs} is the gap between feet in the frontal plane, W_f is the foot width.

According to the symmetry of body, the body is divided into six parts: head and neck, trunk, upper limb, thigh, shank and foot. To simplify the formula, *s* means sin, *c* means cos. Then:

$$\begin{aligned}
X_C &= \frac{m_h X_h + m_{tk} X_{tk} + m_{ul} X_{ul} + m_t X_t + m_s X_s + m_f X_f}{M} \\
&= \frac{m_h(l_h s\theta + \Delta d) + m_{tk}(l_{tk} s\theta + \Delta d) + m_{ul}(l_{ul} s\theta + \Delta d) + m_t(l_s + \alpha l_t) c\theta_{A0} s\Delta\theta_A + m_s \beta l_s c\theta_{A0} s\Delta\theta_A}{M} \\
\Delta d &\approx L(s\theta_A - s\theta_{A0}), \quad \Delta\theta_A = \theta_A - \theta_{A0}, \quad L = l_t + l_s \\
X_C &= A s\theta_A + B s\theta_{A0} + C s\Delta\theta_A + D \\
A &= \frac{m_h l_h + m_{tk} l_{tk} + m_{ul} l_{ul}}{M} \\
B &= \frac{(m_h + m_{tk} + m_{ul})L}{M} \\
C &= \frac{[m_t(l_s + \alpha l_t) + m_s \beta l_s] c\theta_{A0}}{M} \\
D &= -\frac{(m_h + m_{tk} + m_{ul}) L s\theta_{A0}}{M}
\end{aligned} \tag{6}$$

m_h, X_h, m_{tk}, X_{tk}, m_{ul}, X_{ul}, m_t, X_t, m_s, X_s, m_f, X_f are the mass of the head and neck and its centroid (C_h) coordinate, the mass of the trunk and its centroid (C_{tk}) coordinate, the mass of the upper limb and its centroid (C_{ul}) coordinate, the mass of the thigh and its centroid coordinate, the mass of shank and its centroid coordinate, the mass of foot and its centroid coordinate, respectively. $X_f = 0$. l_h, l_{tk}, l_{ul} are the distances from C_h, C_{tk}, C_{ul} to the point of rotation of the pelvic bone (O') respectively. l_t, l_s, L are the length of the thigh, shank and the sum of them respectively. θ is the dip angle of upper body in frontal plane, and positive value means the upper body tilts to the right. θ_A is the angle of AEI, and positive value means the right ankle rotates right and left ankle rotates left. θ_{A0} is the initial value of θ_A when the person stands on both feet, and it will change with the change of W_{fs}. α is the ratio of the distance between the thigh's CoM (C_t) and the tibia point to the length of thigh, β is the ratio of the distance between the shank's CoM (C_s) and the internal malleolus point to the length of thigh. The parameters can be found in the Figure 5. When the parameters (dimensions, masses and centroids of various parts of the human body) refer to the second edition of "MODERN SPORTS BIOMECHANICS":

Figure 5 The static analysis of the proposed rehabilitation gait



$$\begin{aligned}
X_{C-male} &= 0.09418Hs\theta + 0.3133Hs\theta_A + 0.1150Hc\theta_{A0}s\Delta\theta_A \\
&\quad - 0.3133Hs\theta_{A0} \\
X_{C-female} &= 0.08948Hs\theta + 0.3017Hs\theta_A + 0.1187Hc\theta_{A0}s\Delta\theta_A \\
&\quad - 0.3017Hs\theta_{A0}
\end{aligned} \tag{7}$$

H is the height of the person. If the gap between the feet (W_{fs}) is too large, the dip angle of upper body in frontal plane ($\boldsymbol{\theta}$) will be large, which will make the person discomfort. Most of people will make W_{fs} as small as possible during walking. Usually, $W_{fs} = 0.2 \sim 1 W_f$.

3.1.2 Sample data collection and analysis

We let healthy volunteers simulate the proposed gait and master it, and collected the gait information of the volunteers with the proposed gait. The gait information acquisition system is shown in Figure 6. The inertial unit movement measurement units were used to measure dip angle of the upper body ($\boldsymbol{\theta}$), the flexion/extension angle of hip and knee, and the angle of ($\boldsymbol{\theta_A}$). The plantar pressure was measured by the plantar pressure collection shoe. The layout of plantar pressure collection unit refers to (Yu *et al.*, 2010). We placed marking lines on the ground with 2, 5 and 10 cm spacing, and collected related gait information when W_{fs} was 2, 5 and 10 cm.

Obtaining actual ZMP of HES is practically difficult. The Center of Pressure (CoP) method is used instead to obtain the ZMP-like data. As shown in Figure 7, we calculate the x-coordinate values of CoP through the plantar pressure information:

$$CoP_x = \frac{\sum_{i=1}^{8} X_i N_i}{\sum_{i=1}^{8} N_i} \tag{8}$$

Figure 6 Gait information acquisition system

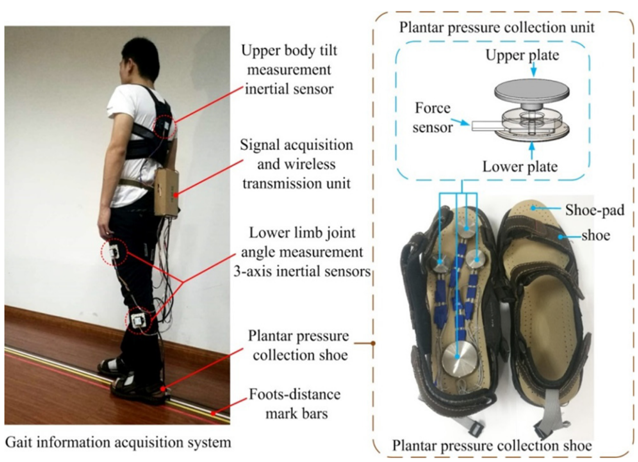
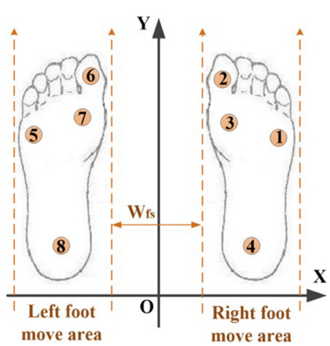


Figure 7 The calculation of CoP with 8 plantar pressure sensors



X_i, N_i is the x-coordinate value of the pressure sensor $\boldsymbol{i}$ and its pressure measurement value, respectively.

Figure 8 shows the gait information of a healthy volunteer with the conventional gait and the proposed gait. The gait data shows that during normal walking, people will unconsciously tilt the upper body to side of support leg slightly to get a better balance. In addition, we can find that there are two to three main differences between the two gaits. One is that the hip angle of the proposed gait has three relatively straight stages:

1. *Stage 1*: From the GS phase to the OHS phase, the support leg holds approximately upright until the gravity center has been transferred to the support leg and the heel of swing leg has touched the ground.
2. *Stag 2*: From the BDS phase to OGS, the hind leg waits until the center of gravity has been transferred to the support foot.
3. *Stag 3*: From HS phase to FDS, legs stay still until the upper body has repositioned.

Other difference is that the dip angle of upper body of proposed gait is larger than the conventional gait. The last difference is that the change of CoP_{gs} lags slightly behind the change of θ_{gs}. This is because when the volunteer tilts his upper body to side of the front leg, the hind leg tends to active support his body. Only when the center of gravity has been shift to the support leg stably, the hind leg will not work hard, thus ensuring a better balance.

Table I shows the physical parameters of the sample persons, the measured related gait information, and the theoretically calculated dip angle of upper body (θ_{Ogs_t}) when the proposed gait has the optimal stability margin. The measured optimal values of dip angle of upper body (θ_{Ogs_a}) are not large (when W_{fs} = 10 cm, $\theta_{Ogs_a} \approx 20°$; when W_{fs} = 5 cm, $\theta_{Ogs_a} \approx 15°$; when W_{fs} = 10 cm, $\theta_{Ogs_a} \approx 12°$), and are similar to the optimal values calculated by the model (θ_{Ogs_t}), which verifies that the model of the proposed gait has a certain degree of correctness.

3.1.3 Analysis of $\boldsymbol{\theta}$ *when patients wear exoskeleton to follow the proposed gait*

Referring to the analysis of the gait mentioned previously, to make HES have the optimal stability margin during walking, the projection of the HES centroid on the ground should fall on the support foot midline:

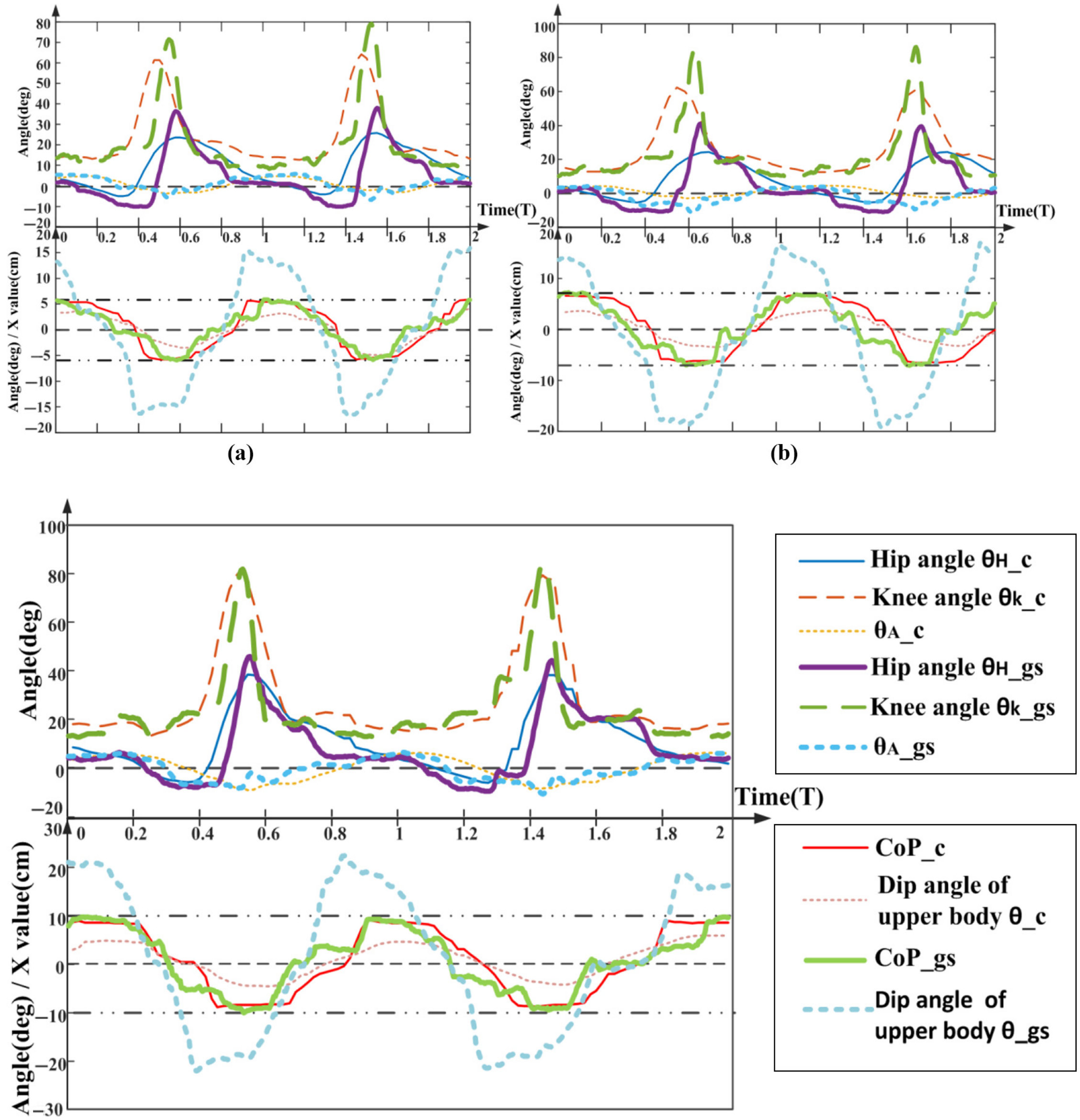
$$X_{C-HES} = \frac{\sum m_i X_i + \sum m_i' X_i'}{M + M'} \tag{9}$$

$$X_{C-HES-Optimum} = X_{sfm}' \tag{10}$$

X_{C-HES} is the abscissa value of HES centroid. m_i', X_i' are the mass of the various parts of the exoskeleton and the abscissa value of its centroid, respectively. M' is the total exoskeletal quality, X_{sfm}' is the abscissa value of the midline of the support foot:

$$X_{C-HES} = \frac{M X_C + M' X_C'}{M + M'} \tag{11}$$

Figure 8 Sample gait information (parameters with subscript *_c* belong to the conventional gait, parameters with subscript *_gs* belong to the proposed rehabilitation gait)



Notes: (a) $W_{fs} = 2\ cm$, $Wf = 10\ cm$; (b) $W_{fs} = 5\ cm$, $Wf = 10\ cm$; (c) $W_{fs} = 10\ cm$, $Wf = 10$ cm

$$\theta_{Optimum} = arcsin\left[\frac{M\left(X_{sfm}{}' - Bs\theta_A - Cs\Delta\theta_A - D\right) + M'\left(X_{sfm}{}' - B's\theta_A - C's\Delta\theta_A - D'\right)}{MA + M'A'}\right] \tag{12}$$

X_C' is the abscissa value of exoskeleton centroid. A,B,C,D are the intrinsic parameters of the human body, refer to formula (6). A', B', C', D' are the intrinsic parameters of the exoskeleton, which are similar to A, B, C, D, respectively. According to Formula (12), to make the patient's dip angle of upper body as small as possible, the quality of the exoskeleton should be as small as possible, and at the same time, the gap between feet in the frontal plane (W_{fs}) of HES

Table I Sample gait information statistics table

Subjects Gender		1 Male	2 Male	3 Male	4 Male	5 Male	6 Male	7 Female	8 Female
Height/cm		170	171	173	176	160	172	167	163
Weight/kg		73	64.5	57	60	57.5	81.5	55	53
range of θ_c	$W_{fs}=10cm$	−6~6	−6~6	−5~5	−3.5~3.5	−5~5	−5~5	−2.5~2.5	−4~4
	$W_{fs}=5$cm	−5~5	−4~4	−4.5~4.5	−3~3	−4-4	−4.2~4.2	−2~2	−3~3
	$W_{fs}=2$cm	−4~4	−3~3	−4~4	−2.5~2.5	−3.5~3.5	−3~3	−1.8~1.8	−2~2
range of θ_gs	$W_{fs}=10$cm	−17.5~17.2	−22~22.5	−20~20	−19~19	−19.5~20	−18~17.8	−17.1~16.8	−16~15.7
	$W_{fs}=5$cm	−15.2~14.6	−16.5~16.8	−16~15.5	−16.2~16.5	−15.2~15.9	−15.2~15.3	−13.8~14.2	−13.1~12.5
	$W_{fs}=2$cm	−12.1~12.5	−14~14	−12.5~12.5	−12.5~13	−13.5~14	−13.1~12.9	−11.2~12	−11~12
θ_{A0}	$W_{fs}=10$cm	1.84	1.88	1.90	1.96	1.63	1.88	1.85	1.76
	$W_{fs}=5$cm	3.53	3.55	3.56	3.59	3.43	3.55	3.56	3.52
	$W_{fs}=2$cm	4.54	4.56	4.56	4.57	4.5	4.56	4.60	4.57
θ_A_Ogs	$W_{fs}=10$cm	5.80	5.38	5.40	5.36	5.28	5.68	6.05	6.26
	$W_{fs}=5$cm	5.93	5.62	5.71	5.79	5.86	5.98	6.16	6.28
	$W_{fs}=2$cm	6.33	6.11	6.36	6.35	6.30	6.26	6.80	6.77
θ_Ogs_a	$W_{fs}=10$cm	17.1	20.1	19.5	18.5	19.3	17.2	16.3	15.1
	$W_{fs}=5$cm	14.2	16.2	15.0	15.2	15.1	14.6	14.4	13.8
	$W_{fs}=2$cm	11.1	13.1	11.5	11.5	12.1	12.5	10.7	10.2
θ_Ogs_t	$W_{fs}=10$cm	16.72	18.98	18.75	18.92	19.55	17.44	17.35	16.38
	$W_{fs}=5$cm	14.90	16.45	15.98	15.60	15.4	14.72	15.2	14.66
	$W_{fs}=2$cm	12.30	13.40	12.18	12.22	12.48	12.67	11.22	11.32

Notes: θ_c: the dip angle of upper body of the conventional gait; θ_gs: the dip angle of upper body of the proposed gait; θ_A_Ogs: the angle of AEI when the proposed gait has the optimal stability margin θ_Ogs_t: the theoretical dip angle of upper body when the proposed gait has the optimal stability margin; θ_Ogs_a: the measured dip angle of upper body when the proposed gait has the optimal stability margin

should be adjusted to a smaller value without motion interference.

4. The control of the rehabilitation gait

To solve the instability issues caused by HES's landing in advance due to the reduce of the hip height of the HES's stepping leg caused by the gravity, we propose a rehabilitation gait based on the transfer of gravity center. The flow chart of the gait control is shown in Figure 9.

The first step is to input the parameters related to the user. To fit different subject conditions and preferences, the parameters listed in Table III are made to be adjustable.

In the second step, according to the input parameters and the collected sample gait database of the proposed gait, polynomial interpolation method is used to fit the gait trajectory. The trend of changes of the angle of the hip and knee joints of this gait is shown in Figure 10. According to Formula (12), the optimal value of the dip angle of upper body ($\theta_{optimum}$) will be calculated.

The third step is to collect the related gait information in real time by the sensory system, and perform amplification, noise reduction, and discretization processing on the collected data.

The fourth step is to make decision and control the movement of the exoskeleton. According to the collected plantar pressure, CoP coordinate is calculated, and whether CoP is in the stable area will be judged. Because the crutch does not work in the ideal situation, the ideal stable area is a well-known support polygon in double support phase and the support surface of the supporting foot in single support phase, both with an additional safety boundary of 15 mm. If the CoP is out of the stable area, HES will go to the double support phase in a reasonable way as fast as possible according to the current state. If the CoP is in the stable area, the current gait phase will be judged. If the current phase is the special phase, gravity shift, whether the center of gravity shift is already completed is determined according to the currently collected plantar pressure and the dip angle of upper body (**θ**). Exoskeleton will guide the user to tilt his upper body to the side of support leg properly. If the center of gravity has been transferred to the support leg, the hind leg will step. Otherwise, HES follows the preset gait trajectory.

In Step 5, whether the training is completed will be determined. If the training is completed, HES will move to the double-legs-upright state with a reasonable trajectory, and then remind the user. Otherwise, HES will reciprocate.

5. An auxiliary game guidance system for the rehabilitation gait

To let the patient tilt his upper body more naturally, we designed a game guidance system with visual and sound guidance, as shown in Figure 11(a). According to the theoretical model mentioned above and the gait information collected in real time, we can calculate the optimal upper body posture A of the patient under the current gait phase. The image guiding interface will display the ideal upper body posture A and the patient's current upper body posture B in real time. The patient can change his upper body posture to let B align A, and in this way, patient can easily find the suitable range of the dip angle of upper body in frontal plane. The main function of sound guidance is to remind and encourage patient to tilt his upper body, and praise the

Figure 9 The control flowchart of the proposed rehabilitation gait

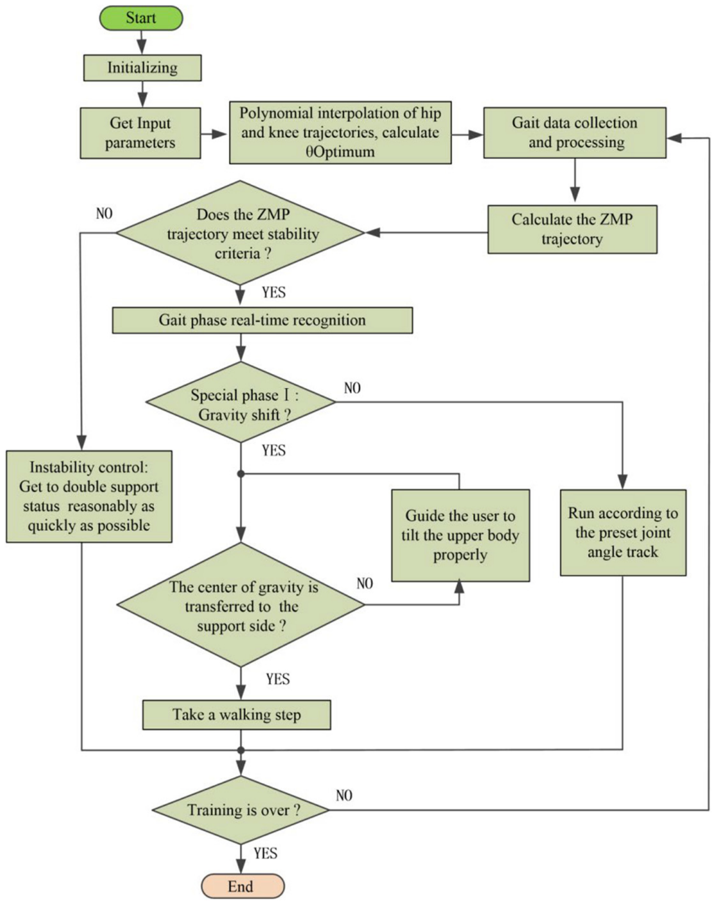


Table II The input parameters

Gender	Height	Weight	Gap between feet	Thigh length	Shank length	Step length	Step time
Male/female	H	W	W_{fs}	l_t	l_s	L_{step}	T_{step}

Table III The proposed rehabilitation gait test result of five subjects

Subjects	Gender	Age	Height/cm	Weight/kg	L*step*/mm	$\overline{\lvert\theta_a\rvert}$/°	$\overline{\lvert CoP_x\rvert}$/cm	θ_t/°
1	male	26	171	64.5	282	18.2	9.1	20.3
2	male	29	170	73	283	18.9	9.2	20.6
3	male	25	176	60	285	19.7	9.6	22.5
4	male	24	160	57.5	276	17.8	8.9	19.8
5	female	23	163	53	275	18.7	9.05	20.2

patient's movements. In the game training mode, we set the recent rehabilitation target according to the patient's condition, specifically the range of the upper body roll angle, which is set as the game level. The system will choose different compliments such as "good", "great", "amazing", "crazy", "unbelievable" based on number of coincidences and coincidence degree of A and B, thus increasing the user's interest and enthusiasm to do rehabilitation training. Furthermore, we only perform upper body tilting training for patient in the game guidance system before the gait training, as shown in the Figure 11(b). After the patient gets used to it, gait rehabilitation training based on gravity shift will be conducted.

6. Mechanical and electrical hardware structure design of exoskeleton

To make the control method proposed in this paper applicable to most of the existing lower extremity exoskeleton products,

Figure 10 Hip and knee motions of the proposed rehabilitation gait

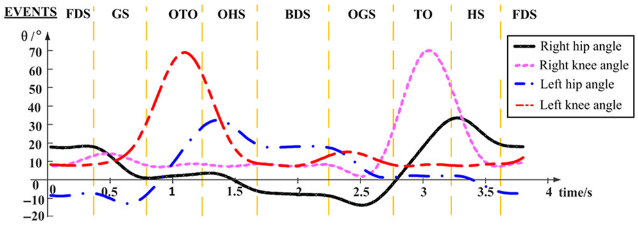
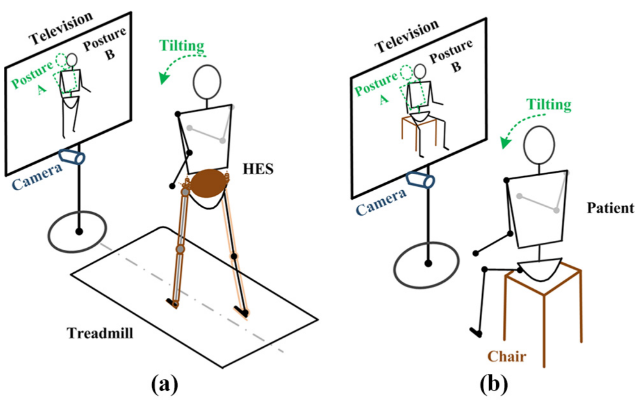


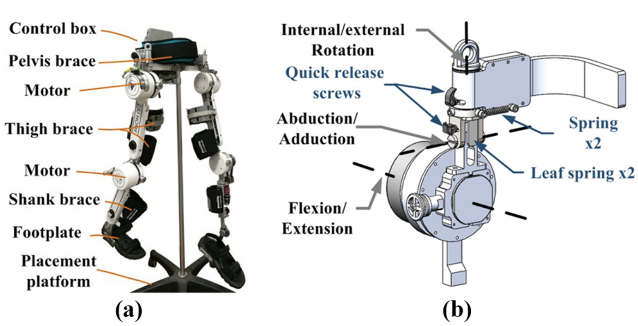
Figure 11 The rehabilitation gait training with the game guidance system



Notes: (a) Walking training with the rehabilitation gait; (b) upper body tilting training

referring to human bionics and the existing exoskeleton products, the Zhejiang University lower extremity exoskeleton (ZJULEEX) was designed as shown in Figure 12(a). ZJULEEX has 6 DOF per leg. As shown in Figure 12(b), the hip joint has 3 DOFs: abduction/adduction, internal/external rotation, and flexion/extension. The flexion/extension joint is driven by Maxon disc motor with harmonic reducer, the abduction/adduction and the internal/external rotation of hip are passive with spring components which make them achieve automatic restoration when the external force is small. Besides we can adjust whether the two passive degrees of freedom are locked according to the patient's conditions and training tasks by using quick release screws, so as to make ZJULEEX more comfortable to wear. The joint of knee flexion/extension is active. The joint of ankle flexion/extension is passive. In addition, the joint of ankle eversion/inversion is passive with -10-$10°$ range of motion. To reduce the exoskeleton mass, most of the parts use carbon fiber materials. ZJULEEX weights 17.5 kg, and is suitable for 155-180 cm users.

Figure 12 The mechanical structure of ZJULEEX



Notes: (a) ZJULEEX; (b) hip structure of ZJULEEX

The electrical hardware structure of ZJULEEX is shown in Figure 13, when the patient wears the exoskeleton for rehabilitation training, the upper computer receives the control signal from the controller and the related gait information measured by the sensory measurement system, and does related signal data processing, and then sends the related control command to the corresponding motor driver, which controls the movement of motor. Sensors in the sensing system include eight thin film pressure sensors for measuring the plantar pressure, a gyroscope for measuring the dip angle of the upper body and angle sensors for measuring the rotation angle of the users' joints.

7. Results

In this section, we describe the experiments conducted to characterize the performance of the proposed rehabilitation gait. By comparing with the experiment results of traditional balance method with crutches, effectiveness of the proposed rehabilitation gait has been initially verified.

The test subjects were approved by the Institutional Review Board of Zhejiang University and informed consents were obtained from the subjects. Subjects were asked to simulate the proposed gait and master it. To ensure safety, there were two safety assistants to monitor the locomotion of the HES without providing the support force. In the experiment, W_{fs} of 10 cm and step time of 2.5s were set and the subjects were asked to walk 10 meters for 5 times.

The proposed rehabilitation gait test on a healthy subject was conducted as shown in Figure 14. In the test, the subject tilted his upper body to the side of support leg as expected, thus HES can receive a better balance. It was worth noting that the crutches in the Figure 14 only played an auxiliary protective role in the case of imbalance. Without the support by the crutches, the CoP trajectory of HES can be

Figure 13 The electrical hardware structure of ZJULEEX

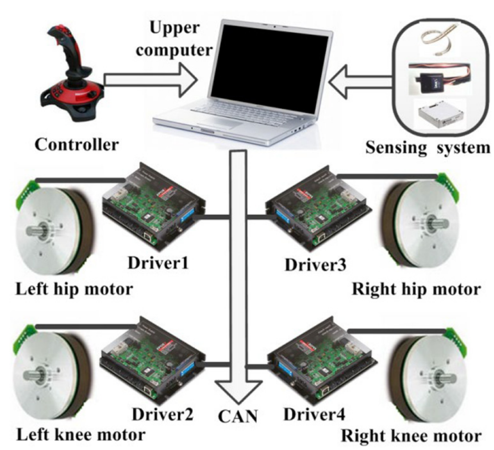


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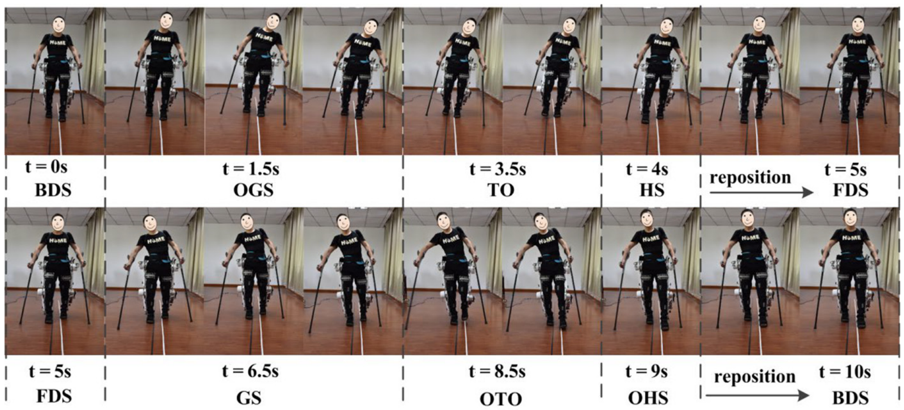
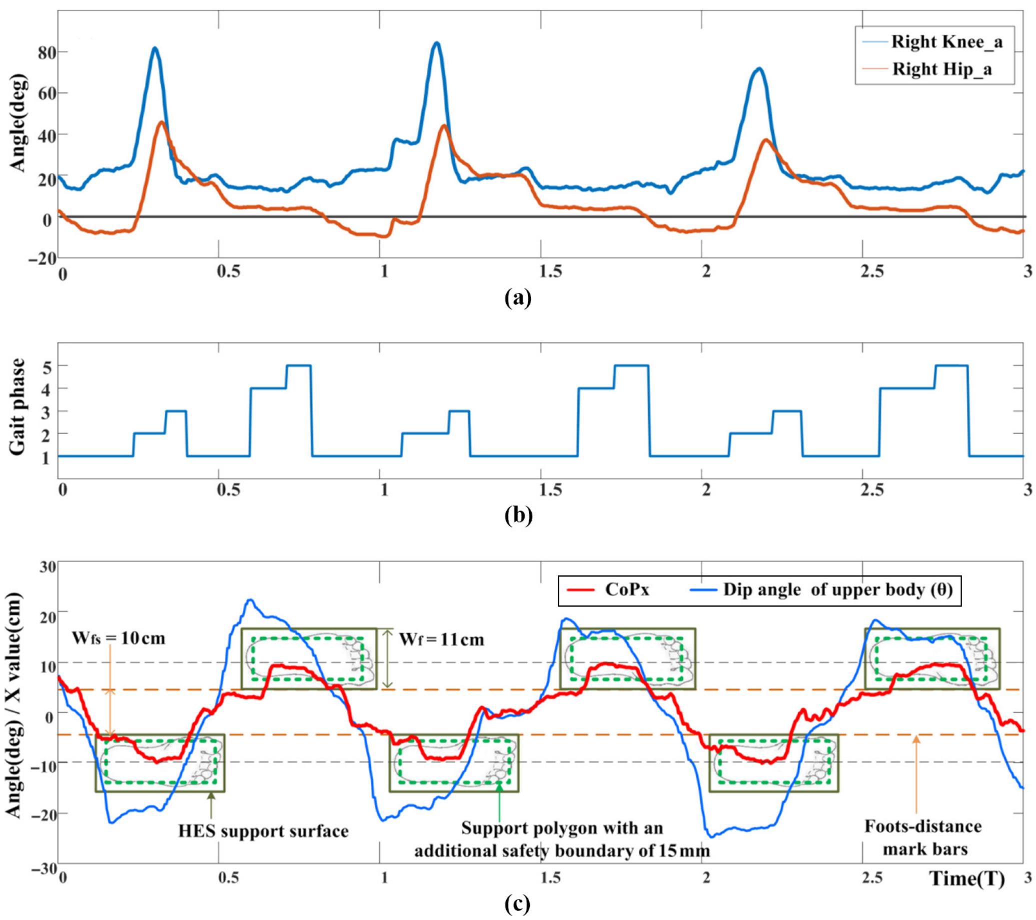
Figure 14 The proposed rehabilitation gait test on a healthy subject



monitored by measuring the plantar pressure with the plantar pressure collection shoe. Result of the proposed rehabilitation gait test was shown in Figure 15. The upper body will be tilted to the side of the support leg before swing phase, which can be seen by the curve of the dip angle of upper body in Figure 15(c). The gait phases in Figure 15(b) and the curve of the X-coordinate value of CoP (CoP_x) in Figure 15(c) showed that the CoP of HES is in the stable

Figure 15 Results of the proposed rehabilitation gait test



Notes: (a) Actually measured joint angles; (b) gait phases: 1- double support, 2- right-leg early swing (flexion), 3- right-leg late swing (extension), 4- left-leg early swing (flexion), 5- left-leg late swing (extension); (c) dip angle of upper body and x-value of CoP support polygon during the swing period, and the maximum absolute value of CoP_x is approximately equal to 10 cm and is close to the optimal value (10.5 cm), which demonstrates that HES can receive a good balance with the proposed rehabilitation gait.

Five healthy subjects participated in the proposed rehabilitation gait test experiment with ZJULEEX, and the results are showed in the Table III. In Table III, $\overline{|\theta_a|}$, $\overline{|CoP_x|}$ are the average value of the maximum absolute value of actually measured dip angle of upper body (θ) and the average values of the maximum X-coordinate value of CoP (CoP_x) during the swing period respectively, θ_t is the calculated theoretical value of the dip angle of upper body according to $\overline{|CoP_x|}$ and Formula (12). The deviation between $\overline{|\theta_a|}$ and $\boldsymbol{\theta_t}$ is small, demonstrating the feasibility of the modeling in section 3. And $\overline{|\theta_a|}$ is about 19° when W_{fs} is 10 cm, it is acceptable for people and will become smaller with the smaller W_{fs}. $\overline{|CoP_x|}$ is about 9cm and is in the stable range (6.5-14.5 cm).

To further illustrate effectiveness of the proposed gait, we conducted a comparative experiment of joint training trajectories of human lower limbs. The subject maintained HES balance by changing the center of gravity of his upper body to the side of the support leg and by crutches respectively. The similar gait control curve is used for walking training in the two tests, and the actual movement of the lower limb joints is measured by the inertial measurement units. The relevant experimental results are shown in Figures 16.

In Figure 16(a), with the safety ensured, we remove the use of crutches. We can see that the tester can achieve a good training effect by tilting his upper body to the support leg side, and the actual length of each step is reasonable. Figure 16(c) also shows that the actual movement curve of the lower limbs joints of the human body is similar to the gait control curve. In Figure 16(b), the tester maintained balance by the use of crutches. From C-1 to C-3, the crutches have a good effect, and the actual length of the step is reasonable. However, from C-4 to C-7, the effect of the crutches is not so good enough that the length of the step was too short. This is due to the decrease in the actual height of the right hip of HES caused by gravity (which can be seen in C-5), which leads to the landing in advance of the HES (which can be seen C-6), and this phenomenon can also be displayed by Figure 17(d). In Figure 17(d), during the first right leg swing period, the actual movement curve of the lower limb joints of the human body is similar to the gait control curve. However, in the following two swings periods, the balance effect of the crutches is not so good that the angle of the actual hip joint is much smaller than expected.

8. Discussions

There are at least three real-time and predictable factors causing the instability of HES, and the factor of lateral tilt caused by gravity should be focused in the balance control of frontal plane as showed in Section 2. The proposed rehabilitation gait based on the transfer of gravity center aims to consider the wearer's effort to improve the balance of exoskeleton rehabilitation training of the hemiplegic patients in the frontal plane, reducing the use of crutches/walking frames. Our preliminary results show the promise of the proposed rehabilitation gait to improve lateral stability of HES.

There are certain differences between the proposed gait and the conventional gait, which may cause some doubts. The following points may help to make it clear:

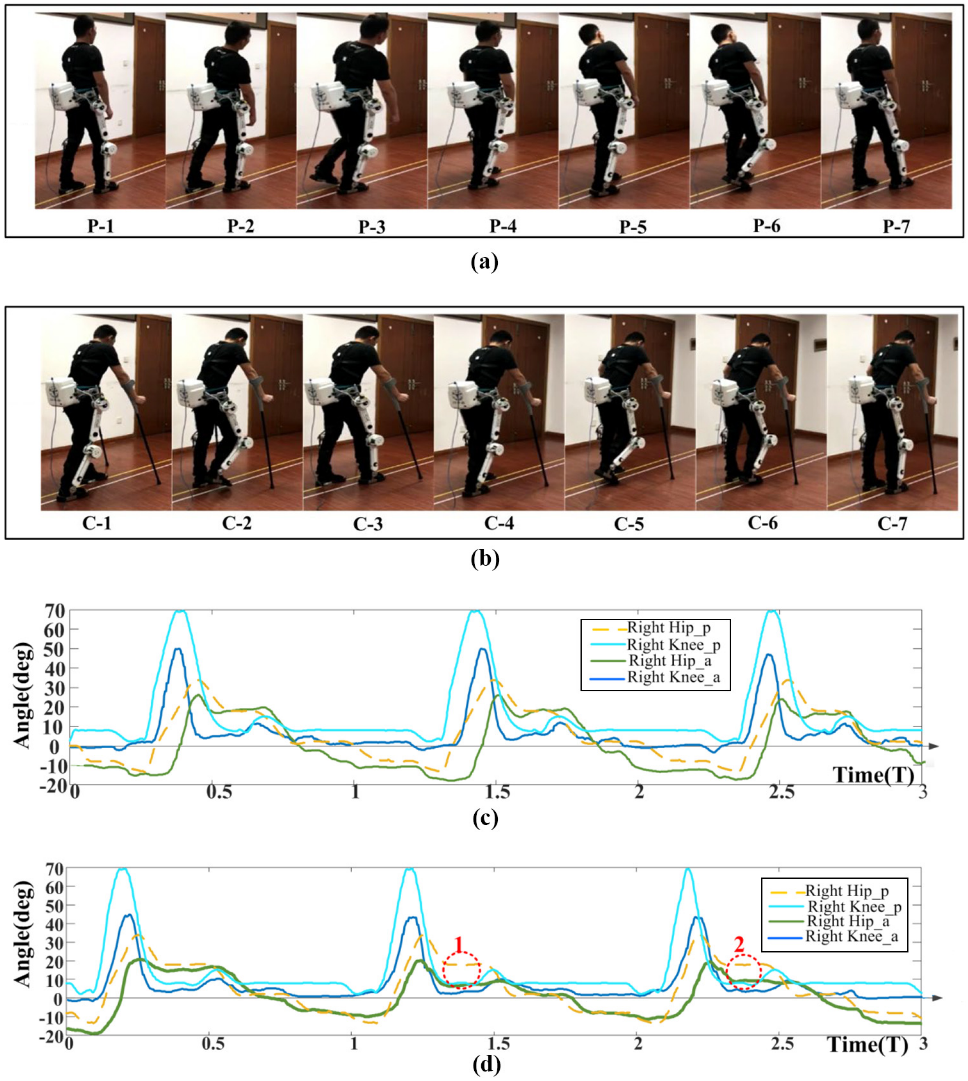
- Person can master the proposed gait after training because of the strong adaptability of human body. In Section 3, it spent average 30 min for a healthy volunteer to master the proposed gait, and the patient can adapt to this gait after short training.
- Figure 8 demonstrated that the difference between the proposed gait and the conventional gait is not so big that people can accept it. In addition, this proposed gait comes from the phenomenon that the blind tilts his upper body to the side of support leg to keep balance more easily when he uses swing leg to detect ground conditions without crutch.
- Most of hemiplegic patients have a certain ability to tilt upper body, although some patients cannot tilt their upper body enough to achieve the optimal balance, the instability problem caused by the gravity effect can be alleviated to a certain degree.
- With the game guidance system, patient can accept to tilt his upper body.
- Several related consultant doctors also pointed out that it is helpful for hemiplegic patients' recovery to do reasonable upper body posture training.
- The proposed gait tries to keep HES in a static-balance-like state at all times to prevent the patient from falling, thereby reducing the dependence on crutches, which can improves feasibility of the rehabilitation training for hemiplegic patients with exoskeleton.
- The gait is only used in the transitional stage of rehabilitation, and we will change the rehabilitation gait when the patient gets a certain recovery.

However, our balance method with the proposed rehabilitation gait still has limitations. First, it may not achieve the ideal balance effect due to patients' limited ability to tilt the upper body. Second, a loss of balance can be induced by a number of perturbation events, such as slipping, tripping, and stumbling [9]. The balance method presented in the paper is not sensitive to these perturbations and does not consider the balance of HES in the in the sagittal plane.

9. Conclusion

This paper analyzed the real-time and predictable causes of the instability of HES and proposed the rehabilitation gait based on the transfer of gravity center. There are three main causes of HES instability without considering the unpredictable perturbation effects: the lateral tilt caused by gravity, human-machine coordination, and the collision caused by the exoskeleton landing in advance. To overcome the lateral tilt caused by gravity, the rehabilitation gait based on the transfer of gravity center was present, and its analysis and related gait information test experiment demonstrate its acceptability to the person. The game guidance system can

Figure 16 Results of the balance method with the proposed rehabilitation gait and with crutches



Notes: (a) Walking process with the proposed rehabilitation gait; (b) walking process with crutches; (c) Joint movement with the proposed rehabilitation gait; (d) joint movement with crutches. Right Hip_P and Right Knee_p mean the preset control value of right hip and right knee respectively. Right Hip_a and Right Knee_a mean the actually measured value of right hip and right knee respectively

make the patient get more naturally used to this gait training. The proposed gait was applied to the prototype exoskeleton ZJULEEX, and the results of the experiments demonstrate the promising potential of the rehabilitation for the balance of HES in the frontal plane with reduce the use of crutches/walking frames during upright locomotion. With the current balance control of the rehabilitation gait, stable walking without crutches has thus far been achieved only with healthy subject. Further research and development of balance control are needed to make exoskeleton useful for clinical populations. Moreover, further research will focus on balance control in response to other types of perturbations to further improve the balance of HES.

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