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Reference Joint Trajectories Generation of CUHK-EXO Exoskeleton for System Balance in Walking Assistance

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ABSTRACT This paper introduces a wearable exoskeleton suit CUHK-EXO that can help paraplegic patients regain their mobility to stand up, sit down, and walk. An offline design and online modification (ODOM) algorithm is proposed for the exoskeleton to generate reference joint trajectories during walking assistance. First, reference trajectories of CUHK-EXO are designed offline based on motion capture data considering leg geometry constraints. Then the human–exoskeleton system (HES) including a pair of crutches is modeled as an eight-link system for analysis. Since the relative position of system center of pressure (COP) is an important factor to indicate system balance during walking, it is estimated in real-time and monitored for exoskeleton control. Based on the system COP position, this paper further proposes the reference trajectories online modification method for CUHK-EXO to counteract disturbances applied to the HES, and hence stabilize system balance in the walking assistance. Finally, walking tests are performed in both healthy subjects and a paraplegic patient to validate the effectiveness of the proposed ODOM algorithm. Testing results demonstrate that knowing the COP desired areas of the wearer, the exoskeleton CUHK-EXO can counteract perturbations and decrease the wearer's efforts, so as to maintain system balance with the ODOM algorithm.

INDEX TERMS Balance control, lower extremity exoskeleton, online modification, trajectory generation, walking assistance.

I. INTRODUCTION

The number of patients with lower limb impairment caused by a stroke, spinal cord injury, or other related diseases is increasing [1]. There are about 0.25 to 0.5 million people who suffer a spinal cord injury every year all over the world [2]. People suffering from paralysis are at an increased risk of secondary medical consequences, such as osteoporosis, muscle atrophy, obesity, and pressure ulcers [1]. In addition, many patients are young people who need to work to support daily life [3]. All those have imposed a heavy and long-term financial burden on both families and the society. Currently, state-of-the-art technologies are utilized to develop robotic assistive devices [4]–[6], such as lower extremity exoskeletons (LEEs), to help these people regain mobility, and hence improve their quality of life. LEEs are wearable robotic systems that are equipped with actuators, and they integrate both human intelligence and robot power. LEEs can apply external force/torque to the wearers' lower limbs under control, and hence provide motion assistance for the wearers according to their motion intentions. By far, enormous progress has been made in the development of LEEs, and some LEEs have been developed and tested, such as ReWalk [7], HAL [8], and the walking supporting exoskeleton [9].

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The most prevalent desires of paraplegic patients are to regain the ability to stand and walk [10]. With the motion assistance provided by LEEs, paraplegic patients are able to perform some essential activities of daily living, such as stand up/sit down, walking, walking upstairs/downstairs, and walking upslope/downslope. Among these motions, walking on level ground is the most important and frequent one. In order to provide effective walking assistance, joint trajectory generation of LEEs has been studied. In the field of bipedal walking, many researchers have investigated on trajectory generation, such as in the humanoid robots [11], [12] and in LEEs developed for gait rehabilitation [13], [14].

In the field of LEEs that are developed to provide motion assistance for the patients who are completely disabled in their lower limbs, some trajectory generation algorithms were proposed as well. Huang *et al.* [15] proposed the trajectory generation algorithm that the exoskeleton knee joint reference trajectory was obtained from that of the hip joint based on the established kinematic model. Tsukahara *et al.* [16] proposed the trajectory generation algorithm in which the reference trajectory of the exoskeleton hip joint was obtained based on other joints angles and the deviation of the center of the ground reaction forces (GRFs) applied to the feet with a kinematic model. In the research of Wang *et al.* [3], reference trajectories of the exoskeleton hip joints in the abduction/adduction direction were generated considering the human-exoskeleton system (HES) lateral stability. Li *et al.* [17] proposed a trajectory generation algorithm that made the exoskeleton zero-moment point trajectory located in the support polygon by establishing the relationship between the zero-moment point trajectory and joint angles, and the system stability was considered.

Most of the currently developed LEEs [7]–[9] can only provide motion assistance for paraplegic patients in the sagittal plane, and a pair of crutches are needed to help the wearer keep balance. Therefore, the crutches are important parts in the whole exoskeleton system. The GRFs applied to crutches can be used to evaluate the supporting forces from the wearer's arms, thus they are related to the wearer's efforts and can indicate the wearer's comfort in the motion assistance of LEEs. Less effort makes the wearer more comfortable. In addition, the GRFs applied to crutches are essential to calculate the system center of pressure (COP) position. The COP position is an important factor to evaluate system balance. However, in the development of LEEs for motion assistance, few researches have considered the GRFs applied to crutches in the trajectory generation for balance control.

The balance of the HES is of important concern in the development of LEEs. In the walking assistance, the HES COP position, which can be estimated by the GRFs applied to the feet and crutches, is a key factor to indicate the system balance. Thus, reference joint trajectories of LEEs should be real-time adjusted according to the COP position to provide stable walking assistance for paraplegic patients. The purpose of this research is to generate reference joint trajectories for the CUHK-EXO exoskeleton to counteract disturbances applied to the HES and make the system COP located in the desired areas by considering the GRFs applied to crutches, and hence ensure the system balance in the walking assistance.

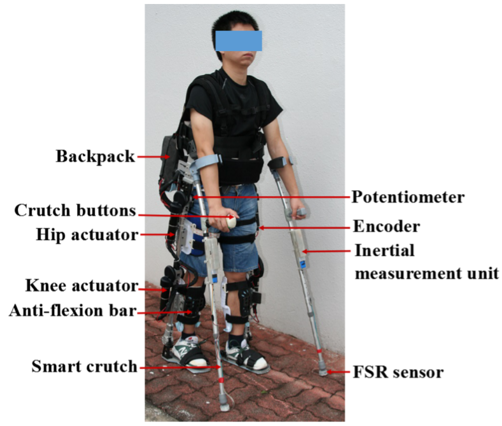

FIGURE 1. The wearable exoskeleton suit CUHK-EXO.

In this paper, a description of CUHK-EXO is firstly presented, followed by the offline design of CUHK-EXO reference trajectories. Then, kinematic model of the HES in the walking motion is established, and COP and center of gravity (COG) of the HES are calculated. Besides, the algorithm of reference trajectories online modification is further proposed. Finally, walking tests are performed to demonstrate the effectiveness of the proposed trajectory generation method.

II. WEARABLE EXOSKELETON SUIT CUHK-EXO

The exoskeleton CUHK-EXO (Fig. 1) is developed to provide motion assistance for paraplegic patients [18]. It weighs about 24 kg (not including the smart crutches) and can accommodate wearers with a range of height from 1.55 m to 1.85 m. It can work continuously for approximate three hours with a fully charged power supply. CUHK-EXO has a total of seven degrees of freedom with three (hip and knee flexion/ extension, ankle dorsiflexion/ plantarflexion) for the right leg and four (hip and knee flexion/ extension, hip external rotation, and dorsiflexion/ plantarflexion) for the left leg. The degree of freedom in the hip external rotation is designed to allow the wearer to transfer into the exoskeleton easily, and is locked after the wearer has put the exoskeleton on. The active joints (hip and knee flexion/extension) of CUHK-EXO are actuated by DC motors through planetary gearboxes and bevel gears. The ankle joints are passive. Anti-flexion bar is designed to prevent the wearer's knee joints from going forward in the standing posture.

A multi-sensor system is built for CUHK-EXO, including the encoders, potentiometers, inertial measurement units (IMUs), and force sensing resistor (FSR) sensors to measure the motion data of the HES. The encoders and potentiometers are placed at the hip and knee joints of the exoskeleton to measure the joint angles and angular velocities; the IMUs are placed at the crutches and the exoskeleton trunk and shanks to obtain the orientation information; and the FSR sensors are placed at the exoskeleton feet and the bottoms of the crutches to acquire the GRF information.

A smart phone App [18] and a pair of smart crutches (Fig. 1) are developed as part of the human-machine interface. The smart phone App is developed with several interfaces: two Login interfaces that are for the patients and therapists to log in, respectively; one Operation interface that is for the therapist to operate the exoskeleton; and one Motion Monitoring interface that is for the therapist to monitor the performance of the exoskeleton such as joint angles and torques in real-time. The smart crutches are designed with FSR sensors, IMUs, Bluetooth modules, microcontroller units, batteries, and several buttons. The buttons are designed on the handle of the right smart crutch to control the power on/power off of the exoskeleton and select the motion types. The crutch has a universal joint, so that the crutch bottom can fully contact the ground. With the smart crutches, more information such as the GRFs applied to crutches and orientation information of crutches can be obtained for the intelligent control of the exoskeleton. The developed smart phone App and smart crutches can improve the system intelligence and make the exoskeleton more easily used by the physical therapists and paraplegic patients.

Three operation modes are developed for CUHK-EXO corresponding to different training stages of paraplegic patients. In the initial stage, since the patient is not familiar with the exoskeleton, the therapist will operate CUHK-EXO through the smart phone App. In the second stage, when the patient gets familiar with the exoskeleton, he/she can operate it through the buttons that are designed on the right smart crutch. In the final stage, when the patient is very familiar with CUHK-EXO, he/she can operate the exoskeleton based on his/her motion intention that can be automatically detected by the exoskeleton through the multiple sensor information. The operation mode through the wearer's motion intention will be done in the future.

III. DESIGN OF REFERENCE JOINT TRAJECTORIES

A. WALKING PATTERN DESIGNED FOR PARAPLEGIC PATIENTS

Generally, human gait and other daily locomotion are three dimensional and can be described in three primary planes of human body [19], including the frontal plane, transversal plane, and sagittal plane. Among these planes, the sagittal plane that describes the person's forward and backward motions is the most important one because most of our movements, the largest torques and powers take place in this plane. A human gait cycle is basically comprised of alternating stance phase and swing phase [20]. It can also be described as two single-step cycles, and a single-step cycle is further divided into the single-leg stance phase (SLSP) and double-leg stance phase (DLSP) [21].

In this research, CUHK-EXO can provide motion assistance for paraplegic patients in the sagittal plane, and a pair of smart crutches are developed to help the wearers to keep balance. Due to the use of crutches, the walking pattern designed for paraplegic patients is different from that of healthy people, as shown in Fig. 2. Here, the gait cycle begins when one leg starts to swing, and ends when the same leg starts to swing again. There are at least three supporting points for the HES to keep balance in each phase. During the SLSP, one leg of the HES is in the swing phase, and the other leg is in the stance phase. Both of the crutches are on the ground. During the DLSP, both legs are in the stance phase, and the crutches are moved forward one after another by the wearer. During this phase, the wearer will also try to transfer his/her COG from the rear leg to the front leg with the help of the upper body strength.

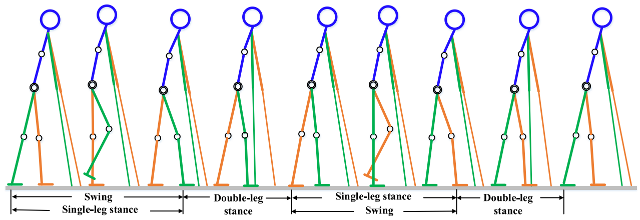

FIGURE 2. Gait phases in the walking pattern for paraplegic patients.

B. CAPTURE OF MOTION DATA

The reference trajectories of CUHK-EXO are obtained from model-based modification of the motion capture data. A preliminary test was conducted with a healthy subject following the walking pattern designed for paraplegic patients introduced above. An optical motion capture system (Vicon Motion Systems Ltd, Oxford, United Kingdom) that is synchronized with a force platform (AMTI, Watertown, USA) is used, as shown in Fig. 3. In the testing, the subject wore CUHK-EXO with the power off, and 16 reflective markers attached at anatomical landmarks of the subject's lower body. The subject was initially in the standing posture with a pair of crutches. Then he performed the walking motion in the walking pattern for paraplegic patients. Three dimensional kinematic data of the subject's lower body are collected, and trajectories of the hip and knee joints in the sagittal plane are obtained.

The captured trajectories were implemented in CUHK-EXO, and walking assistance experiments were conducted with healthy subjects and the exoskeleton power on. In the experiment, the subjects tried not to use their lower limbs strength. However, the subjects could not walk smoothly due to the unmatched joint functions between healthy people's lower limbs and the designed exoskeleton. For example, in the preliminary test, the subject has powered ankle joints, which is helpful to ensure sufficient toe clearance and propel the body forward. Whereas in the following validation experiment, the subjects try to simulate paraplegic patients, thus there is no power in their ankle joints. Therefore, further modification for the captured hip and knee joint trajectories is performed to minimize the differences and enable effective

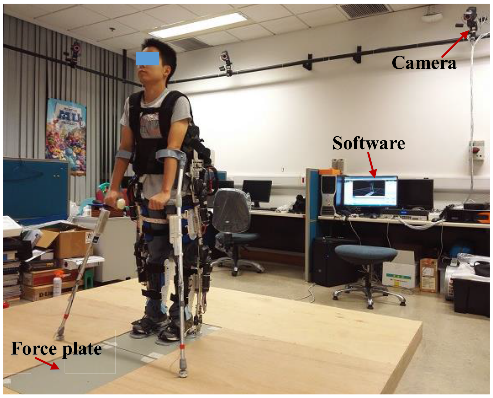

FIGURE 3. Optical motion capture system.

walking. The modification is based on a simple kinematic model of the HES lower limbs, and in the kinematic model, the HES trunk is upright.

C. MOTION CAPTURE DATA MODIFICATION IN SLSP

In the SLSP, the stance leg is responsible for supporting the body weight, and moving the body forward with a desired step-length. The knee joint angle is set to 0° during this phase, thus the hip joints can be in a higher position for better body weight support and leg swing. For the swing leg, the desired motion in this phase is to lift the foot from the ground, move the leg forward, and finally make the foot contact with the ground. Trajectories of the swing leg have a significant effect on the overall performance of the gait. In this research, the SLSP for the swing leg is divided into two stages: knee flexion stage and knee extension stage.

The knee flexion stage is responsible for lifting the foot from the ground with a suitable height for foot clearance, and making it easier to swing the foot forward. In this research, the knee joint of the swing leg flexes to the peak angle at about 50% of the swing phase. At this time, the toe of the swing leg will pass the hip joint with a clearance height of 0.1 m. The motion range of the CUHK-EXO ankle joint is from 0° to 15° in dorsiflexion, and the plantarflexion motion is limited. When the foot leaves the ground, it will drop down because of the foot weight. Thus, the joint angle of the swing leg ankle joint is 0° during the swing phase. For a subject with a height of 1.70 m (thigh length: 0.43 m, shank length: 0.41 m, and ankle height: 0.10 m), the peak joint angles during this stage can be obtained according to the leg geometry constrains as shown in Fig. 4(a). The peak joint angles are 38° for the hip joint and 89° for the knee joint. In this research, the flexion direction of each joint angle is positive, and the joint angle is 0° when the HES is in the upright posture.

In the knee extension stage, the knee joint of the swing leg will extend to move the foot forward. At the end of this stage, the swing foot will be parallel to the ground approximately and contact with the ground with a full foot plate. Similarly, for the same subject, we obtain the hip and knee joint angles at the end of this phase as shown in Fig. 4(b). They are 15°

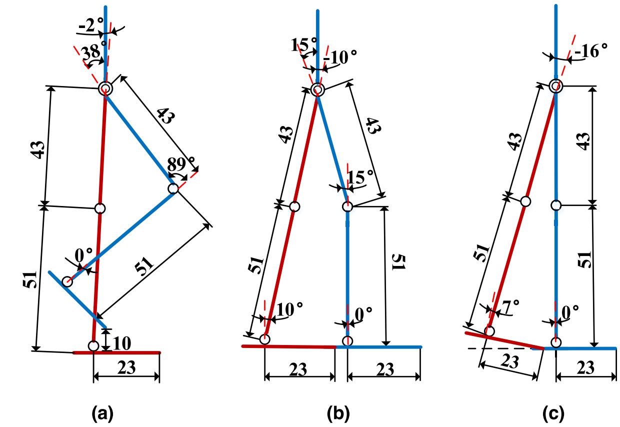

FIGURE 4. Geometrical relationship of the human-exoskeleton system (HES) lower body at different timing. (a) End of the knee flexion stage. (b) End of the knee extension stage. (c) End of the double-leg stance phase.

for the hip joint and 15° for the knee joint. As for the knee joint, the bending angle also makes the weight transfer easier for paraplegic patients.

D. MOTION CAPTURE DATA MODIFICATION IN DLSP

During the DLSP, the HES needs to prepare to get into the next SLSP. The desired motion of the exoskeleton is to transfer the COG forward. Both the hip and knee joints of the front leg will extend. At the end of the DLSP, the heel of the rear leg will leave the ground, which makes it easier to lift the foot from the ground. At this time, joint angles of the exoskeleton can also be obtained with the same subject and leg geometry constrains as shown in Fig. 4(c). They are 0° for both the hip and knee joints of the front leg, and for the rear leg, they are −16° for the hip joint and 0° for the knee joint.

With the gait data collected from motion capture and designated joint angles at special timing considering leg geometry constrains, a sinusoidal function is used to fit the data. Finally, reference trajectories of the exoskeleton hip and knee joints in a complete gait cycle are designed as shown in Fig. 5. In the first single-step cycle (1% - 50% in Fig. 5), the knee flexion stage for the right leg is from 1% to 12.5%, and the knee extension stage for the right leg is from 12.5% to 25%. The DLSP is from 25% to 50%.

Different from the joint power and torque that are related to an individual's body weight, the joint angles of human lower limbs are similar for people with different body weight and dimension [22]. Thus, the designed reference joint trajectories (Fig. 5) were applied during walking assistance. However, the gait period of the designed reference trajectories will be offline modified according to the physical conditions and training stages of the different wearers. In the initial training stage when the wearers are not familiar with the exoskeleton, they cannot transfer their COG quickly. Thus the gait period will be longer. When the wearers get familiar with the exoskeleton and can transfer the COG quickly, the gait period will be shorter.

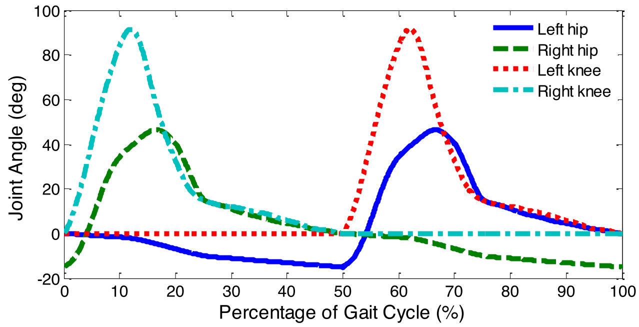

FIGURE 5. **Reference trajectories designed for the exoskeleton hip and knee joints.**

IV. KINEMATICS MODELING OF WALKING MOTION

Kinematic model of the HES during walking motion is established in this section. Since COP and COG positions of the HES are important to indicate system balance [23] in the walking assistance, they are also calculated based on the established kinematic model.

A. KINEMATIC MODEL OF HES

Human body is composed of bones that are linked by various kinds of joints. Assumption that bones behave as rigid segments is made to analyze human motions, thus human body can be divided into several links. For the HES, motions mainly occur in the sagittal plane, thus all lower limb joints only have one degree of freedom in the sagittal plane. The following assumptions are made before modeling the HES during walking: 1) distance between the waist and hip joints in the vertical direction is ignored; 2) left and right hip joints are at the same height during walking; that is the two hip joints are in the same position in the sagittal plane; 3) foot weight of the stance leg is ignored, and the ankle joint is on the ground; 4) the bottoms of crutches are approximatively in the same position in the sagittal plane after the wearer's swing; 5) weight of the links that are composed of the arms and crutches is ignored.

Since the crutches are essential in the walking assistance, they are also included in the modeling of the HES. In the HES kinematic model, the crutch and the wearer's arm that holds it are regarded as one link. The wearer's trunk is divided into two parts, and they are assumed as two links. The trunk lower part is covered with braces and belts of the exoskeleton, and it is controlled by the exoskeleton to follow its movements. The trunk upper part, including the head and neck, is controlled by the wearer himself/herself. For the HES lower body, two links are assumed for each leg, one for the shank and one for the thigh. Therefore, the HES system is modeled as an eight-link system during the walking motion as shown in Fig. 6. In the figure, the left leg is the front leg.

The basic model parameters are defined as follows. m_i is the mass of link i ($i = 1, 2, \ldots, 6$); l_i is the length of link i; and d_i is the length from the center of mass (COM) of link i to joint i. Link 3 is covered with braces and belts, and l_3 is 0.3 m. l_{41} is the distance between the bottom of the trunk upper part

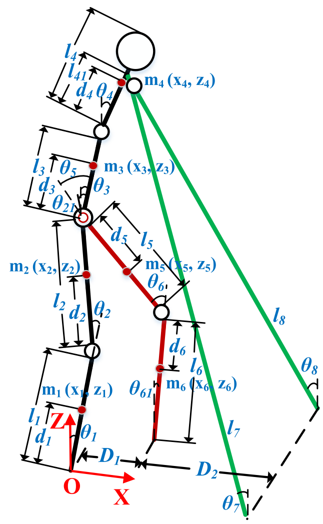

FIGURE 6. **Eight-link model of the HES.**

and shoulder joint. l_7 is the distance between the shoulder joint and bottom of the crutch while the arm is straight. θ_2 and θ_6 are the relative joint angles of the rear and front knee joints with respect to the adjacent segments, respectively; θ_{21} and θ_5 are the relative joint angles of the rear and front hip joints with respect to the adjacent segments, respectively. These joint angles are measured by potentiometers. θ_1, θ_{61}, θ_3, θ_4, θ_7, and θ_8 are the angles of the rear shank, front shank, trunk lower part, trunk upper part, right crutch, and left crutch with respect to the vertical direction, respectively. The clockwise direction is positive for these angles, and they are measured by IMUs. For the sake of simplicity, θ_7 and θ_8 are assumed to be equal in this research. *O-XZ* is the coordinate system in the sagittal plane. The original point O is located at the ankle joint of the stance leg if the HES is in the SLSP, and located at the rear leg ankle joint if the HES is in the DLSP. The X-axis is along the ground, and the Z-axis is vertical to the ground upwards. (x_i, z_i) is the COM position of link i.

In Fig. 6, D_1 is the step-length, and D_2 is the distance between the ankle joint of the front leg and the bottoms of the crutches in the X direction. At the beginning of the DLSP, D_1 can be calculated based on the kinematic model with (1). In addition, D_2 can be calculated with (2) when one of the crutches has been swung forward, and it varies during every single-step cycle.

$$\begin{aligned} D_1 &= l_1 \left(sin\theta_1 - sin\theta_{61}\right) \\ &\quad + l_2(\sin\left(\theta_5 - \theta_3\right) - \sin\left(\theta_{21} - \theta_3\right)) \end{aligned} \tag{1}$$

$$\begin{aligned} D_2 &= l_6 sin\theta_{61} - l_5 \sin\left(\theta_5 - \theta_3\right) \\ &\quad + l_3 sin\theta_3 + l_{41} sin\theta_4 + l_7 sin\theta_7 \end{aligned} \tag{2}$$

B. COP POSITION ESTIMATION IN WALKING

GRFs are important information for analysis and evaluation of human motions since they are associated with human gait and can indicate foot-ground contact conditions [24]. Generally, GRF has two components, including the horizontal

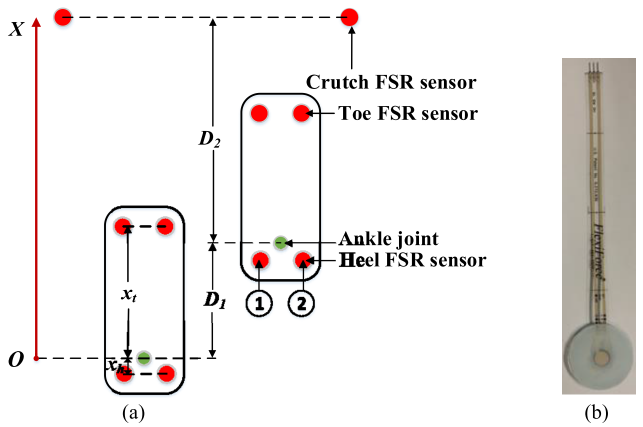

FIGURE 7. The force sensing resistor (FSR) sensors. (a) Locations of the FSR sensors. The red circles stand for the FSR sensors, and the green circles stand for the ankle joints. (b) The FSR sensor implemented inside a silicone pad.

force and vertical force. The horizontal force is parallel to the ground, like the frictional force when a person is walking, and the vertical force is the supporting force perpendicular to the ground. Vertical GRFs can be used to indicate the foot-ground contact conditions and estimate the system COP position. In this research, we only consider the vertical GRFs. In the multi-sensor system of CUHK-EXO, four FSR sensors are mounted on each sole of the exoskeleton, and one FSR sensor is mounted at the bottom of each crutch, as shown in Fig. 7(a). The FSR sensors have a measuring range from 0 to 445 N, and each FSR sensor is implemented inside a silicone pad (Fig. 7(b)). With the force platform (AMTI, Watertown, USA), each FSR sensor has been calibrated to obtain the relationship between the input force and output voltage (which is a function of the output resistance of the FSR sensor).

During different phases of one gait cycle, the HES has different foot-ground and crutch-ground contact conditions, and different foot-crutch relative positions, which have significant effects on the estimation of system COP position. In this research, the COP position estimation is performed in three stages, as shown in Fig. 8. Stage 1 is the SLSP, stage 2 is the DLSP before the crutches are brought forward including the swing forward, and stage 3 is the DLSP when the wearer has swung the crutches forward. In stage 1, the HES has three supporting points (Fig. 8), i.e., one foot and two crutches, and the coordinate system is located at the ankle joint of the stance leg. Then, the COP position of the stance foot can be obtained as follows:

$$x_{sf} = \frac{(f_{sh1} + f_{sh2})x_h + (f_{st1} + f_{st2})x_t}{f_{sh1} + f_{sh2} + f_{st1} + f_{st2}} \tag{3}$$

where $\mathrm{f_{shj}}f_{shj}$ ($j = 1, 2$) and $\mathrm{f_{stj}}f_{stj}$ are the GRFs measured by the FSR sensors mounted at the heel area and toe area of the stance foot, respectively, x_t and $\mathrm{x_h}x_h$ are the toe and heel FSR sensors positions, respectively.

Thus, the COP position of HES during stage 1 can be obtained as follows:

$$x_{cop_1} = \frac{f_{sf}x_{sf} + (f_{c1} + f_{c2})D_{21}}{f_{sf} + f_{c1} + f_{c2}} \tag{4}$$

where f_{sf} is the resultant GRF applied to the stance foot, D_{21} (Fig. 8) is the distance $D_2\mathrm{D}_2$ calculated in the last single-step cycle, and f_{c1} and f_{c2} are the GRFs applied to crutches.

In stage 2 and stage 3, the HES has four supporting points, i.e., two feet and two crutches (Fig. 8), and the coordinate system is located at the ankle joint of the rear leg. Then, the COP positions of the front and rear feet can be obtained as follows:

$$x_{ff} = \frac{(f_{fh1} + f_{fh2})x_h + (f_{ft1} + f_{ft2})x_t}{f_{fh1} + f_{fh2} + f_{ft1} + f_{ft2}} + D_1 \tag{5}$$

$$x_{rf} = \frac{(f_{rh1} + f_{rh2})x_h + (f_{rt1} + f_{rt2})x_t}{f_{rh1} + f_{rh2} + f_{rt1} + f_{rt2}} \tag{6}$$

where f_{fhj} and f_{ftj} are the GRFs measured by the FSR sensors mounted at the heel and toe areas of the front foot, respectively, f_{rhj} and f_{rtj} are GRFs measured by the FSR sensors mounted at the heel and toe areas of the rear foot, respectively.

The foot-crutch relative position D_2D_2 changes in stage 2 since the wearer swings the crutches forward in this stage. Thus the COP position in stage 2 is calculated in different situations with (7). In situation 1, no crutches have been brought forward. In situation 2, the front crutch contacts the ground and the rear crutch does not leave the ground. In situation 3, the front crutch contacts the ground and the rear crutch leaves the ground. The swing conditions of the crutches can be detected by the GRFs applied to the crutches.

$$x_{cop_2} = \begin{cases} \dfrac{f_{ff}x_{ff} + f_{rf}x_{rf} + (f_{c1} + f_{c2})D_{21}}{f_{ff} + f_{rf} + f_{c1} + f_{c2}}, & \text{situation1} \\ \dfrac{f_{ff}x_{ff} + f_{rf}x_{rf} + f_{rc}D_{21} + f_{fc}(D_1 + D_{22})}{f_{ff} + f_{rf} + f_{rc} + f_{fc}}, & \text{situation2} \\ \dfrac{f_{ff}x_{ff} + f_{rf}x_{rf} + (f_{c1} + f_{c2})(D_1 + D_{22})}{f_{ff} + f_{rf} + f_{c1} + f_{c2}}, & \text{situation3} \end{cases} \tag{7}$$

where f_{ff} and f_{rf} are the resultant GRFs applied to the front and rear feet, respectively, f_{fc} and f_{rc} are the GRFs applied to the front and rear crutches, respectively. D_{22} D_{22} (Fig. 8) is the distance D_2 calculated in the current single-step cycle.

As for stage 3 (Fig. 8), the crutches have been swung forward, and the COP position can be obtained as follows:

$$x_{cop_3} = \frac{f_{ff}x_{ff} + f_{rf}x_{rf} + (f_{c1} + f_{c2})(D_1 + D_{22})}{f_{ff} + f_{rf} + f_{c1} + f_{c2}} \tag{8}$$

C. CALCULATION OF COG POSITION

Based on the HES eight-link model (Fig. 6), the COM position of each link (except for link 7 and link 8) can be calculated as follows:

$$\begin{cases} x_1 = d_1 sin\theta_1 \\ z_1 = d_1 cos\theta_1 \end{cases} \tag{9}$$

$$\begin{cases} x_2 = l_1 sin\theta_1 - d_2 \sin(\theta_{21} - \theta_3) \\ z_2 = l_1 cos\theta_1 + d_2 \cos(\theta_{21} - \theta_3) \end{cases} \tag{10}$$

$$\begin{cases} x_3 = l_1 sin\theta_1 - l_2 \sin(\theta_{21} - \theta_3) + d_3 sin\theta_3 \\ z_3 = l_1 cos\theta_1 + l_2 \cos(\theta_{21} - \theta_3) + d_3 cos\theta_3 \end{cases} \tag{11}$$

$$\begin{cases} x_4 = l_1 sin\theta_1 - l_2 \sin(\theta_{21} - \theta_3) \\ +l_3 sin\theta_3 + d_4 sin\theta_4 \\ z_4 = l_1 cos\theta_1 + l_2 \cos(\theta_{21} - \theta_3) \\ +l_3 cos\theta_3 + d_4 cos\theta_4 \end{cases} \tag{12}$$

$$\begin{cases} x_5 = l_1 sin\theta_1 - l_2 \sin(\theta_{21} - \theta_3) \\ \qquad +d_5 sin(\theta_5 - \theta_3) \\ z_5 = l_1 cos\theta_1 + l_2 \cos(\theta_{21} - \theta_3) \\ \qquad -d_5 cos(\theta_5 - \theta_3) \end{cases} \tag{13}$$

$$\begin{cases} x_6 = l_1 sin\theta_1 - l_2 \sin(\theta_{21} - \theta_3) \\ +l_5 sin(\theta_5 - \theta_3) - d_6 sin\theta_{61} \\ z_6 = l_1 cos\theta_1 + l_2 \cos(\theta_{21} - \theta_3) \\ -l_5 cos(\theta_5 - \theta_3) - d_6 cos\theta_{61} \end{cases} \tag{14}$$

Thus, the COG position of the HES in the sagittal plane can be obtained as follows:

$$\begin{cases} x_{COG} = \dfrac{m_1x_1 + m_2x_2 + m_3x_3 + m_4x_4 + m_5x_5 + m_6x_6}{m_1 + m_2 + m_3 + m_4 + m_5 + m_6} \\ z_{COG} = \dfrac{m_1z_1 + m_2z_2 + m_3z_3 + m_4z_4 + m_5z_5 + +m_6z_6}{m_1 + m_2 + m_3 + m_4 + m_5 + m_6} \end{cases} \tag{15}$$

V. REFERENCE TRAJECTORIES ONLINE MODIFICATION

During the walking motion of paraplegic patients with the assistance provided by LEEs, internal or external perturbations may apply to the HES without any prior notice to the wearer and system, which have significant effects on the system balance. The internal perturbations could be unexpected upper body postures such as trunk inclination, and the external perturbations could be unexpected force applied to the HES such as a push on the wearer. The COP of HES is located at the stable region defined by the convex polygon of the supporting points (feet and crutches). If the system COP is near the center of the stable region, the system will be very stable because of large stability margin. On the other hand, if the COP is near the edge of the stable region, the system would easily lose balance. Therefore, relative position of the system COP is important to evaluate the system balance.

Safety is an important factor during the development of robotic systems that include physical human-robot interactions [25]. Based on the kinematics analysis in the previous section, we know that the joint trajectories of the HES determine the system posture, and hence can affect the system COP position. Therefore, in order to counteract disturbances applied to the HES and provide paraplegic patients with stable walking assistance, the offline design and online modification (ODOM) algorithm is developed for CUHK-EXO to generate reference trajectories. The offline designed reference trajectories (Fig. 5) of the exoskeleton will be online modified at each control cycle based on the COP position.

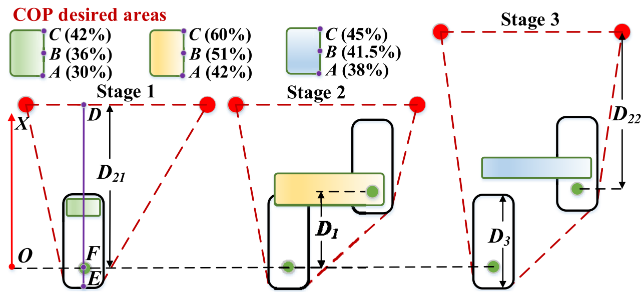

FIGURE 8. The desired areas of the HES center of pressure (COP) in different stages during walking.

A. DESIGN OF COP DESIRED AREAS

As shown in Fig. 8, during the SLSP, the COP of the HES is located in the Triangular Region (TR) defined by three supporting points, and during the DLSP, it is located in the Quadrilateral Region (QR) defined by four supporting points. The relative position of COP varies in TR and QR during walking. Thus we would like to find the desired areas for the HES COP during walking where the system is balanced and the wearer does not feel uncomfortable because large supporting forces from the arms are not required.

A preliminary test was conducted to obtain the COP desired areas in TR and QR during a single-step cycle. From the test, we found that the COP locates at different positions of TR and QR in different stages. In addition, for a specific stage, the COP position varies with different wearers or even the same wearer but with different body postures. Repeated measures ANOVA ($P \leq 0.05$) were performed. We found that the variation of COP position in a specific stage is small, and there is not a statistically significant difference. Therefore, the COP desired areas of the HES in different stages of a single-step cycle are obtained as shown in Fig. 8. The colored rectangular areas are the COP desired areas in different stages where the system is balanced and the wearer does not feel uncomfortable. In this research, the walking assistance occurs in the sagittal plane. Thus we only consider the COP position variation in the sagittal plane, i.e., in the X direction, and hence TR can be represented by the solid purple line ED in stage 1, and the COP desired area can be represented by line AC. In stage 1, point A located at about 30% of line FD is the lower bound of the COP desired area, and point C located at about 42% of line FD is the upper bound. Point B is the COP desired position in stage 1. It is the midpoint of line AC and located at about 36% of line FD. As for stage 2 and stage 3, the definition of the COP desired areas is similar to that of stage 1.

B. TRAJECTORY ONLINE MODIFICATION IN SLSP

In general, if a system is in a quasi-static mode, the ground projection of the system COG will be close to its

COP [26], [27]. The walking motion of HES is not static, but the walking pattern we developed for paraplegic patients has the following features: small step-length and large gait period. Thus the walking motion of the HES can be assumed to be quasi-static, and the change of the HES COG position can significantly affect the system COP position. A feedback control algorithm for the trajectory online modification based on the COP position is proposed, and the general algorithm is given in (16). Here,Δx_{COG} is the COG deviation between the predefined (offline designed) and modified joint trajectories, and Δx_{COP} is the deviation between the actual and desired COP positions. Parameter k_j $(j = 1, 2)$ is set as constant and determined in a preliminary experiment.

$$\Delta x_{COG} = k_j \Delta x_{COP} \tag{16}$$

As for the parameter k_j, k_1 is for the trajectory online modification in stage 1, and k_2 is for stage 2 and stage 3. With (15), the COP deviation Δx_{cop} can be expressed as follows:

$$\begin{aligned} & k_j \Delta x_{COP} \\ & = k_j \left(x_{act_{cop}} - x_{des_{cop}}\right) = \Delta x_{COG} \\ & = \frac{m_1x_{12}+m_2x_{22}+m_3x_{32}+m_4x_{42}+m_5x_{52}+m_6x_{62}}{m_1+m_2+m_3+m_4+m_5+m_6} \\ & \quad - \frac{m_1x_{11}+m_2x_{21}+m_3x_{31}+m_4x_{41}+m_5x_{51}+m_6x_{61}}{m_1+m_2+m_3+m_4+m_5+m_6} \end{aligned} \tag{17}$$

where x_{i1} $(i = 1, 2, \ldots, 6)$ and x_{i2} are the COM positions of each link in the X direction before and after the modification of lower limb joint trajectories, respectively.

For human beings, multiple strategies have been used to keep balance while walking, including the ankle strategy, hip strategy, and strategy of their combination [28]. In this research, the ankle joints of the HES are passive, thus we only modify the reference trajectories of CUHK-EXO hip joints in real-time according to the COP position. In stage 1, reference trajectory of the stance leg hip joint are online modified, and the modification angle φ_1 is defined. Then, (17) can be rearranged as follows:

$$k_1 \Delta x_{cop} = \frac{p[\sin(\alpha + \varphi_1) - sin\alpha]}{q} \tag{18}$$

where the various terms are expressed by (19), (20), and (21):

$$\alpha = \theta_{21} - \theta_3 \tag{19}$$

$$p = -m_2 d_2 - (m_3 + m_4 + m_5 + m_6) l_2 \tag{20}$$

$$q = m_1 + m_2 + m_3 + m_4 + m_5 + m_6 \tag{21}$$

Rearrange (18), we obtain φ_1 as follows:

$$\varphi_1 = \arcsin\left(\frac{q}{p} k_1 \Delta x_{cop} + sin\alpha\right) - \alpha \tag{22}$$

C. TRAJECTORY ONLINE MODIFICATION IN DLSP

In stage 2 and stage 3, reference trajectories of both hip joints are modified, and the modification angles φ_2 and φ_3 are defined for the rear leg and front leg, respectively. Similarly, (17) can be rearranged as follows:

$$k_2 \Delta x_{cop} = \frac{p\,[\sin(\alpha + \varphi_2) - sin\alpha] + r[\sin(\beta + \varphi_3) - sin\beta]}{q} \tag{23}$$

where the various terms are expressed by (24) and (25):

$$\beta = \theta_5 - \theta_3 \tag{24}$$

$$r = m_5 d_5 + m_6 l_5 \tag{25}$$

In this research, the modification angle φ_2 φ_2 is set equal to $\varphi_3\varphi_3$ for the sake of simplicity. Rearrange (23), we obtain:

$$\varphi_2 = \varphi_3 = -\arcsin\left(\frac{qk_2\Delta x_{cop} + psin\alpha + rsin\beta}{\sqrt{p^2 + r^2 + 2prcos\,(\alpha - \beta)}}\right) - \emptyset \tag{26}$$

where

$$\tan\emptyset = \frac{psin\alpha + rsin\beta}{pcos\alpha + rcos\beta} \tag{27}$$

At each control cycle of the walking motion, the COP position is calculated. If the COP is located in the desired areas, the exoskeleton will be controlled to follow the predefined reference trajectories. On the other hand, if the COP gets out of the desired areas, the modification angles φ_1, or φ_2 and φ_3 will be calculated and added to the predefined reference trajectories of the hip joints to make the COP located in the desired areas.

VI. PERFORMANCE EVALUATION

In order to validate the effectiveness of the proposed ODOM algorithm, walking tests were performed. The controller architecture of CUHK-EXO and testing results are presented in the following subsections.

A. CONTROLLER ARCHITECTURE OF CUHK-EXO

The exoskeleton and wearer's lower limbs are well connected through braces, and they have formed a closed loop as a human-exoskeleton cooperative system. Impedance control has been widely used in exoskeletons for rehabilitation application [1], [29], and the exoskeleton can provide the assistance according to the wearer's voluntary participation during the rehabilitation. The target users of CUHK-EXO are paraplegic patients who have lost motor and sensory functions in their lower limbs. Thus, there is no voluntary force/torque from the wearer's lower limbs applied to the exoskeleton. In this research, position control is adopted for CUHK-EXO. As compared with other control strategies, position control has the advantages of reduced hardware complexity, sensor need, and computational complexity. Besides, the disadvantage of tracking fixed references, such as limited capability to counteract external perturbations and keep the system balanced, can also be solved with the proposed ODOM algorithm.

The controller architecture of CUHK-EXO is shown in Fig. 9, which is composed of a high-level controller and a low-level controller. The high-level controller is to recognize

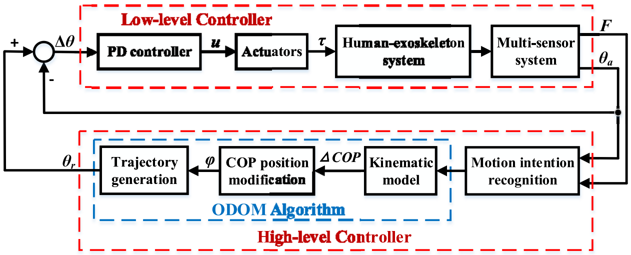

FIGURE 9. CUHK-EXO controller architecture.

the wearer's motion intention, analyze and evaluate his/her motion conditions, and finally generate the references for the exoskeleton. The function of the low-level controller is to collect the feedback sensor data, send them to the high-level controller, and regulate the actuators to output desired motions with proportional-derivative (PD) control. Finally, assistive torques in the HES lower limb joints are generated from the actuators to help the patients perform activities of daily living.

B. EVALUATION OF ODOM ALGORITHM

Preliminary tests in three healthy subjects (age: 25-30, height: 1.70 ± 0.05 m, weight: 62 ± 8 kg) were conducted firstly, and the testing results validated the effectiveness of the ODOM algorithm. Then, walking tests with a paraplegic patient were conducted with disturbances to further evaluate the performance of the ODOM algorithm. The patient was a poliomyelitis patient who had no strength and limited sensory function in his lower limbs. He was 29-year-old (1.65 m, 66 kg) and suffered from the disease when he was 2 years old. In his daily life, he relied on a wheelchair for mobility. The tests were conducted under the clinical ethical approval that was approved by the Joint Chinese University of Hong Kong – New Territories East Cluster Clinical Research Ethics Committee (Ref. No.: 2015.262). Before participating in the tests, the patient was informed of the test protocol and gave his consent.

In this research, the patient used the developed smart crutches for balance, and his motion intention was detected through the buttons on the right crutch. The patient was initially in the sitting posture. After pressing the button corresponding to the sit-to-stand motion, he got into the standing posture with the assistance from CUHK-EXO. Then, he pressed the button corresponding to the walking motion and walked with CUHK-EXO. Two walking tests were conducted with external pushes applied to the patient's trunk in a laboratory environment of The Chinese University of Hong Kong in this research. In the first walking test, the exoskeleton was controlled to track the offline designed reference joint trajectories without the ODOM algorithm; while in the second walking test, the exoskeleton was controlled with the proposed ODOM algorithm.

The high-level controller of CUHK-EXO was implemented on a remote PC, and the main code of the exoskeleton

TABLE 1. The HES parameters in the kinematic model.

Link	Weight m (kg)	Length l (m)	COM Position d (m)
LAC	NA	1.30	NA
RAC	NA	1.30	NA
TUP	22	0.46	0.304
TLP	30	0.3	0.15
Rear thigh	9.0	0.33	0.187
Rear shank	4.19	0.33	0.187
Front thigh	9.0	0.33	0.143
Front shank	4.19	0.33	0.143

Abbreviation: LAC, the link that is composed of the left arm and left crutch; RAC, the link that is composed of the right arm and right crutch; TUP, trunk upper part; TLP, trunk lower part; NA, not applicable.

FIGURE 10. Snapshots of the walking tests.

system was written in the software MATLAB. In the tests, the parameters k_1 and k_2 are set to 0.85 and 0.45, respectively. The sampling frequency of the system is 50 Hz, which is fast enough for human locomotion analysis [30], [31]. The HES weighs about 90 kg with a height of 1.65 m. According to the literature [22], [32], the parameters in the kinematic model (Fig. 6) of the HES are identified as given in Table I.

In the walking tests, the GRFs applied to the patient's feet and crutches were measured to calculate the system COP position in real-time. During the SLSP, the patient held the smart crutches to keep balance, and his leg was swung forward with the assistance from CUHK-EXO. During the DLSP, the patient moved the crutches forward one by one and transferred his COG from the rear leg to the front leg with his upper body strength. Snapshots of the walking tests are shown in Fig. 10. The tests were performed under the supervision of two physical therapists. For safety consideration, one author of this paper stood behind the patient to protect him from potential unexpected falling in the case of system failure during the tests.

Testing results of the two walking tests in one gait cycle are shown in Fig. 11 and Fig. 12. The GRFs applied to crutches and the HES actual COP positions and COP desired areas in the first walking test are shown in Fig. 11(a) and Fig. 11(b), respectively. The GRFs applied to crutches in the second walking tests are shown in Fig. 12(a). The HES actual COP positions, COP desired areas, and COP deviation between the actual and desired COP positions are given in Fig. 12(b). The hip joint modification angles are shown in Fig. 12(c). The offline designed and online generated reference trajectories,

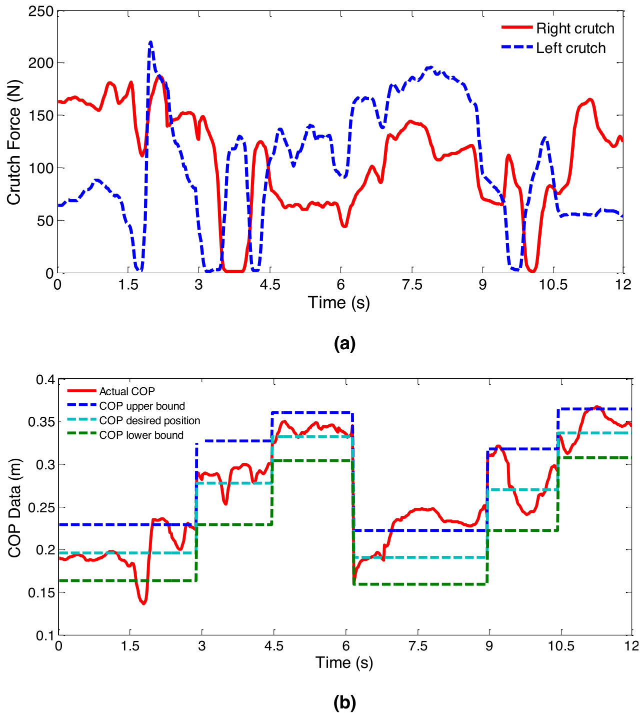

FIGURE 11. **Testing results of the first walking test. (a) The ground reaction forces (GRFs) applied to crutches. (b) The HES COP positions.**

and actual joint angles of the exoskeleton hip joints are shown in Fig. 12(d). In the tests, the step-length is about 0.27 m. The single-step period is 6 s, and the single-leg stance time and double-leg stance time are both about 3 s.

In the first walking test, a push was applied to the patient's chest at about 1.5 s (SLSP) without any notice to the patient to simulate an external disturbance. In this situation, due to additional backward momentum, the GRFs (Fig. 11(a)) applied to crutches became small suddenly, and the COP exceeded the desired areas (Fig. 11(b), 1.6 s – 2.3 s) and got into an area where the system had the possibility of losing balance. Since the exoskeleton was controlled to follow the predefined reference trajectories, the patient had to try his best to maintain balance with his upper body strength. After trying to lean his trunk forward, he also moved the left crutch slightly backwards to keep balance. And the largest GRF applied to the left crutch is about 220 N at this time (Fig. 11(b), 2 s). It took the HES 0.7 s to drive the COP back into the desired areas.

In addition, a push was applied to the patient's back at about 6.9 s (SLSP) in the first walking test. Due to additional forward momentum, the GRFs (Fig. 11(a)) applied to crutches became large suddenly, and the COP also exceeded the desired areas (Fig. 11(b), 7 s – 9 s) and got into an area where the wearer had to apply large force to the crutches to keep balance. It took the HES 2 s to drive the COP back into the desired areas with an average force of 179 N and 123 N applied to the left and right crutches during this period, respectively. During the period of 6 s – 7 s when the COP was located at the desired areas during the SLSP, the calculated

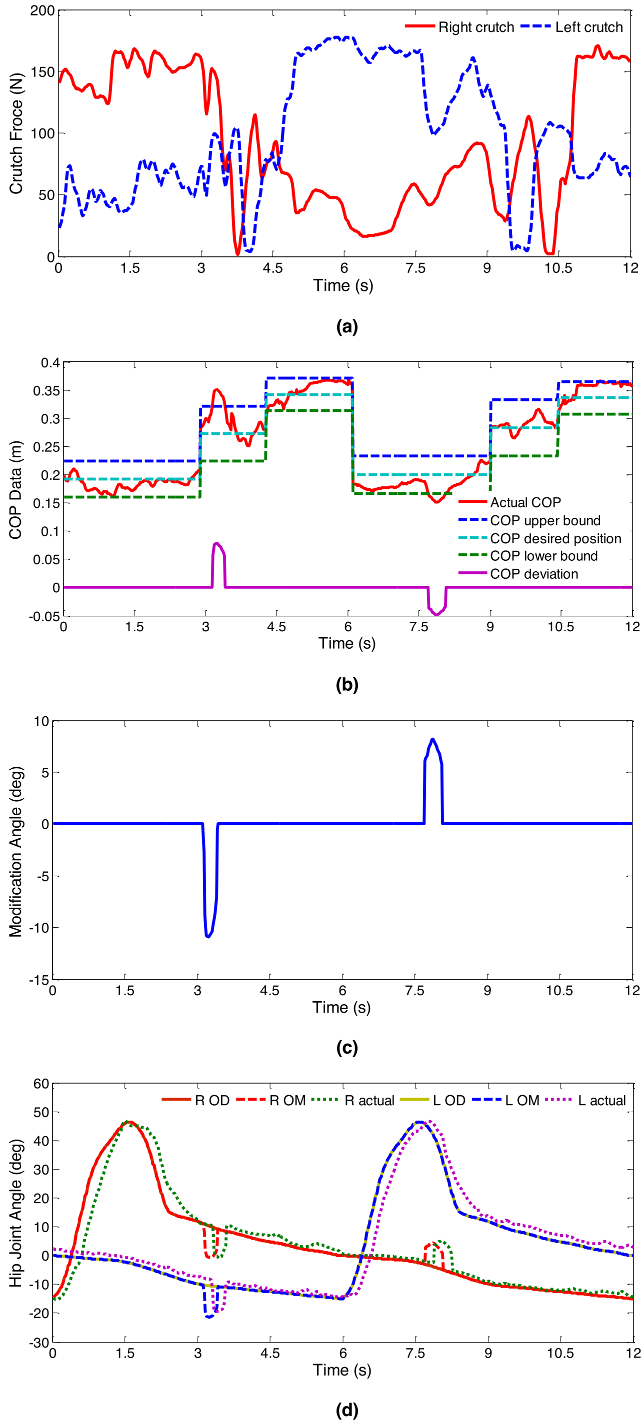

FIGURE 12. **Testing results of the second walking test. (a) The GRFs applied to crutches. (b) The HES COP positions. (c) Online generated modification angles for the hip joints. (d) Reference and actual trajectories of the hip joints. In the figure, L stands for the left, R stands for the right, OD stands for the offline designed references, and OM stands for the online modified references.**

average force applied to the left crutch and right crutch were 155 N and 85 N, respectively.

In the second walking test, a push similar to that in the first walking test was applied to the patient's back at about 3.1 s (DLSP). Due to additional forward momentum, the GRFs

(Fig. 12(a)) applied to crutches became large suddenly, and the COP exceeded the desired areas (Fig. 12(b), 3.2 s – 3.4 s) and got into an area where the wearer had to apply large force to the crutches to keep balance. Thus, the modification angles φ_2 and φ_3 (Fig. 12(c)) were calculated and added to the predefined reference trajectories of hip joints. Then, the reference trajectories were modified (Fig. 12(d)) to drive the COP back into the desired areas (Fig. 12(b)), which stabilized the system balance and decreased the wearer's efforts under disturbances. It took the HES 0.2 s to drive the COP back into the desired areas with an average force of 78 N and 119 N applied to the left and right crutches during this period, respectively. During the period of 9 s – 12 s (DLSP) when the COP was located at the desired areas, the calculated average force applied to the left crutch and right crutch were 74 N and 99 N, respectively.

In addition, at about 7.6 s (SLSP) in the second walking test, a push similar to that in the first walking test was applied to the patient's chest. Due to additional backward momentum, the GRFs (Fig. 12(a)) applied to crutches became small suddenly, and the COP exceeded the desired areas (Fig. 12(b), 7.7 s – 8.1 s) and got into an area where the system has the possibility of losing balance. Thus, the modification angles φ_1 (Fig. 12(c)) were calculated and added to the predefined reference trajectories of hip joints. Then, the reference trajectories were modified (Fig. 12(d)) to drive the COP back into the desired areas (Fig. 12(b)), which ensured the system balance under disturbances. It took the HES 0.4 s to drive the COP back into the desired areas. During the period when the system COP was located at the desired areas, the exoskeleton just followed the offline designed reference trajectories (Fig. 12(d)).

After the two walking tests, the patient was asked about the feeling while walking with and without the ODOM algorithm. The patient reported that with the ODOM algorithm, he can achieve balance more easily with less efforts from his arms under perturbations.

VII. DISCUSSIONS

The most prevalent desires of paraplegic patients are to regain the ability to walk around from their wheelchairs, and hence have independent lives. A number of LEEs have been developed with crutches to provide motion assistance for paraplegic patients in the sagittal plane. However, for LEEs applied to motion assistance, few researches have considered the GRFs applied to the crutches in the trajectory generation for balance control. The goal of this research is to propose the reference trajectories generation algorithm ODOM for CUHK-EXO during the walking assistance. With the ODOM algorithm, reference trajectories of the exoskeleton are designed offline first, and then online modified according to the system COP position, thus disturbances applied to the HES can be counteracted to ensure the system balance.

Walking tests were conducted to validate the effectiveness of the ODOM algorithm. In the walking tests with a paraplegic patient, pushes were applied to the HES without notice to the patient to simulate disturbances. When a push was applied to the patient's back during the SLSP, it took the HES controlled without the ODOM algorithm 2 s to drive the COP back into the desired areas with an average force of 302 N applied to the crutches. The crutches GRFs (302 N) had an increase of 25.8% as compared with that (240 N) when the COP was located in the desired areas during the SLSP. For the exoskeleton controlled with the ODOM algorithm, it only took the HES 0.2 s to drive the COP back into the desired areas with an average force of 197 N applied to the crutches when a similar push was applied to the patient's back during the DLSP. The crutches GRFs (197 N) only had an increase of 13.9% as compared with that (173 N) when the COP was located in the desired areas during the DLSP.

In the situation when a push was applied to the patient's chest during the SLSP, it took the HES 0.7 s to drive the COP back into the desired areas with a peak GRF of 220 N (Fig. 11(b), 2 s) applied to the left crutch. The patient also moved the left crutch slightly backwards to achieve balance during the period of 1.6 s – 2.3 s, which was dangerous since the situation existed when there was only two supporting points for the HES in this period. For the exoskeleton controlled with the ODOM algorithm, when a similar push was applied to the patient's chest during the SLSP, it only took the HES 0.4 s to drive the COP back into the desired areas without moving the crutches.

As compared with the results of the exoskeleton controlled without the ODOM algorithm, it took the HES controlled with the ODOM algorithm less time and less wearer's efforts to drive the COP back into the desired areas in a safe mode. Therefore, with the implementation of the proposed ODOM algorithm, disturbances applied to the HES can be counteracted and the wearer's efforts can be reduced under perturbations, and hence stable walking assistance can be provided for the wearer.

However, there are limitations in the current research. The number of subjects in the clinical tests is small, thus more paraplegic patients will be recruited to further evaluate the performance of the ODOM algorithm in future studies, and the statistical analysis will then be performed. In addition, the capability to counteract disturbances of CUHK-EXO with the proposed ODOM algorithm is limited. If the disturbances are too large and the bottoms of the crutches almost leave the ground in the walking motion, the HES COP would be near the edge of TR or QR. In this situation, the HES would be like an inverted pendulum due to the passive ankle joints. In addition, the calculated modification angle would be very large due to the large COP deviation, and discontinuity in motion would occur. Thus, the HES would be easy to tip over in this situation. It is similar if the heels of the HES leave the ground and the system COP is close to the bottoms of the crutches under large disturbances. Thus the active ankle joints can be combined into the ODOM algorithm to improve the exoskeleton capability of counteracting disturbances [28].

VIII. CONCLUSION

In this paper, the reference trajectory generation algorithm ODOM was proposed for the exoskeleton CUHK-EXO during the walking assistance. The gait data collection and modification, establishment of the HES kinematic model, and hip joint trajectory online modification were presented. The effectiveness of the ODOM algorithm was verified by human trials on both healthy subjects and a paraplegic patient. Testing results demonstrated that CUHK-EXO can counteract disturbances applied to the HES, and hence provide stable walking assistance for the wearer with the ODOM algorithm.

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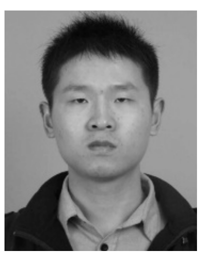

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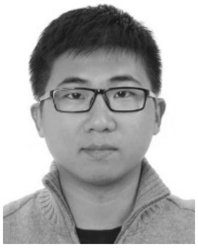

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