Actively Controlled Lateral Gait Assistance in a Lower Limb Exoskeleton*

Letian Wang, Shiqian Wang, Edwin H. F. van Asseldonk, Herman van der Kooij

Abstract— **Various powered wearable lower limb exoskeletons are designed for paraplegics to make them walk again. Control methods are developed and implemented in these exoskeletons to provide active gait assistance in the sagittal plane while active control in the frontal plane is still missing. This paper proposed a control method that provided gait assistance in both lateral and sagittal plane. First, in the lateral plane, the exoskeleton was controlled to support the weight shift during stepping by providing assisting hip ab/adduction torques when the subject initiated a small amount of weight shift to the stance side to trigger a step. Second, the exoskeleton's hip ab/adduction during stepping was controlled to improve lateral stability. This was achieved by altering the amount of hip ab/adduction to change step width at heel strike. Using these controls, an able-bodied subject could walk in the exoskeleton without any external balance aids, i.e. crutches or a walker, where his hip and knee joints were controlled by the exoskeleton and his ankle joints were constrained by the exoskeleton. The next step is to test whether the proposed method improves balance in spinal cord injured subjects.**

I. INTRODUCTION

Loss of mobility is the direct consequence of neurologic injuries such as stroke or spinal cord injuries. Patients with decreased mobility suffer from a great inconvenience in their daily life and a limited participation in social life. Developing devices to allow paraplegic patients to walk again would drastically change their personal and social lives.

A wearable lower limb exoskeleton is an invention that has the potential to let paraplegics regain locomotion capability. A wearable exoskeleton consists of a pair of anthropomorphic artificial legs, of which selected degrees of freedom (DoFs) are usually actuated by electrical motors. Patients can wear the exoskeleton by attaching the exoskeleton to their legs at selected locations. Over the last couple of years, different exoskeletons were developed and evaluated for research purposes such as Mina [1], WPAL [2], Vanderbilt exoskeleton [3,4]. Some others are already introduced to the market like ReWalk [5] and Ekso [6].

All of these exoskeletons only have actuation in the sagittal plane. The lack of actuated DoFs in the frontal plane limits the capability of these exoskeletons to contribute to maintaining lateral stability and to provide active lateral weight shift. Studies show that a lateral displacement of the Center of Mass (CoM) toward the stance leg (lateral weight shift), precedes the initiation of a step [8, 9]. During walking, adapting step width is crucial for lateral stability [10]. Actively assisting lateral motion could potentially improve gait stability, reduce the usage of external balance aids like crutches and walkers, and ultimately allow paraplegic patients to walk with the exoskeleton with free hands so they can use hands to carry something and pick up an object etc.. Therefore we developed an exoskeleton, the MINDWALKER [11] that allows assisting movement in the frontal plane, by incorporating active hip ab/adduction.

For MINDWALKER, bipedal locomotion and the interaction between the exoskeleton and the users has to be controlled. Bipedal locomotion is a mixture of discrete and continuous control problems. The discrete control problem is the transition between events such as starting, stepping, and stopping. This can be modeled and solved by using state machines with users' inputs. From literature, various methods are used to trigger gait events such as push-button interface [1, 5 ,6], trunk motion [5] and position detection of the Center of Mass (CoM) of the user [4,6]. The continuous control problem is the generation of gait trajectory. Literature suggests that gait patterns can either be predefined trajectories based on offline simulations or captured gait data of healthy subjects[1, 3, 5, 6], or online generated [2] based on balance indicators such as Zero Moment Point [7]. The generated reference patterns are generally tracked using position control at powered joints.

In this paper, we will focus on the control of the frontal plane. The control of sagittal plane movements is performed in a similar way as in the existing devices. We present the design and evaluation of a controller that detects the intention of a subject to shift weight and assists the subject in completing the weight shift. Furthermore, we present a controller that detects when lateral stability is threatened and assists the subject in adapting the lateral foot placement to maintain stability.

The content of this paper is organized as follows. The hardware design of the MINDWALKER exoskeleton is briefly described in section II. In section III, the proposed control method and its implementation are elaborated. In section IV, the preliminary test results are presented and discussed. Finally the conclusions and future work are given in section V.
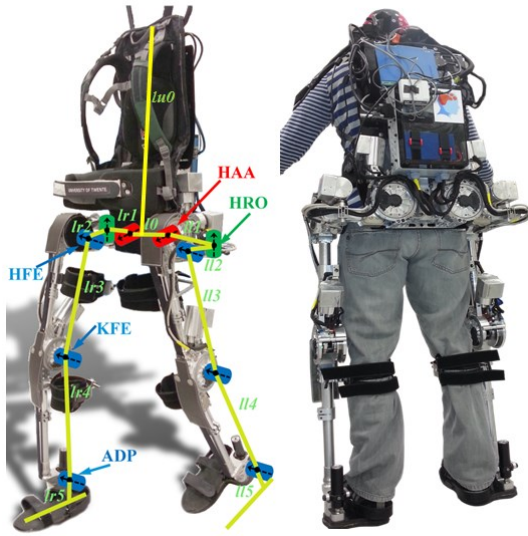
Figure 1. The MINDWALKER exoskeleton. Left: the sketch of DoFs of the MINDWALKER. Different joints (Blue: sagittal. Red: frontal. Green: transverse) and the linkages are indicated and labeled. Right: the back view of a user wearing the exoskeleton.

II. The MINDWALKER Exoskeleton

The MINDWALKER is a powered lower limb exoskeleton designed for paraplegics to regain locomotion capability.

The exoskeleton weighs 28kg excluding batteries and it bears its own weight by transferring the weight via its footplates to the ground. Shown in Fig. 1, the MINDWALKER exoskeleton has five DoFs at each leg, three of which, namely hip ab/adduction (HAA), hip flexion/extension (HFE) and knee flexion/extension (KFE), are powered by series elastic actuators (SEAs); two of which namely, hip endo/exo rotation (HRO) and ankle dorsi/plantar flexion (ADP), are passive but provided with certain stiffness (800Nm/rad at HRO, 180Nm/rad at ADP). The range of motion of all five DoFs are listed in Table I.

The slave electronics are integrated in each joint. They are responsible for communicating with the sensors and motors, and for data preprocessing. At this moment, the control PC and battery for electronics are located in the backpack, the motor battery is off board. Communication between the control PC and slave electronics utilizes EtherCAT network-based architecture.

In each leg, the physical sensing of the MINDWALKER consists of

- Integrated sensors in each motor measuring the motor velocity, temperature.

- Encoders at HAA, HRO, HFE and KFE measuring joint angle positions,

- Encoders at powered joints measuring the deflection (therefore the joint torque) of the series (spiral shaped) springs

- Inertia Measurement Units (IMUs) at the segments of thigh (both lateral bar and sagittal bar) and shank

TABLE I. RANGE OF MOTION AT EACH JOINT

Joint	Degree of Freedom	Range of Motion
Hip	Ab/Adduction (HAA)	17/19°
	Flexion/Extension (HFE)	110/18°
	Endo/exo Rotation (HRO)	10/10°
Knee	Flexion/Extension (KFE)	120/1.5°
Ankle	Dorsi/Plantar Flexion (ADP)	20/20°

measuring the acceleration, velocity and orientation of the corresponding segments in the world frame.

III. Control Method

A. Finite State Machine

Nine states are defined for assisted walking, shown in Fig. 2. The switching conditions for right stepping are described in Table II. The same conditions are symmetrically applied to left stepping.

Two types of transitions are defined, namely, triggered and automatic transitions. Triggered transitions are initiated by the user (either the wearer or the operator). Automatic transitions allow automatic switch from one state to another without user command when certain conditions are fulfilled.
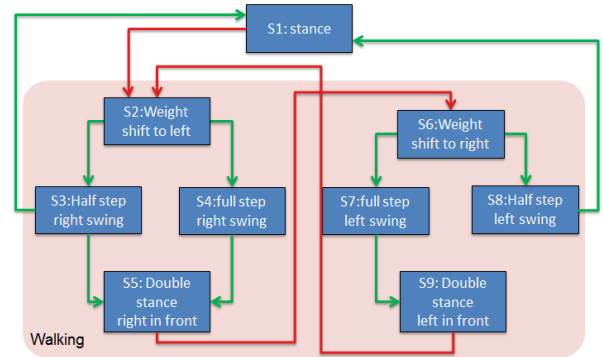

Figure 2. Finite state machine for the stance and gait assistance. Two types of transition among states are represented by colored arrows. Green=Automatic. Red=Triggering by the users.

TABLE II. STATE MACHINE SWITCHING CONDITIONS FOR RIGHT STEPPING

S1 to S2	User inputs the command "start".
S2 to S3	1.The weight is shifted at a desired amount to the left. 2. The previous user input is either "start" or "stop".
S2 to S4	1.The weight is shifted at a desired amount to the left. 2. The previous user input is "next step".
S3 to S5	1.Heel strike. 2. The previous user input is "start".
S3 to S1	1.Heel strike. 2. The previous user input is "stop".
S4 to S5	Heel strike.
S5 to S6	User inputs the command "next step".

$$\varphi_{lu0\_z} = \varphi_{lr3} - \theta_{RHFE}$$

$$\varphi_{lu0\_x} = \varphi_{lr1} - \theta_{RHAA}$$

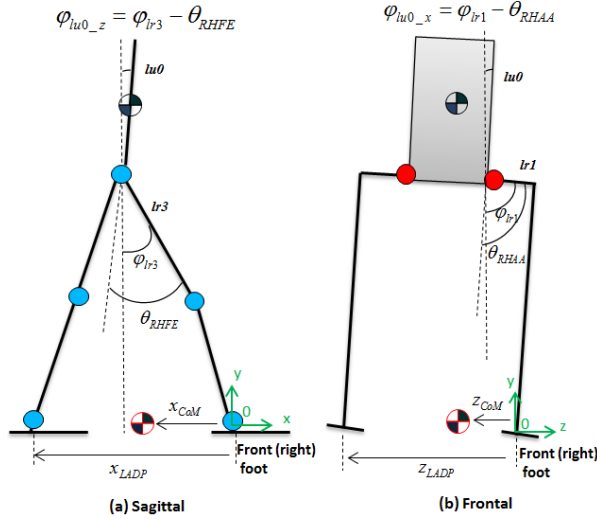**(a) Sagittal**

(b) Frontal

Figure 3. Sketch of estimation of the CoM position in the sagittal plane (a) and in the frontal plane (b). φ denotes the orientation of linkages measured by IMU sensors. θ denotes the joint angels measured by the joint encoders. The 0-axis locates at the front ankle (in this case the right ankle RADP). x and z are the positions of the interested points in sagittal or in frontal. Blue and red circles represent the joints in sagittal and frontal respectively.

For the triggered transitions, two methods are implemented:

1) Trigger by remote control

The user or the system operator can use push-button interface to trigger different operating modes. In this paper we focus on the operating mode of assisted walking. In this mode, start, stop walking and stepping can be triggered using the remote control. Other operating modes like zero-torque, high impedance, sitting, and standing up were implemented and tested but will not be discussed.

2) Trigger by CoM position

State transitions can also be triggered when the user manipulates the CoM position of the user-exoskeleton system by, e.g., leaning forward and sideways. A trigger to initiate a step will be generated when the projection of the sagittal and lateral CoM positions on the ground fall in the desired quadrant. The sagittal and lateral CoM positions are estimated based on sensor data, the geometry and mass property of the exoskeleton and the human anatomical data from [12].

The upper body orientation in sagittal (pitch) and frontal (roll) plane is required in the estimation of the aforementioned CoM positions. As currently there is no IMU above the exoskeleton legs, the upper body orientation is
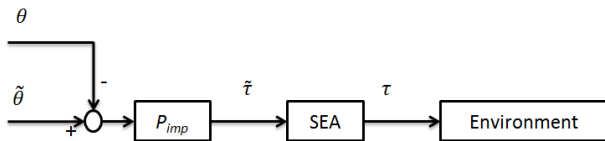
estimated using the IMUs and joint encoders installed in the lower extremities of the exoskeleton. The pitch angle of the upper body φ_{lu0_z} is estimated by the HFE joint position θ_{RHFE} and the orientation of the sagittal bar of the thigh φ_{lr3} shown in Fig. 3a. The roll angle φ_{lu0_x} is estimated in the same manner shown Fig. 3b.

After knowing the estimated CoM positions, we can further calculate the relative CoM positions with respect to the stance foot. Two ratios are defined to quantify the CoM shift between two feet by using the relative CoM position in the sagittal and the frontal plane,

$$r_{\text{sagittal}} = \frac{x_{CoM}}{x_{LADP}} \qquad (1)$$

$$r_{\text{lateral}} = \frac{z_{CoM}}{z_{LADP}}. \qquad (2)$$

r_{sagittal} and r_{lateral} stand for the weight shift ratios in the sagittal and frontal plane. x_{CoM} and x_{LADP} are the CoM position and left ankle position in the sagittal plane. z_{CoM} and z_{LADP} are the positions in the frontal plane. For $r = 0.5$ the weight is equally distributed between two feet and for $r = 0$ and $r = 1$ the weight is shifted to the front foot and rear foot respectively.

Two thresholds are determined empirically for r_{sagittal} and r_{lateral} respectively by taking into account the facts that 1) the trigger should not be too sensitive causing false alarm and 2) the user should not spend too much effort to reach the thresholds. Only if both ratios exceed their thresholds, a trigger to take a step is generated.

B. Joint Control

The motion and the posture defined in the 9 states in Fig. 2 are impedance controlled. Proportional (P) feedback controllers were implemented to track the predefined joint trajectories (in states S2-S4,S6-S8) or to maintain given joint positions (in state S1, S5 and S9). Fig. 4 depicts the block diagram of the impedance joint controller. The detailed description of the joint controller can be found at [13].

The proportional gain P_{imp} in Fig. 4 can be regarded as virtual stiffness. It differs per state and DoF and is described in the terms of "High" or "Low" for a high impedance or low impedance mode. The values for different states are listed in Table III.

Figure 4. Impedance controlled trajectory tracking diagram at a powered joint. The feedback loop are Proportional (P) controlled by P_{imp} to track the desired joint reference angle $\tilde{\theta}$ given the sensor feedback of the joint angle θ. The desired joint torque $\tilde{\tau}$ is tracked by the Series Elastic Actuator (SEA) which outputs the actual joint torque τ acting on the joint.

TABLE III. SUMMARY OF THE PROPERTIES OF THE P CONTROLLER FOR THE 9 STATES

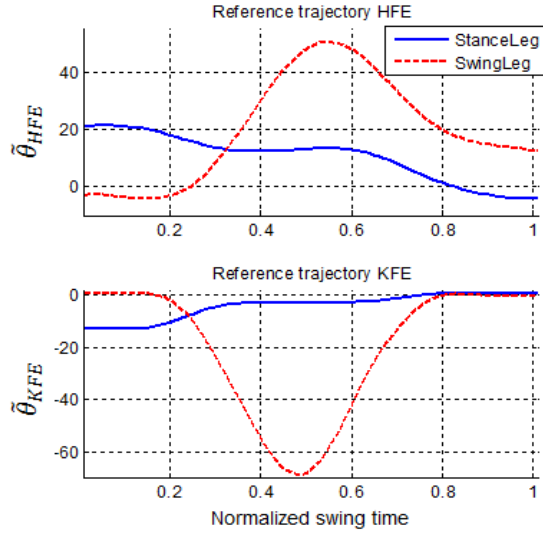| State	Plane	Impedance Mode	P Gains (Nm/rad)	Reference type
S1: stance	Sagittal	High	400	Position
	Lateral	Low	50	
S2,S6 Weight Shift	Sagittal	High	400	Trajectory
	Lateral	High	500	
S3,S4,S7,S8: half/full swing	Sagittal	High	600	Trajectory
	Lateral	High	800	
S5,S9: double stance	Sagittal	High	600	Position

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Figure 5. Reference swing trajectory for the sagittal joints HFE and KFE. The unit of Y-axes is degree. Hip flexion is positive. Knee extension is positive. Time = 0 denotes 0% of the swing phase. Time = 1, denotes 100% of the swing phase.

C. Walking Trajectory Generation

The walking trajectory is divided into three parts; stance/double stance, weight shifting and swing. The reference positions for stance/double stance (S1, S5, and S9) are predefined such that the user-exoskeleton is in a posture that is in equilibrium (CoM within the base of support). The reference trajectories for weight shifting (S2 and S6) are defined at the start of the state by smooth interpolation between the end posture of the double stance phase (when the step is triggered) and the beginning of the swing phase. The reference trajectories for the swing phases (S3, S4, S7, and S8) are defined separately in the sagittal plane and the frontal plane. All the reference trajectories are predefined. The reference trajectories in the frontal plane will be online adjusted. The adjustment method will be elaborated later.

1) Sagittal plane

The trajectory for the HFE and KFE joints during the swing phase were defined based on walking patterns of a healthy subject walking in the MINDWALKER while it was in zero-torque mode. As an example, the trajectory for the right swing (S4) is plotted in Fig. 5.

2) Lateral plane

The trajectory for the HAA joints should resemble the following pattern. The two powered HAA joints will shift the weight (CoM) of user-exoskeleton to the stance side before toe off. Just before heel strike, the HAA joints will move back to their zero positions which are the same as the reference pre-defined for the double stance phase. The key parameter for this pattern is amplitude of the weight shift which is quantified by $r_{lateral}$ in (2).

We need to determine a proper value of $r_{lateral}$ to match the timing between the sagittal and frontal movements. Although the sagittal and frontal plane motions are separately controlled, the movements are coupled in the way that they share the same swing time. Too small $r_{lateral}$ causes that the
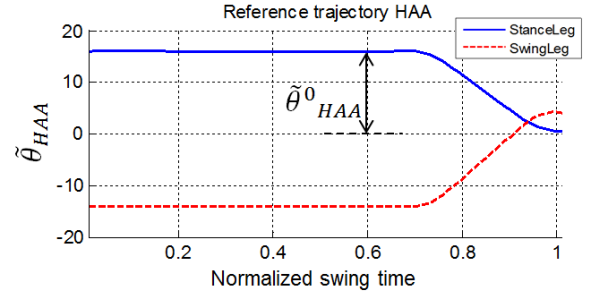

Figure 6. Reference swing trajectory for the lateral joints HAA. Hip adduction is positive.

user-exoskeleton falls to the neutral position too fast after lifting the swing foot. This may result a stumble. Too large $r_{lateral}$ has the risk that the user-exoskeleton falls over to the stance side.

Since MINDWALKER has no ankle in/eversion, $r_{lateral}$ can be only manipulated at the HAA joints. The HAA joint reference can be generated with a parameter, the nominal hip ab/adduction angle $\tilde{\theta}_{HAA}^{0}$, to control the desired amount of weight shift. The value of $\tilde{\theta}_{HAA}^{0}$ was determined by "trial and error". As an example, the HAA joint reference for the swing phase is shown in Fig. 6.

D. Trajectory Online Correction

To prevent the user-exoskeleton from falling sideways, we implement online correction of the step width by adapting the amount of hip ab/adduction needed during the swing phase. The required adjustment of hip ab/adduction is determined using XCoM [14]. If the user-exoskeleton system falls towards one side due to external perturbations such as being pushed at the shoulder or internal perturbations such as user's upper body motion, the foot placement is adjusted resulting in a wider or a narrower step width to counteract such perturbations.

In the concept of XCoM, the single stance phase of a bipedal gait is modeled as an inverted pendulum. The XCoM ζ in the frontal plane is defined in [14] as

$$\zeta = z_{CoM} + \frac{v_{CoMz}}{\omega_0} \tag{3}$$

where z_{CoM} and v_{CoMz} are the lateral position and velocity of the CoM of the inverted pendulum respectively. ω_0 is the eigenfrequency of the pendulum.

In fixed gait pattern without any perturbation, the averaged value of XCoM at certain period (e.g. mid-swing) in the swing phase is constant at every step with a constant z_{CoM} and v_{CoMz}, say z_{CoM}^{bs} and v_{CoMz}^{bs} where the superscript bs stands for baseline. When the system is perturbed during the swing phase, the averaged magnitude of the perturbation $\Delta \zeta$ at the same period in the swing phase can be expressed as

$$\Delta \zeta = \Delta z_{CoM} + \frac{\Delta v_{CoMz}}{\omega_0}$$

$$= z_{CoM} - z_{CoM}^{bs} + \frac{v_{CoMz} - v_{CoMz}^{bs}}{\omega_0} . \tag{4}$$

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Ignoring the CoM position change, which is usually much smaller comparing to the latter term, we have

$$\Delta \zeta = \frac{v_{CoMz} - v_{CoMz}^{bs}}{\omega_0} .$$

(5)

In order to compensate the perturbation such that the XCoM gets to a desired value just at heel strike, we need to adapt the step width. This was achieved by altering the desired HAA reference $\tilde{\theta}_{HAA}$. When the perturbation exceeds a threshold, a correction on the desired HAA reference is performed. The correction amplitude was derived from $\Delta\zeta$ as follows:

$$\Delta\zeta = L_{Leg} \sin(\Delta\tilde{\theta}_{HAA})$$

$$\Delta\tilde{\theta}_{HAA} = \arcsin\left(\frac{p(v_{CoMz} - v_{CoMz}^{bs})}{L_{Leg}\omega_0}\right)$$

(6)

L_{Leg} is the distance from the ground to the HAA joint. The parameter p scales the velocity changes in (6). It is empirically tuned to change the sensitivity of the correction. $p = 3$ was used in the actual implementation.

At this moment we implemented a one-time adjustment of the HAA trajectory during swing phase when the averaged perturbation $\Delta\zeta$ calculated at mid-swing increases above a certain threshold.

IV. EXPERIMENT AND RESULTS

Experiments were performed with one able-bodied subject to test the controllers and user interface as described in the previous section. The subject was a 28-year-old male (1.83 m, 76 kg).

The aim of the experiment was to test whether the subject was able to walk under the assistance of the exoskeleton without external balance aids (crutches or a walker). Straight walking of 10 steps was performed. At some steps, the
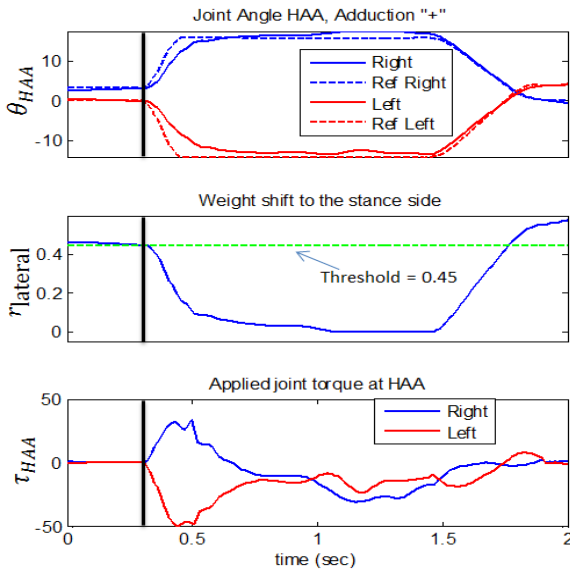

Figure 7. Weight shift assisted by MINDWALKER during one step. The black vertical line separates the double stance and the weight shift. The unit of joint torque is Nm.

Figure 8. A snapshot of a left step during the straight walking. The crutches were only used as safety protection and didn't touch the ground during the whole procedure.

subject was slightly pushed at the shoulder from the stance side to the swing side at the beginning of the swing phase by another person. Joint angles and torques at powered joints were recorded. Step width at double stance (S5 and S9) was computed from the recorded joint angles.

During the experiment, the subject was instructed to be passive in his lower limbs to emulate a paraplegic person.

A. Weight Shifting Assisted by MINDWALKER

In Fig. 7, the controller detected the intention of the subject to initiate weight shifting as soon as $r_{lateral} < 0.45$ at approximately $t = 0.3s$. The state transited from double stance to weight shift. The generated torques to track the desired hip abduction angles and to assist the weight shift amounted up to 50 Nm. At the end of the swing phase, approximately $t = 1.7s$, the HAA joints returned to their neutral position and $r_{lateral}$ indicated the weight was shifted back to the middle and further to the other side for the next step.

B. Straight Walking with Corrected Step Width

A sample video of the straight walking in the MINDWALKER exoskeleton was recorded during the experiment[1]. A snapshot of a left step is shown in Fig. 8.

In Fig. 9, the subject walked at approximately 0.21m/s with a step length of 0.43m. During the normal (unperturbed) step between $t = [0.5,2]s$, $\Delta\zeta$ in (4) did not surpass the threshold and nominal joint trajectories were tracked. During the perturbed step, the user-exoskeleton was pushed and the subject felt quicker to the swing side. At $t = 3.2s$, online correction of the HAA angles took place because $\Delta\zeta$ exceeded the threshold. This resulted in a larger hip abduction at heel strike and a larger step width at $t = 4s$.

C. Discussions

The goal of this paper was to design, implement and evaluate controllers to assist lateral weight shift for the user-exoskeleton and to adapt foot placement during walking.

From the results, the exoskeleton was able to impose a weight shift towards the stance side (the leading leg at the double stance) to unload the swing leg for the coming step. This suggests the proposed method was effective on providing lateral support for step initiation.
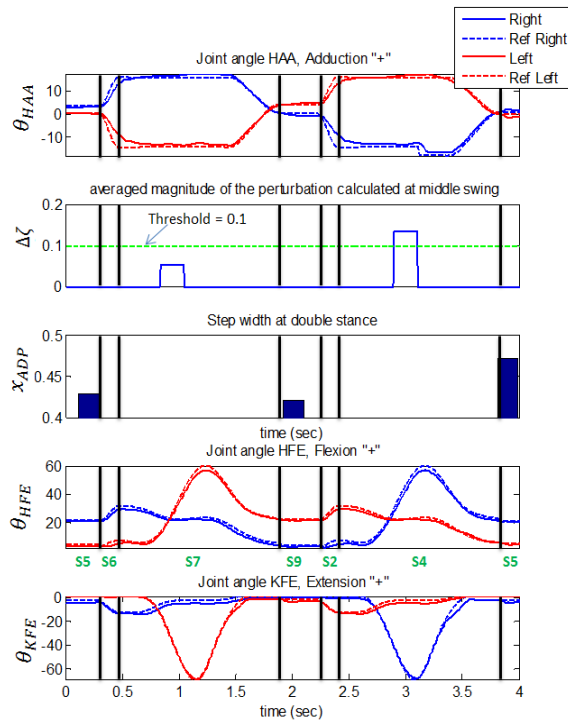
Figure 9. A gait cycle with 2 steps triggered by the subject. The black vertical lines separate the different states (labeled in green at x-axis below the 3ʳᵈ plot) during the walking. S5/S9: double stance with left/right foot in front. S2/S6: weight shifting to left/right. S4/S7: swing right/left leg.

During walking, no crutches or walkers were used to keep body balance. It indicates that self-balanced walking could be realized by applying the proposed method. To the author's knowledge, no self-balanced walking (without walking aids) has been reported or demonstrated using the exoskeletons [1-3,5,6] mentioned in the introduction section, even with healthy subjects.

In the experiment we instructed the subject to emulate a SCI patient. As the lower limb muscle activity of the able-bodied subject was not recorded and not controlled during the experiment, we cannot exclude that the healthy subject contributed to walking. Especially at the ankle joints (ADP), the subject might help the user-exoskeleton control the stability in the sagittal plane. This is because the passive stiffness at MIDNWALKER's ADP joint is relative small considering the magnitude of torque that a human subject can provide. In general, most of the subject's activities, especially in the frontal plane, were either overruled by the corresponding powered exoskeleton joints HAA, HFE and KFE (the smooth and reproducible joint trajectories in Fig. 8), or constrained by the passive joints of HRO with high stiffness, or even eliminated due to the fact that no DoF at ankle in/eversion is designed.

The next step is to investigate the use of the exoskeleton in paraplegic patients who have no motor control in their lower limbs and to investigate if stable walking can be achieved without using support aids. The added value of the powered hip/adduction can by systematically assessed in experiment by having subjects walking in the device with this DoF actively assisted by the exoskeleton or with this DoF locked. The performance of the online step width correction on walking stability can be systematically evaluated in the future by analyzing the ground reaction force and the CoM position with respect to the base of support during walking.

V. CONCLUSION

In this paper, a control method for assisted walking is proposed, implemented and evaluated using the powered lower limb exoskeleton MINDWLKER. The exoskeleton allows actively supporting the lateral weight shift to initiate a step and to control the step width during walking. A healthy subject could walk in the exoskeleton in a pre-defined gait pattern without any balance aids where his hip, knee and ankle joints were controlled or constrained by the exoskeleton. The control method and the MINDWALKER exoskeleton will be further tested and evaluated for paraplegic persons in the near future.

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