

An Omnidirectional Assistive Platform Integrated With Functional Electrical Stimulation for Gait Rehabilitation: A Case Study

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Abstract—This paper presents a novel omnidirectional platform for gait rehabilitation of people with hemiparesis after stroke. The mobile platform, henceforth the “walker”, allows unobstructed pelvic motion during walking, helps the user maintain balance and prevents falls. The system aids mobility actively by combining three types of therapeutic intervention: forward propulsion of the pelvis, controlled body weight support, and functional electrical stimulation (FES) for compensation of deficits in angular motion of the joints. FES is controlled using gait data extracted from a set of inertial measurement units (IMUs) worn by the user. The resulting closed-loop FES system synchronizes stimulation with the gait cycle phases and automatically adapts to the variations in muscle activation caused by changes in residual muscle activity and spasticity. A pilot study was conducted to determine the potential outcomes of the different interventions. One chronic stroke survivor underwent five sessions of gait training, each one involving a total of 30 minutes using the walker and FES system. The patient initially exhibited severe anomalies in joint angle trajectories on both the paretic and the non-paretic side. With training, the patient showed progressive increase in cadence and self-selected gait speed, along with consistent decrease in double-support time. FES helped correct the paretic foot angle during swing phase, and likely was a factor in observed improvements in temporal gait symmetry. Although the experiments showed favorable changes in the paretic trajectories, they also highlighted the need for intervention on the non-paretic side.

Index Terms—Stroke, rehabilitation robotics, gait recognition, wearable sensors, functional electrical stimulation (FES).

I. INTRODUCTION

APPROXIMATELY two-thirds of stroke survivors experience walking impairment [1]. The hemiplegia that commonly follows a stroke is characterized by impaired balance,

Manuscript received July 22, 2019; revised November 18, 2019; accepted January 30, 2020. Date of publication February 6, 2020; date of current version March 6, 2020. This work was supported in part by the Science and Engineering Research Council, Agency of Science, Technology and Research, Singapore, through the National Robotics Program under Grant No. 192 25 00045 and in part by the Faculty of Engineering, National University of Singapore under the Faculty Research Committee Tier 1 Grant No. R397-000-302-114. (Corresponding author: Haoyong Yu.)

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Digital Object Identifier 10.1109/TNSRE.2020.2972008

asymmetric voluntary movements and severe muscle weakness or paralysis. These alterations in motor function significantly compromise the person’s level of independence and affect their quality of life [2].

Assistive robotics combined with Functional Electrical Stimulation (FES) has been proposed as a method for early lower-extremity and gait rehabilitation post-stroke, and the number of rehabilitation robots utilizing FES has risen rapidly over the past ten years. Much of the effort in this field has been toward developing devices that integrate a treadmill with FES [3]–[5]. When using treadmill-based therapies, patients have shown improvements in walking speed [6], functional walking ability [7], [8] and range of motion of lower limb joints [6]. Moreover, the use of FES for drop-foot in stroke rehabilitation has shown to be effective in improving muscle strength [9] and motor function [10].

Although these studies suggest that individuals with stroke can recover gait capability by training on treadmill in combination with FES, there are certain inherent limitations to this method. Firstly, treadmill walking is biomechanically different to walking over ground [11]. Walking on a treadmill increases cadence by shortening step length, while decreasing anterior-posterior trunk acceleration and increasing forward tilt of trunk [12]. These changes alter dynamic gait stability and increase metabolic cost [12], [13]. Moreover, it has been shown that walking in a treadmill usually reduces the motion of the pelvis [14] which can severely affect the gait dynamics [15], [16].

In terms of muscle stimulation, most of the FES systems available for drop-foot treatment are ‘open-loop’, in the sense that the stimulation intensity, i.e. the combination of pulse width and pulse amplitude is kept constant [17]. Although systems like the WalkAide (Innovative Neurotronics, Austin, TX) employ motion data feedback, it is only in order to control the stimulation timing, i.e. to determine when to switch stimulation on and off during the gait cycle. The main limitation of open-loop FES is that it cannot adapt the stimulation intensity to changes in residual muscle activity and spasticity, which can occur in as few as five strides [18]. This requires frequent manual adjustment of the stimulation parameters.

To make FES more effective, the stimulation intensity should vary according to changes in the muscle’s response [19]; this requires the use of feedback, thus making

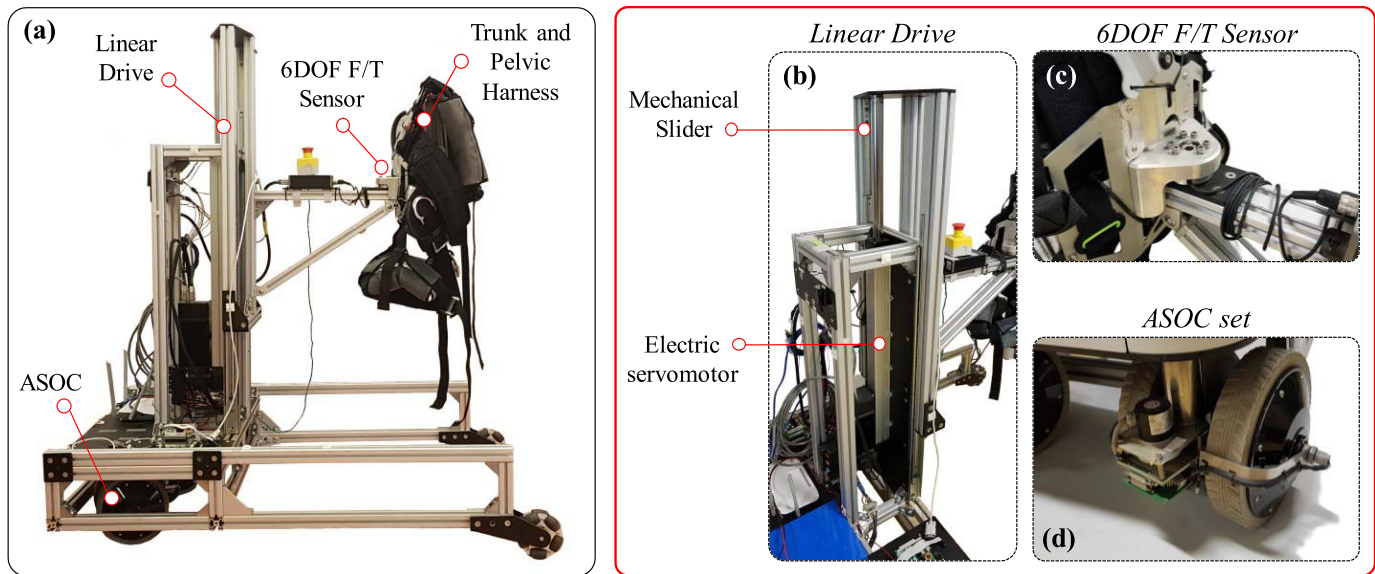


Fig. 1. The walker system. (a) Omnidirectional mobile platform and its associated subsystems. The platform is driven by a pair of active split-offset casters (ASOC). The trunk and pelvic harness connects to its support frame by means of the 6 degree of freedom (DOF) force/torque (F/T) sensor. The support frame of the harness moves vertically by means of a linear drive. Admittance control of the linear drive allows vertical displacement of the trunk during walking as well as body weight support (BWS). (b) Detail of the linear drive. (c) Detail of the 6 DOF F/T sensor. (d) Detail of the ASOC. The end-effector velocity commanded by the platform’s admittance control is converted to angular velocity commands for the ASOC by inverse kinematics.

the FES a ‘closed-loop’ system. In this research, we propose implementing the closed-loop approach by varying the FES pulse width in response to changes in joint flexion angle, which can be measured in real time from gait data. The underlying assumption is that changes in joint flexion reflect changes in the patient’s muscle activation.

In order to allow gait training over ground, and to help correct the patient’s gait patterns, we developed a rehabilitation system consisting of three main components [20]:

- An omnidirectional mobile platform or “walker” (Figure 1), which runs under admittance control to allow unrestricted motion of the user’s body along all three axes (forward-backward, lateral, and vertical), as well as full rotation about the vertical (yaw) axis. The control can also be programmed to provide active assistance or resistance while the user executes said motions.
- An inertial measurement unit (IMU)-based gait recognition system that computes lower limb kinematic data and detects gait phases.
- A closed-loop FES system that controls the timing of the stimulation and adjusts its intensity based on the kinematic data from the IMU system. In this way, the system accounts for natural variations in muscle response to stimulation.

In this study, the admittance controller was set to actively assist only the forward motion of the pelvis. The purpose of this study was to investigate the potential outcomes of gait training post-stroke using the walker with forward propulsion, combined with closed-loop FES. One chronic stroke survivor underwent five sessions of gait training, each involving a total of 30 minutes moving in the walker while FES was applied to the ankle dorsiflexors.

We analyzed the experimental data for changes in the patient’s gait which could be attributed to the forward propulsion provided by the platform, particularly in cadence and gait speed. We also sought evidence that closed-loop FES can elicit physiological foot clearance when applied to the ankle dorsiflexor muscles (i.e. tibialis anterior and peroneus longus muscles). Finally, we looked for changes in the patient’s levels of gait asymmetry. A number of favorable changes were observed in temporal symmetries, specifically in the proportion of stance and swing, and in single-support time. The experiments also yielded a few favorable changes in the paretic trajectories when compared to a clinically normal trajectory. Evolution of the non-paretic trajectories was more complex, in part because no specific intervention was applied to the non-paretic leg.

II. SYSTEM DESCRIPTION

A. The Omni-Directional Mobile Platform

Here we discuss the main features of the omnidirectional mobile platform. Further details are provided in [20]. The choice of making the platform omni-directional is based on the fundamental mechanism of locomotion during walking. During normal walking, pelvic forward-backward, side-to-side (lateral) and rotational movements are adapted to decrease the movement of the body’s center of mass (COM) in the vertical and horizontal direction, which reduces metabolic cost and mechanical work exerted in the lower extremity [21]. Restrictions of pelvic motion, on the other hand, can lead to abnormal muscle activation and gait kinematics causing abnormal gait patterns [21].

Our omni-directional mobile platform consists of a wheeled platform with two sets of active split offset casters (ASOC),

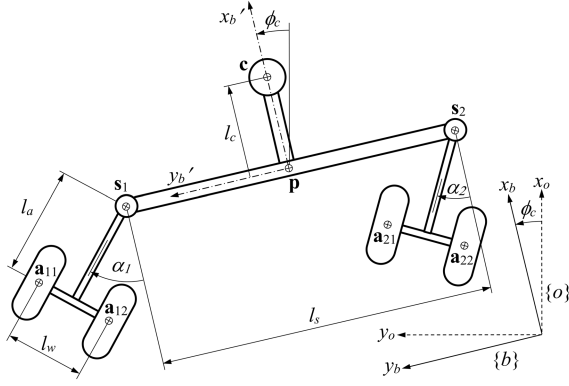


Fig. 2. Kinematic model of the omnidirectional platform. Passive casters are omitted for clarity.

a pelvic and trunk harness, a linear drive for BWS [22] and a force/torque (F/T) sensor for control (Figure 1). The F/T sensor and harness are mounted on the linear drive to support vertical movement of the user's COM. The combination of the mobile platform and the linear drive allows motion of the patient's pelvis along all three axes (forward-backward, lateral, and vertical), as well as full rotation of the body about the yaw axis.

Admittance control is employed to control the dynamic behavior of the platform during operation. The control's objective is to partially cancel the resistance to movement caused by the inertia and friction of the platform mechanism, thereby making the walker respond quickly to the patient's attempted motion. Input to the admittance controller is provided by the 6-axis F/T sensor, which detects the forces and torques exerted by the user on the platform through the harness. Based on the measured forces and torques, speeds in the forward, lateral and rotational direction are generated for the actuated wheels. The controller's admittance model is:

$$\mathbf{M}_d^b \dot{\mathbf{v}}_c + \mathbf{B}_d^b \mathbf{v}_c = \mathfrak{F}_{in}^b(\mathbf{F}_m) \quad (1)$$

where $\mathbf{v}_c = [v_{c,x} \ v_{c,y} \ \dot{\phi}_c]^T$ is the reference vector of end-effector linear velocities and body angular velocity about the z axis in coordinate frame $\{b\}$ (Figure 2). The virtual dynamics of the platform consist of the inertia matrix \mathbf{M}_d and the damping matrix \mathbf{B}_d , both of which are diagonal. In general, the inertia and damping values are chosen to be lower than those of the physical platform, as to reduce the effort exerted by the user in moving with the platform. On the right-hand side of (1), \mathbf{F}_m is the vector of measured forces and torques from the six-axis FT sensor, in coordinate frame $\{b\}$. We will assume straightforward use of the sensor readings, such that $\mathfrak{F}_{in}^b(\mathbf{F}_m) = [F_{x,m} \ F_{y,m} \ \tau_{z,m}]^T$. Integrating (1) generates the reference velocity \mathbf{v}_c , which is converted to velocity commands to each of the ASOC units through inverse kinematics.

In order to aid the user's motion on the horizontal plane, a computed force term ${}^b F_{asst}$ is added to the input of the admittance control:

$$\mathbf{M}_d^b \dot{\mathbf{v}}_c + \mathbf{B}_d^b \mathbf{v}_c = \mathfrak{F}_{in}^b(\mathbf{F}_m) + {}^b \mathfrak{F}_{asst} \quad (2)$$

Adding this computed force to the control causes the walker to exert an actual horizontal force on the patient's pelvis. In our current implementation, the horizontal force is assistive, i.e. it has the same direction as the patient's forward movement. The assistive force along the local ${}^b x$ axis (Figure 2) is given by

$${}^b \mathfrak{F}_{asst,x} = L_{asst,x} F_{asst,x} \quad (3)$$

where $F_{asst,x}$ is a reference force value and $L_{asst,x}$ is a modulating function that controls the amount of assistance felt by the user. As part of this modulation, the user experiences zero assistive force while at rest, and maximum assistance only when a certain walking speed has been reached.

Vertical motion of the linear drive is controlled with a separate admittance controller consisting of virtual mass $m_{z,d}$, virtual damping $b_{z,d}$ and virtual stiffness $k_{z,d}$. The admittance controller allows vertical movement of the pelvis during walking and can provide assistance in the form of partial BWS. The linear actuator's admittance model is:

$$m_{z,d} \ddot{z} + b_{z,d} \dot{z} + k_{z,d} (z - z_0) = F_{z,m} \quad (4)$$

where z corresponds to the vertical position of the linear drive and z_0 is the equilibrium point of the virtual spring with stiffness $k_{z,d}$. The desired BWS force is generated by adjusting z_0 . In order to apply a desired support force F_{BWS} , we set

$$z_0 = \frac{F_{BWS}}{k_{z,d}} \quad (5)$$

The strategy of using a virtual spring for BWS rather than commanding a force of the right-hand side of (4) was dictated in part by the need to guarantee the stability of the linear drive when coupled to the human body. The only drawback of this approach is that it causes small fluctuations in the vertical force acting on the trunk due to the vertical displacement of the pelvis. The virtual spring also acts as a safety measure: if the patient becomes fatigued during training, he or she can rest by supporting their body weight fully against the spring.

B. IMU-Based Gait Recognition System

Most commercially available FES systems for drop-foot correction are unable to adapt the intensity of the stimulation to the patient's response [17]. Our gait training system employs a set of wearable 9-DOF inertial measurement units to provide the kinematic data required to adapt the FES to changes in the patient's gait. The complete gait detection system (Figure 3) consists of seven IMUs (ADIS16405BMLZ, Analog Devices, Norwood, MA) connected in series. However, for the present application we only employ data from the IMU mounted on the paretic foot. The objective is to estimate the foot-to-ground angle (θ_{ilt}) during the swing phase. The peak value of θ_{ilt} will be employed as feedback to adjust the intensity of FES on the ankle dorsiflexors once per stride.

The foot-to-ground angle is estimated by means of a complementary filter. Prior to the gait training experiment, the ankle joint axis normal to the sagittal plane is determined using a calibration procedure adapted from [23]. The calibration involves a predefined set of motions performed by the subject.

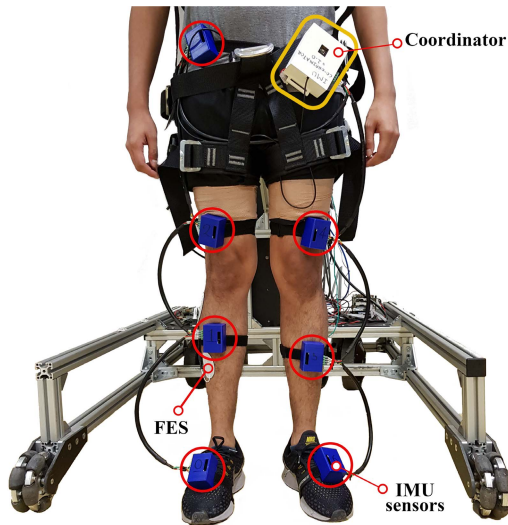


Fig. 3. The IMU system used in this pilot study was composed of seven inertial sensors placed on the right and left legs and at the waist.

During walking, the complementary filter obtains the best estimate for θ_{ilt} from two independent angle estimates: θ_{gyr} , which results from the numerical time integral of the angular velocity (ω_z) about the normal axis, and θ_{acc} , which is the angle estimate from the IMU acceleration vector. θ_{ilt} is obtained by evaluating the following update equations at each time step:

$$\theta_{gyr} = \theta_{gyr} + \omega_z \Delta t \quad (6)$$

$$\theta_{ilt} = \alpha \theta_{acc} + (1 - \alpha) \theta_{gyr} \quad (7)$$

where α is the complementary filter parameter (chosen to be 0.01).

FES is applied only during the swing phase. A finite state machine (FSM) with two states is implemented to detect the stance and swing phases of the paretic leg using data from the paretic foot IMU [23]. The detected gait phases and foot orientation angles are transmitted to the FES control computer over an RS232 data link. This information is used by the FES control algorithm to control the stimulation parameters in real time.

C. Closed-Loop Functional Electrical Stimulation

We targeted drop-foot in this initial study because it is a prevalent deficit among stroke patients [24], [25], and because drop-foot correction is a well-established objective for stroke rehabilitation [10], [24]. Although FES could have been applied to the knee muscles as well, we decided to use FES only on the ankle in order to limit the number of simultaneous interventions experienced by the patient, as the mobile platform already provides forward propulsion and balance support.

Our system uses a battery-operated FES unit (RehaStim, Hasomed GmbH, Germany) to stimulate the ankle dorsiflexors during the swing phase to correct drop-foot. Stimulation is synchronized to the gait phases derived from the IMU finite state machine. The stimulation pulses consist of biphasic

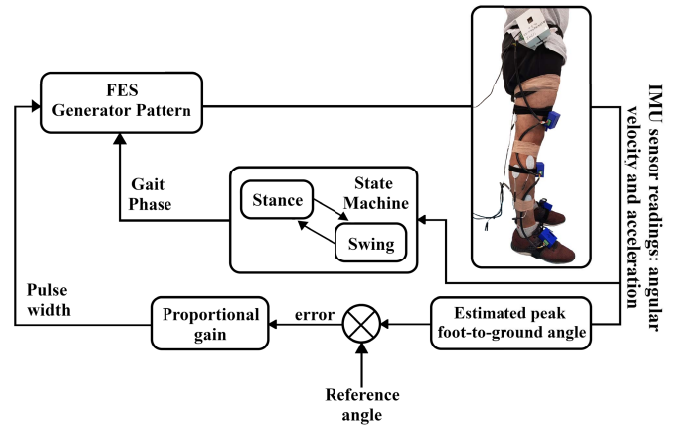


Fig. 4. Proportional foot-to-ground angle control. A method for controlling foot-to-ground angle which is proportional to the angle error, i.e. the difference between reference angle and the measured angle.

square-waveforms sampled at 35 Hz. Modulation of the force produced by the muscles is achieved by varying the pulse width. This allows more precise control and requires less charge per pulse than modulating the pulse amplitude [26].

The FES control uses a scheme similar to proportional control to help the patient achieve a desired peak value of foot-to-ground angle; we chose a reference value of 20° , which ensures enough foot clearance [16]. The control method is shown in Figure 4. After a stride is completed, the control computes an error value equal to the difference between the reference angle and the maximum measured value of the angle. The control output is a pulse width value proportional to the error; the computed pulse width is used to apply FES to the dorsiflexors during the swing phase of the next stride. It should be noted that, although FES is active during the entire swing phase, the pulse width is updated only once per stride.

The controller imposes upper and lower saturation limits on the pulse width. The lower saturation value corresponds to the minimum pulse duration (pulse width) required to induce muscle contraction [9]. The upper saturation value is the pulse duration capable of producing near-maximal muscle contraction without pain.

III. EXPERIMENTAL SETUP AND DATA ANALYSIS

A. Subject Description

Experiments took place at the Rehabilitation Center of the National University Hospital (Singapore). One chronic stroke patient (female, 57 y/o, 59 kg body mass, 165 cm height, 19 mo. after stroke) with right-sided paralysis underwent gait training with our system. She had drop-foot and presented weakness in dorsiflexion of the right foot and ankle. She also had problems in stabilizing the paretic leg during the stance phase and to support her own weight. On the other hand, she had good muscle response to FES. The patient gave her informed consent to take part in the study. The experimental protocol was approved by the Domain Specific Review Board of the National Healthcare Group, Singapore (DSRB reference 2017/00696).

B. Protocol

The patient underwent five sessions of gait training with the walker and FES system within a period of two weeks. On the first day, and prior to the first experiment, the subject underwent a 6-minute test of free walking. Gait data was recorded from the IMUs to provide a baseline against which to compare the experimental results.

Due to day-to-day changes in muscle state [27], FES pulse amplitude (i.e. the stimulation current) was recalibrated prior to each training session. The selected amplitude was the highest value that elicited functional response without causing pain at 250 μ s pulse width (upper saturation value). For the five sessions, the corresponding pulse amplitudes were 32, 34, 38, 50 and 48 mA.

Each experiment consisted of two periods of 15 minutes of uninterrupted walking with the assistance of the walker and FES. The subject walked back and forth along a straight 11-m long track. A 5-minute rest period was allowed between walking periods. The patient received forward assistance with a nominal value ($F_{asst,x}$) of 7.5% of the body weight. In order to limit the amount of intervention, no BWS was provided.

After placement of the FES electrodes, the patient was fitted with the set of 9-DOF IMUs. Sensors were placed on the foot, thigh, and shank of each leg. An additional sensor was placed on the waist. The patient was then coupled to the walker through the pelvic and trunk harness. The initial vertical position of the harness was adjusted to the patient's height.

C. Data Analysis

For each experiment, temporal gait parameters were computed based on the gait events toe-off (TO) and heel-strike (HS) from two consecutive steps. For the i -th step, the times at which events TO and HS occur are denoted, respectively, by $t_{TO}(i)$ and $t_{HS}(i)$. To calculate the parameters in seconds, each parameter was divided by the IMU sensor's sampling rate f_s . The following indices were calculated as follows [28]:

$$\text{Swing time}(i) = \frac{t_{HS}(i) - t_{TO}(i)}{f_s} \quad (8)$$

$$\text{Stance time}(i) = \frac{t_{TO}(i+1) - t_{HS}(i)}{f_s} \quad (9)$$

$$\text{Stride time}(i) = \frac{t_{HS}(i+1) - t_{HS}(i)}{f_s} \quad (10)$$

Subsequently, the durations of the stance and swing phases were normalized as percentages of the gait cycle. Double limb support time was calculated from the period in which one foot made initial contact while the opposite foot was still on the ground. Single-support times were calculated from the period in which only one foot (right or left) was in the ground.

To assess temporal asymmetry in the patient's gait, we analyzed inter-limb differences in swing percentage, stance percentage, stride time and double support time. Our chosen metric was the Symmetry Angle (SA), which allows to compare between discrete values obtained from the left and right sides [29]. The SA is related to the angle formed when a right-side value X_{right} is plotted against a left-side value X_{left} .

Two identical values generate a point on a line passing through the origin and forming a 45° angle with the horizontal (X_{right}) axis. Values that are not identical create a vector that forms an angle δ with respect to the 45° line and can be quantified as $\delta = 45^\circ - \arctan(X_{left}/X_{right})$. Thus δ provides a measure of asymmetry. The symmetry angle is then computed as

$$SA = \begin{cases} \frac{\delta}{90^\circ} \times 100\% & \text{for } \delta \leq 90^\circ \\ \frac{\delta - 180^\circ}{90^\circ} \times 100\% & \text{for } \delta > 90^\circ \end{cases} \quad (11)$$

A value of $SA = 0$ indicates full symmetry, whereas $SA = 100\%$ indicates the two values are equal in absolute value and opposite in sign. $SA = 50\%$ means the left value is zero, and $SA = -50\%$ means that the right value is zero.

IV. RESULTS

A. Temporal Gait Parameters

Analysis of patient's data revealed increases in cadence and selected walking speed, with respect to baseline, for all experiments except the first one (Table I). The reduction in cadence and walking speed during experiment #1 was most likely due to the patient being unaccustomed to walking in the mobile platform. The patient's effort to adapt to the walker during the first experiment is also evidenced by an increase in double support time (Figure 5(a)).

As it should be expected, for each experiment the stride times on the paretic side and the non-paretic one were essentially identical (Table I). However, there were significant differences among sides in the gait cycle percentage corresponding to swing. (And by extension, in the gait cycle percentage corresponding to stance.)

In baseline condition, these percentage differences reflect the extent of the patient's gait impairment and are evidenced by large symmetry angle values (SA). Remarkably, the symmetry angles of the swing and stance percentages decreased in all walker experiments including the first one. Thus, in terms of stance to swing proportion, the patient walked with greater temporal symmetry in the robotic device. There was also a marginal improvement in the symmetry of single-support time (Table I).

After an initial increase in double-support time during experiment #1, the patient consistently reduced the time spent with both feet in contact with the ground (Figure 5(a)). Single support time was essentially unchanged for the paretic limb (Figure 5(b)), whereas for the non-paretic limb it was slightly lower than baseline (Figure 5(c)).

B. Joint Angle Trajectories

In the baseline condition, the paretic side exhibited a severe anomaly in foot-to-ground angle during the swing phase, characterized by a mostly negative angle value, which is evidence of drop-foot condition (Figure 6). With the application of FES, the peak values of foot angle were consistently above the 20° reference, (Figure 6), with the maximum occurring during experiment #5 (Table II).

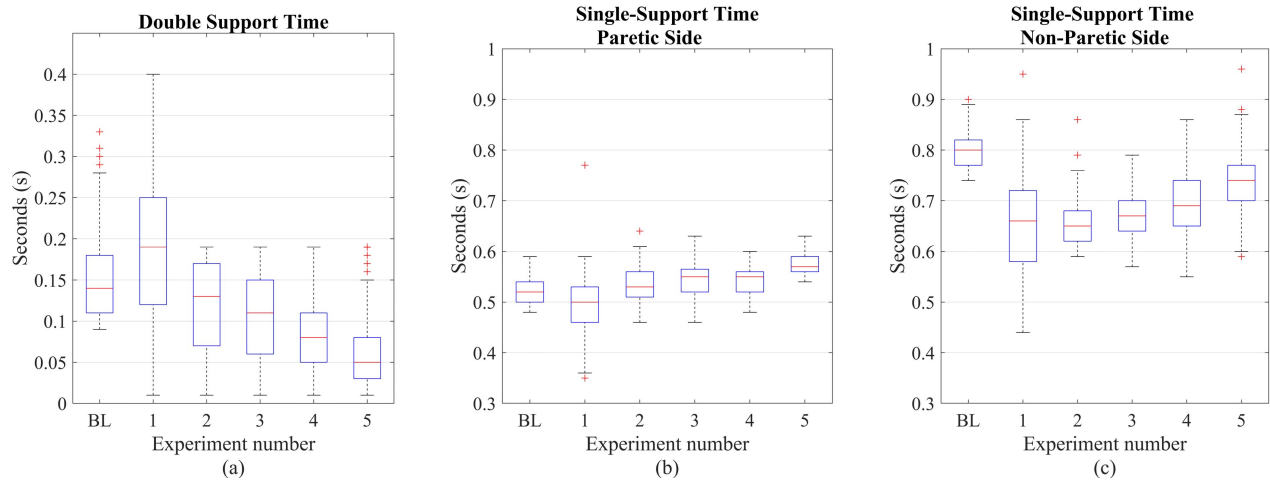


Fig. 5. (a) Timing of double-support, (b) Right single-support (affected side) and (c) left single-support (unaffected side) noted in the stroke patient during experiments. BL stands for baseline.

TABLE I

THE GAIT PARAMETERS OF THE STROKE SUBJECT DURING ROBOTIC GAIT TRAINING WITH THE ROBOTIC WALKER AND FES

	Experiment Number						
	Baseline	1	2	3	4	5	
Number of steps analyzed	235	170	223	241	255	265	
Cadence (steps/min)	35.2 ± 1.3	30.7 ± 5.0	41.2 ± 7.1	40.8 ± 6.7	46.1 ± 6.2	44.04 ± 6.5	
Walking speed (m/s)	0.21 ± 0.2	0.11 ± 0.1	0.19 ± 0.11	0.25 ± 0.12	0.28 ± 0.17	0.29 ± 0.14	
Swing (% gait cycle)							
Paretic side	42.9 ± 2.6	40.3 ± 6.6	44.7 ± 3.1	44.3 ± 2.7	55.1 ± 2.5	52.3 ± 2.6	
Non-paretic side	20.6 ± 0.8	32.4 ± 4.1	36.5 ± 3.7	36.5 ± 3.3	42.5 ± 2.6	40.2 ± 2.5	
SA (%)	-21.5	-6.89	-6.41	-6.13	-8.17	-8.28	
Stance (% gait cycle)							
Paretic side	57.0 ± 2.6	59.6 ± 6.6	54.9 ± 3.1	55.6 ± 2.7	44.6 ± 2.5	47.5 ± 2.6	
Non-paretic side	79.4 ± 0.8	67.5 ± 4.1	63.4 ± 3.7	63.4 ± 3.3	57.4 ± 2.6	60.0 ± 2.5	
SA (%)	10.4	3.95	4.57	4.17	7.95	7.37	
Single-support time (s)							
Paretic side	0.57 ± 0.043	0.49 ± 0.057	0.53 ± 0.035	0.54 ± 0.034	0.54 ± 0.029	0.57 ± 0.04	
Non-paretic side	0.78 ± 0.033	0.65 ± 0.096	0.65 ± 0.059	0.67 ± 0.046	0.69 ± 0.062	0.74 ± 0.053	
SA (%)	9.82	8.88	6.45	6.81	7.73	8.22	
Stride time (s)							
Paretic side	1.50 ± 0.22	1.51 ± 0.06	1.44 ± 0.05	1.45 ± 0.04	1.34 ± 0.03	1.40 ± 0.03	
Non-paretic side	1.52 ± 0.14	1.56 ± 0.08	1.45 ± 0.05	1.45 ± 0.05	1.33 ± 0.03	1.39 ± 0.03	
SA (%)	0.42	1.04	0.22	0	-0.24	-0.23	

SA stands for Symmetry Angle.

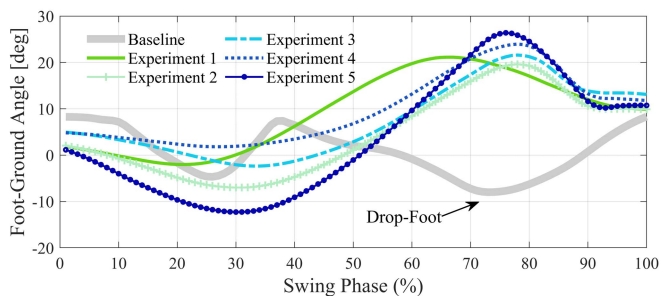


Fig. 6. Average foot-to-ground angle of the paretic limb during swing phase of gait cycle. The swing phase was defined from toe-off (0%) to heel-strike (100%).

As the foot-to-ground angle increased, average values of the pulse width consistently decreased, from 248.2 μs during experiment #1, to 203.5 μs during experiment #5 (Table II).

This suggests a gradual increase in the contribution of the patient’s muscles to the control of the foot’s orientation.

To assess the evolution of the patient’s gait, we compared her joint trajectories against the averaged trajectories from healthy subjects (Appendix). Figure 7 and Figure 8 show, respectively, the trajectory comparisons for foot dorsi/plantar flexion angle and knee flexion/extension angles.

Both the paretic and the non-paretic trajectories of the foot dorsi-plantar flexion angle exhibited severe anomalies at baseline. These anomalies evolved considerably with gait training (Figure 7). In baseline, the paretic foot essentially only exhibited plantarflexion. During the walker experiments, the situation was reversed: the foot angle trajectory exhibited mostly dorsiflexion, no doubt because of the electrical stimulation. During experiment #1 the dorsiflexion was relatively constant throughout the gait cycle, whereas during experiment #5 the angle trajectory became closer to normal, which

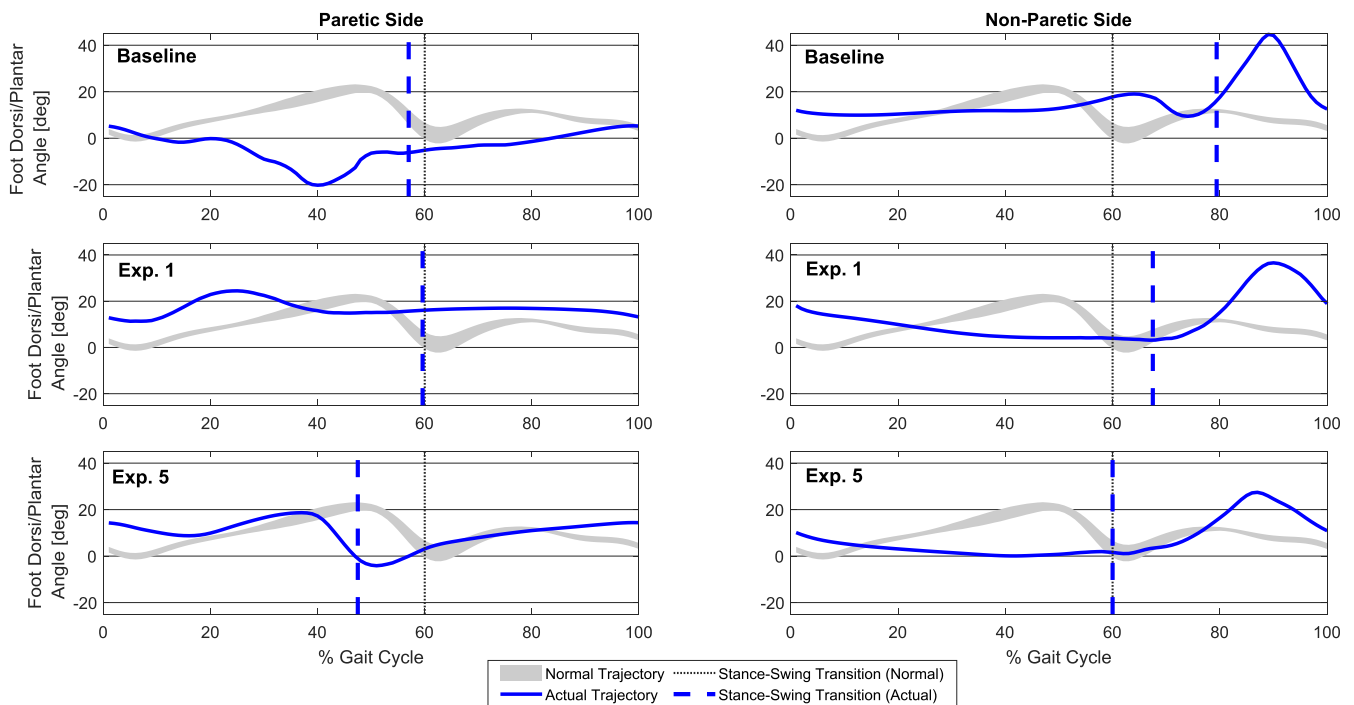


Fig. 7. Kinematics of the foot dorsi/plantar flexion of the paretic and the non-paretic limbs of the stroke patient. Data is presented from baseline, experiment #1 and experiment #5. Gray shaded area indicates the normal range of motion from healthy individuals.

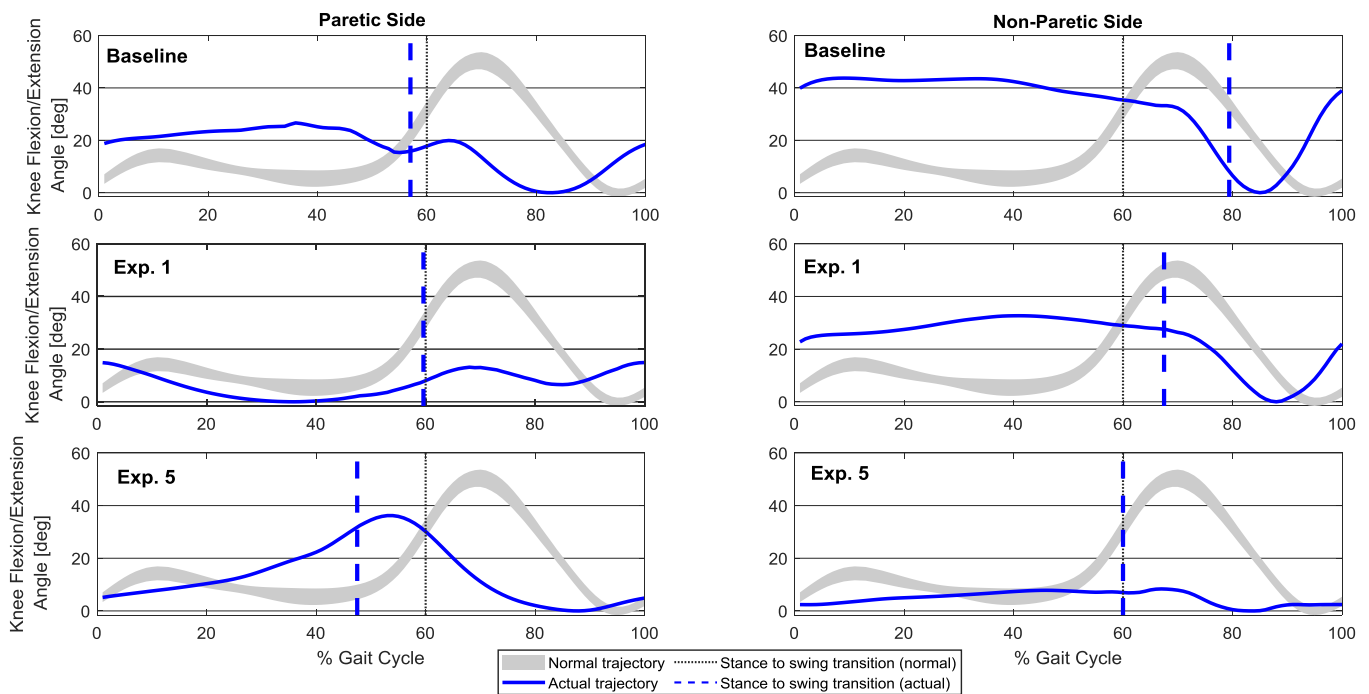


Fig. 8. Kinematics of the knee flexion/extension of the paretic and non-paretic limbs of the stroke patient. Data is presented from baseline, experiment 1 and experiment 5. Gray shaded area indicates the normal range of motion from healthy individuals.

suggests some adaptation in muscle activation on the part of the patient (Figure 7, left column). The foot-angle trajectory of the non-paretic side exhibited anomalies as well, most notably a marked dorsiflexion occurring during the swing phase. This is perhaps a result of the non-paretic side compensating for the motor deficits on the paretic side.

Even more striking anomalies occurred in the knee flexion angle trajectories (Figure 8). At baseline, the patient tended to

walk with both knees flexed for most of the gait cycle, with the largest flexion angle occurring on the non-paretic side. With gait training the patient developed something akin to a normal knee angle trajectory on the paretic side, but with the stance-to-swing transition and the peak flexion angle occurring earlier than normal in the gait cycle. On the non-paretic side, the outcome was a gradual disappearance of knee flexion from the gait cycle.

TABLE II
AVERAGE PULSE WIDTH AND MAXIMUM FOOT-TO-GROUND ANGLE OVER THE 5 DAYS OF EXPERIMENTS

BL	Exp. 1	Exp. 2	Exp. 3	Exp. 4	Exp. 5
Pulse Width (μs):					
N/A	248.2	204.2	211.6	209.4	203.5
	± 10.3	± 38.1	± 37.7	± 38.8	± 40.4
Maximum foot-to-ground angle ($^\circ$):					
7.43	17.66	20.3	20.01	23.22	25.08
± 1.79	± 5.47	± 5.40	± 4.74	± 4.42	± 4.97
Percentage of steps at upper saturation (250 μs):					
N/A	93.5	23.8	32.4	28.2	26.0
Percentage of steps at lower saturation (150 μs):					
N/A	0.6	13.5	10.4	16.9	21.1

Standard deviation in the pulse width values is rounded to the second decimal point. BL stands for Baseline.

V. DISCUSSION

This research aimed to determine the potential outcomes of gait training post-stroke using the omnidirectional platform in combination with closed-loop FES. To that end, we ran a short study consisting of five training sessions with a chronic stroke patient. Then we analyzed the gait data for clues about the potential outcomes that could be expected from a longer training program. A second objective was to use the experimental results as a guide for improving the experiment design and determining intervention dosages in future studies.

The system presented here can be placed in the category of balance support devices for walking over ground [30]. However, ours is the first of its kind in that it provides two complementary interventions that can be modulated with full independence of each other: forward propulsion and FES of the lower-limb muscles. Our robotic walker supports forward propulsion of the pelvis on the horizontal plane and provides body weight support if required. By contrast, most walker platforms are limited to providing balance support [31]–[33], lacking mechanisms to directly improve gait speed.

The most salient points of the experimental results can be summarized as follows:

- In general, the patient's gait parameters worsened during experiment #1, most likely as a result the effort involved in adapting to the walker and FES.
- That was followed by progressive improvement in several measures of gait performance, from experiment #2 onwards (Table I).
- At baseline, the anomalies in joint angle trajectories were severe for both the paretic and non-paretic sides. Evolution towards a normal trajectory could be observed on the paretic side; the corresponding evolution on the non-paretic side was harder to interpret.

A. Temporal Gait Parameters: Effect of Assistive Force

The platform's admittance control was intended to contribute to the patient's mobility, in part by providing forward assistive force. As early as experiment #2, the walking speed, cadence and double-support time compared favorably against baseline and continued to improve with each new experiment (Table I, Figure 5). This suggests that the patient learned to make use of forward assistance to propel her COM faster.

During experiment #1 the patient's gait was quite unsteady; she frequently needed to support herself on both feet before taking the next step. This is consistent with the fact that patients with stroke generally spend more time in double support to maintain balance [34]. As the experiment series progressed, the patient consistently reduced her double support, achieving a third of the time reported at baseline (Figure 5). This decrease suggests use of the walker system enhanced dynamic balance during ambulation.

The patient also displayed a typical stroke gait in that the non-paretic side expended more time in stance than the paretic side (Table I). Stance time on the non-paretic side decreased over the course of the experiment and resulted in improved temporal symmetries (stance and swing percentage). This too can be attributed in part to the forward propulsion provided by the walker.

B. Drop-Foot Correction: Effect of FES

The closed-loop FES served well its primary purpose of correcting the patient's drop-foot. By helping to normalize the swing trajectory, it may also have been a factor in improving the symmetry of the swing and stance percentages. On the other hand, FES tended to produce peak angles well above the 20° target (Figure 6). Although this may be a result of changes in the patient's muscle activation, another likely factor was that the feedback control of the FES tended to operate at its lower saturation value during part of the experiments. In this regard, Table II shows how the percentage of steps taken at lower saturation evolved, from almost negligible, to 21.1% during the last experiment. This means that part of the time the patient received a stronger FES than what the angle error dictated. Thus, the guidelines for calibrating FES pulse width may have overestimated the lower saturation value of pulse, in which case new criteria may be needed when closed-loop control is involved.

C. Joint Angle Trajectories: Assessment

Initially, the joint angle trajectories for both the paretic and non-paretic sides were highly abnormal. On the paretic side, the ankle joint exhibited almost pure plantarflexion in baseline (Figure 7). Angle trajectories also showed that the patient habitually bent the knees during stance (Figure 8), which indicates that she had difficulty supporting her weight despite being ambulatory. This effect was more marked on the non-paretic side, meaning that the patient tended to support a larger proportion of her weight on the healthy leg.

1) *Trajectories on the Paretic Side:* On the paretic side, the ankle joint angle trajectory evolved, from almost pure plantarflexion in baseline, to a trajectory with predominantly dorsiflexion that bore similarities to a healthy trajectory. This is attributable mainly to the FES.

A less expected result was that the paretic knee joint angle also evolved towards a healthy trajectory (Figure 8) even though no direct assistance was provided to the knee flexor or extensor muscles. A possible explanation is that, during the experiment series, the patient's gait changed from having repeated phases of near-static support, to involving mostly

dynamic balance. In other words, the gait became closer to normal human gait. The implication is that dynamic balance requires less gravitational load bearing from the limbs, and consequently would cause less bending of the knee.

2) *Trajectories on the Non-Paretic Side*: The non-paretic ankle joint angle (Figure 7) had a highly unusual trajectory, with large dorsiflexion occurring during swing. With repeated experiments the anomaly tended to disappear. We speculate the abnormal dorsiflexion may be a consequence of an adaptation mechanism to compensate gait deficiencies on the paretic side; with the combined interventions of assistive force and FES, the need for such compensation would diminish.

The non-paretic knee joint angle showed a similar evolution. The near-constant levels of knee flexion decreased gradually until, in experiment #5, the patient walked with her knee practically extended at all times. While this is still abnormal behavior, it also seems to confirm that the patient achieved some degree of dynamic balance, which would in turn require less gravitational support from the knee joint.

Notably, although the stimulation intensity in our closed-loop FES was proportional to the ankle dorsiflexion error, the stimulation was not intended to enforce a motion trajectory on the joint. The purpose of FES was to promote motor learning, as previous studies have shown that frequently repeated movements induced by FES help to reinforce neural network connection patterns in stroke subjects [35]. By contrast, other studies have found that enforcing a joint angle trajectory (by means of a powered orthosis or a similar device) may impair motor learning as the effects of the training essentially vanish once the robot-assisted therapy is removed [36].

D. Consequences for Experiment Design

The main value of the experimental results presented here lies in the fact that the patient's paretic foot angle (Figure 6) and ankle angle (Figure 7) trajectories evolved in a manner consistent with motor learning after only 5 sessions. Specifically, the patient went from almost pure plantarflexion at baseline, to a trajectory with predominantly dorsiflexion during swing that bore similarity with the healthy trajectories.

Given that post-stroke gait training programs involving FES tend to run for 10-12 weeks [37], we plan to test whether the combination of forward propulsion of the pelvis and FES can enhance motor learning, and do so to an extent that allows reducing the duration of the training program. Propulsion of the pelvis partially reduces the amount of torque required to flex or extend the lower-limbs' joints. Thus, the specific hypothesis is whether this torque reduction can facilitate the re-learning of symmetric joint angle trajectories.

The single most important lesson we derived from this study is that attempting to correct gait deficits only on the paretic side may not be enough for rehabilitation. Although we saw a reduction in the anomalies of the non-paretic joint trajectories, we did not see signs of a normal trajectory emerging on that side. A key point is that, just as recovery after stroke requires motor learning, compensatory movements on the non-paretic side are a learned strategy as well [38], except that they involve learning an abnormal muscle activation pattern. And to the extent the patient becomes reliant on them, compensatory movements tend to limit proper recovery [39]. In consequence,

attempting to correct only the motion deficits directly attributable to the stroke (such as ankle or knee flexion deficits) may have little effect the compensatory movements that the patient has already acquired.

Furthermore, there is growing evidence that bilateral electrical stimulation can be more effective than unilateral intervention in improving motor function after stroke [40]–[42], mainly because it targets not only the motion deficits on the paretic side but also the compensatory movements on the non-paretic side. Accordingly, we plan to test whether trying to actively correct the compensatory movements via FES will result in better outcomes than treating only the primary motion deficits, i.e. the ones directly linked to the stroke.

Patients' difficulty in supporting their weight may prove an obstacle to improving their gait pattern, especially if the weight support distribution is not uniform. Therefore, patients may benefit from receiving BWS as part of the intervention even if they are considered ambulatory. However, BWS has the drawback of altering the dynamics of the gait cycle. When only BWS is provided, subjects tend to lower their stride frequency in proportion to the amount of support provided. However, in preliminary experiments we found that forward assistance can be used in combination with BWS to eliminate the alteration in stepping frequency. This suggests that a simple functional relationship between BWS and forward assistance could be derived for users of the walker. The resulting function would allow to determine, for a given amount of BWS, the amount of forward assistance needed to minimize disruption of the patient's stepping frequency.

VI. CONCLUSION

We conducted an initial evaluation of a gait training system composed of an omnidirectional platform and a closed-loop FES system for correcting joint trajectory deficits. The platform employs admittance control to maximize its mobility; the same control allows to provide horizontal assistive forces to the user, which are modulated with gait speed. The FES system helps the user compensate deficits in joint angular motion by means of proportional feedback control.

We obtained preliminary evidence that the platform's assistance is effective in increasing in cadence and self-selected gait speed, as well as improving temporal gait symmetries. The FES system on its part proved effective at compensating joint motion deficit, although it tended to exceed the target value of peak foot angle. At the same time, the study highlighted the need for correcting gait anomalies on the non-paretic side as well, together with the potential value of using BWS by system. We plan to use these findings in the design of future experiments. The key research question is whether training with the pelvic propulsion and FES combination can accelerate motor learning with respect to conventional methods.

APPENDIX: GAIT DATA FROM HEALTHY SUBJECTS

Average joint trajectories were calculated using gait data from 10 healthy subjects (5 males and 5 females, age 25.5 ± 2.17 years, height 1.71 ± 0.09 m, body mass 68.3 ± 9.5 kg). We chose to use a three-dimensional motion capture system for maximum reliability [43]. The system features 8 infrared cameras (VICON Motion Systems, Oxford,

United Kingdom). 15 retro-reflective markers were placed on the lower limbs of the subjects following the Plug-in-Gait marker placement model and kinematic data were collected at a sampling rate of 100 Hz.

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