A Hybrid Approach towards Assisting Ankle Joint of Paretic Patients

Victor Arnez-Paniagua, Weiguang Huo, Ivan Colorado-Cervantes, Samer Mohammed, Yacine Amirat

***Abstract*— Active orthoses that assist the ankle joint of patients with hemiplegia require accurate real-time estimation of the gait cycle's time evolution. In this paper, we present a fuzzy logic based algorithm to accurately determine the sub-phases of the gait cycle using ground reaction forces (GRF) sensors. A hybrid system integrating an active ankle-foot orthosis (AAFO) and a functional electrical stimulation (FES) is used to provide appropriate assistance to the ankle joint based on the gait cycle sub-phases evolution. To generate a push-off at pre-swing, the AAFO's actuator is activated to produce a forward impulse of the foot towards the mid-swing where the FES is activated to provide toe-clearance during terminal swing. Preliminary results show the feasibility of the proposed approach in terms of providing appropriate assistance as a function of gait sub-phases detection. Experiments were conducted with two healthy subjects walking at fixed speed on a treadmill.**

I. INTRODUCTION

Nowadays stroke is a leading cause of long-term adult disability making the walking function as one of the primary concerns for stroke patients [1]. Post-rehabilitation patients still present residual gait deficits such as foot drop, a common post-stroke gait impairment estimated to affect 20% of survivors and it is caused by total or partial paresis of ankle dorsiflexion muscles. As a result, ground clearance is a difficult task during the swing phase leading generally to inefficient gait compensation by moving the foot arc away from the body. This gait inefficiency is due to the fact that the toes remain in contact with the support surface while the hip abduction of the unaffected limb increases during the stance phase [2]. This will lead to an increasing metabolic cost of the walking activities along with decreasing endurance and increasing risk of falls of those patients [1].

In the past few decades, an increasing number of assistive devices have been developed to overcome gait impairments [3]. Those devices range from passive and active ankle joint orthoses, to full size exoskeletons [4], including neuro-orthosis, such as FES [5].

One main advantage of AAFO systems is the relatively low fatigue of the lower limb muscles during walking activities, as external energy sources are used to enhance movements of the wearer's joints, thus achieving good results in repeatability and assistance [4]. The effectiveness of FES has been proved to produce positive orthotic effects on many gait parameters, such as increasing walking speed and improving symmetry index [6]. FES offers many advantages compared to AAFOs, such as active muscle contraction, muscle strength improvement [7][8], muscle tone reduction and efficient energy use of proximal lower limb [9]. However, there are some challenges facing the development of such systems for long-term daily use such as the rapid muscular fatigue, the great physical effort required, the need to extensive training programs.

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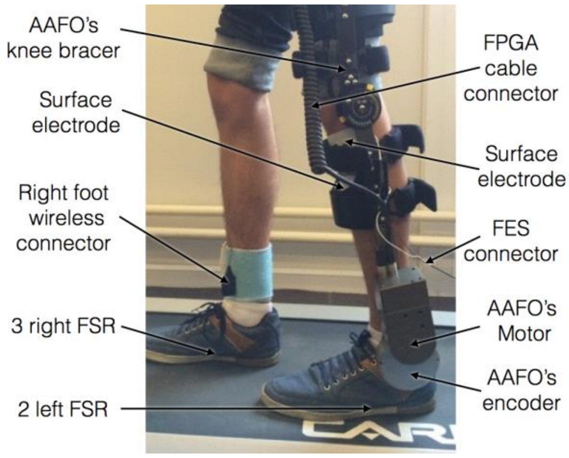

Figure 1. System setup.

In this study, a hybrid system that incorporates an AAFO and a FES is proposed. The goal is to prevent foot drop in hemiparetic patients while improving muscle tone by using FES, and reducing fatigue due to the assistance provided by the AAFO. The setup is shown in Fig. 1. The feasibility of using both actuators in one setup is evaluated.

Several actuated orthoses use gait detection in order to decide actuator trigger of such devices [10]. The detection process is mainly based on the use of simple foot switches [3], shoes with embedded pressure sensors [11] and inertial sensors [12], etc. Furthermore, they use different algorithms such as finite-state machines [3], fuzzy logic [11], model of the gait cycle as a function of shank [10], or regression models to identify the gait phases using in-shoe pressure mapping system [13].

In order to trigger the proposed hybrid system, a fuzzy logic algorithm based on embedded FSR sensors in the insoles of both feet, is used.

II. GAIT PHASE DETECTION

Human walking involves repetitive patterns known as gait phases. The gait or locomotion cycle consists of two basic phases, the stance phase and the swing phase [13][14]. The stance phase starts when the foot makes contact with the ground, while the swing phase occurs when the foot is swinging without touching the ground. Furthermore, the stance phase can be divided into five sub-phases that are: loading response (LR), early mid-stance (EMS), late mid-stance (LMS), terminal stance (TS) and pre-swing (PS). The swing phase consists of three sub-phases that are: initial swing (ISw), mid-swing (MSw) and terminal swing (TSw).

The beginning and duration of each sub-phase are detected using the following events: initial contact (IC), opposite toe off (OTO), heel rise (HR), opposite initial contact (OIC), toe off (TO), foot adjacent (FA) and tibia vertical (TV). Note that IC refers to the heel strike in this study.

The sub-phases of the stance phase are detectable by analysing the GRF. During the swing phase, an approximation of the duration of the sub-phases is proposed by measuring the contralateral extremity's GRF. Initial swing is detected when contralateral is in early mid-stance sub-phase; mid-swing occurs when contralateral is in late mid-stance, and terminal swing is detected when contralateral is in terminal stance. Therefore, we will call the swing sub-phases as approximate initial swing (AIS), approximate mid-swing (AMS) and approximate terminal swing (ATS).

To measure the GRFs, five FSR sensors were used (Fig. 2). Two FSRs are placed on the surface of the left foot's insole (at the heel and at the base of the fourth metatarso) and three FSRs are placed under the right foot; one FSR is aligned with the heel and the second and third are aligned with the first and second metatarsophalangeal joint.

Based on a fuzzy logic algorithm, the likelihood of each sub-phase is determined using a fuzzy inference system with membership function (FMV). The rules that define each sub-phase's probability are described in Table I. Firstly, we define the magnitude range for each sensor r_i and the absolute threshold value N_i in order to calibrate the sensors for each trial:

$$r_i = max(\boldsymbol{F}_i) - min(\boldsymbol{F}_i) \quad (1)$$

$$N_i = r_i h + min(\boldsymbol{F}_i) \quad (2)$$

where $\boldsymbol{F}_i$ denotes the vector that contains the acquired data, h is the relative threshold, normally a 5% of the range (r_i) is effective. The membership function for each sensor is expressed as follows:

$$f_i = \frac{1}{2}\left(tanh\left(\frac{F_i - N_i}{r_i} - 1\right) + 1\right) \quad (3)$$

where F_i represents the sensor's reading. The FMV for each sub-phase is calculated based on Table I and using Zadeh operators, e.g., the FMV for LR is as follows:

$$\mu_{LR} = min(f_{lh}, 1 - f_{lt}, 1 - f_{rh}, f_{rt}) \quad (4)$$

Only the sub-phase having the maximum probability will be considered. Afterwards, the choice between the ankle joint robot or FES as the current actuator, will be done based on the current sub-phase. Fig. 3 shows the FMVs for each sub-phase of the gait cycle as defined above.

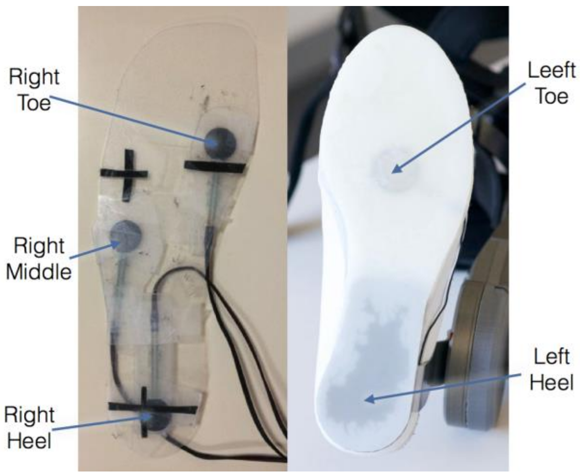

Figure 2. FSR sensors. The left foot's insole is attached to the AAFO, it has two fixed sensors. The right foot's insole has three adjustable sensors.

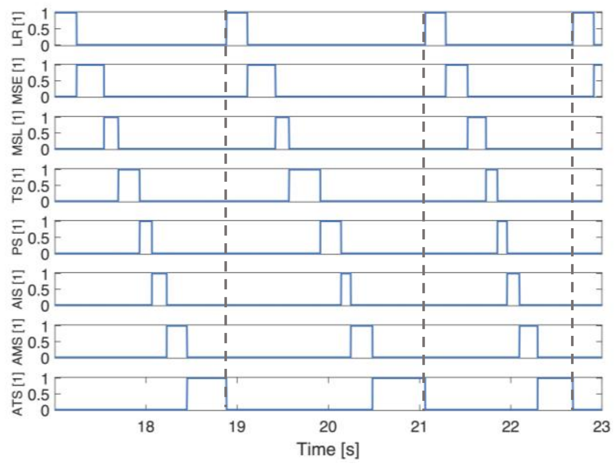

Figure 3. Maximum probability selection for each sub-phase's likelihood during walk on treadmill. The dotted lines indicate the separation between strides.

TABLE I. RULES DEFINING THE SUBPHASES PROBABILITIES

f_{lh}	f_{lt}	f_{rh}	f_{rm}	f_{rt}	μ_i
large	small	small	N/A	large	μ_{LR}
large	small	small	N/A	small	μ_{MSE}
large	large	small	N/A	small	μ_{MSL}
small	large	small	N/A	small	μ_{TS}
large	N/A	small	N/A	N/A	μ_{PS}
small	small	large	N/A	N/A	μ_{AIS}
small	small	small	large	N/A	μ_{AMS}
small	small	small	N/A	large	μ_{ATS}

f is the membership function, the underscores are: l: left foot; r: right foot; h: heel sensor; t: toe sensor; m: middle sensor, μ represents the fuzzy variable that contains the sub-phase probability. The value N/A means that the sensor's value does not influence the sub-phase's FMV.

III. HYBRID CONTROLLER

Adults with hemiplegia often demonstrate two common gait impairments: inadequate dorsiflexion causing drop-foot, and plantar flexor spasticity or stiffness causing decreased push off. Based on the gait phase detection presented in the previous section, a hybrid control approach using both FES and AAFO is used in this study. The goal is to assist the wearer to have a better ankle plantar flexion movement between terminal stance and mid-swing sub-phases using AAFO while the tiabialis anterior muscle is stimulated using FES between mid-swing and terminal swing, as seen in Fig. 4.

A. Actuated Ankle-Foot Orthosis

A recent study [15] has shown that powered plantar-flexion movements are greater in healthy than in paretic patients, which is probably due to an increase in plantar-flexion co-contraction. Patients tend to compensate the resulting lack with hip hiking and circumduction. Therefore, adding an assistive force to the ankle joint at an appropriate moment of the gait cycle could improve the locomotion activities of the paretic patient. Preliminary results with patients show that plantar-flexion assistance through pneumatic artificial muscles, in combination with on/off controller, contribute to improve lower limb joint kinematics [16].

During terminal stance, the body weight moves ahead of the forefoot, requiring powered plantar-flexion to stabilize the ankle and prevent over dorsiflexion movement. In pre-swing, when the limb has to be positioned for swing, the lack of powered plantar-flexion movement can lead to inefficient gait compensations. Therefore, in terminal stance, the AAFO assists the ankle by providing a plantar-flexing torque, then in pre-swing, a higher torque is generated to create the initial push-off impulse. To achieve this objective, an open-loop control strategy is implemented, where the gait phase detection

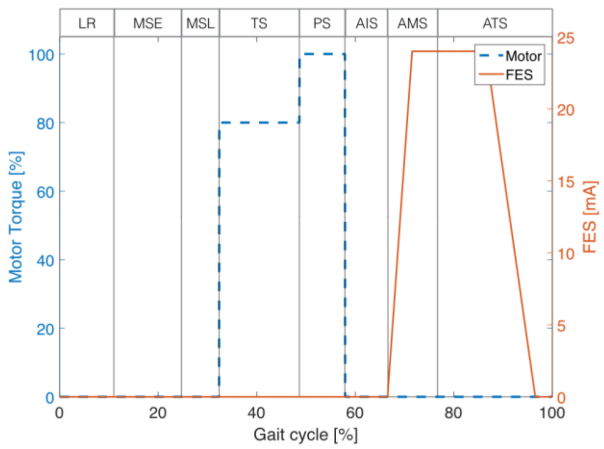

Figure 4. The AAFO is triggered during the TS and PS sub-phases, while the FES is triggered with AMS and ATS.

algorithm described above triggers a constant assistive torque during terminal stance. During pre-swing; a higher control torque value is set in order to generate the desired push-off assistive torque. This lasts for the duration of the pre-swing sub-phase. The two constant torque values were tuned heuristically. This was achieved by applying continuously increasing values to the AAFO's motor, while observing the ankle joint reaction during a treadmill walking session prior to the experiment.

B. *Functional Electrical Stimulation*

Despite the fact that AAFOs contribute to afferent peripheral input beneficial to the paretic patient, it does not always produce functional meaningful afferent feedback. During dorsiflexion actuation, for example, the AAFO provides pressure at the plantar surface of the foot, and thus may contribute to inappropriate afferent input during the swing phase. Therefore, using FES during swing phase is appropriate to provide meaningful afferent feedback [17].

Trapezoidal FES stimulation profiles are mostly applied in clinics to face foot drops of paretic patients. FES is linearly ramped up to its maximum value at toe-off, stimulation current amplitude is then kept constant until an established condition is reached, where it is ramped down to zero. These stimulation profiles are trapezoidal in shape, ramping up to prevent spastic reaction in the triceps surae and down to provide smooth cessation of stimulation.

In this study, a trapezoidal stimulation profile is chosen, it starts when the mid-swing sub-phase is detected and lasts for a fixed amount of time; 600 *ms*, with a rise time of 100 *ms*, and a drop time of 200 *ms*, if IC occurs before the trapezoid ends then the muscle is no longer stimulated. The maximum stimulation current is set prior to the experiment by applying a continuously increasing current value until the wearer feels a clear dorsiflexion movement. Fig. 4 shows the behaviour of both actuators during one gait cycle. A maximum current of 16 and 24 *mA* was enough to achieve sufficient dorsiflexion for subject 1 and 2 respectively. The motor value was set to 20% AAFO's maximum torque for plantar-flexion movement during pre-swing. During terminal stance, a 16% AAFO's maximum torque was set. These values are normalized using the maximum torque value (Pre-Swing).

IV. Experimental Setup

The experimental setup consists in using two actuators; the AAFO and the FES. The AAFO is an actuated orthosis (AJ-1000, SG Mechatronics) powered by an electrical motor with embedded rotational encoders at the ankle joint and two FSR embedded in the insole. It incorporates a FPGA (NI myRIO-1900), Li-Po batteries and an adjustable frame. It has a weight of 2.35 *Kg* not taking neither the batteries nor the FPGA into account. Fig. 1 shows the ankle joint robot and the FES stimulation electrodes. The RehaStim portable stimulation device (HASOMED) is used in this study. The current-controlled 8-channel stimulator includes two independent current sources, which are multiplexed to 4 outputs each. The output current ranges from 0 to 126 *mA* with 2 *mA* step and could be configured with a pulse width from 0 to 500 μs with 1 μs step. Besides, an extra insole with embedded force sensors is used for the right foot.

Non-assisted and assisted scenarios were applied during the experimentation conducted with healthy subjects. The non-assisted scenario was implemented by turning off both the AAFO's and the FES. Note that the non-assisted scenario was applied to take into account the effects of the AAFO's inertia and frictions on the overall impedance. In the assisted scenario, both AAFO and FES are activated. During both scenarios, the wearer walked at a constant speed of 2 *Km/hr* on a treadmill. For each trial, data were acquired at a sampling rate of 1 *KHz* from FSR sensors, the relative encoder coupled to the motor, the motor's torque and the FES stimulation's current. During the experiments, a gait cycle was measured from the initial heel contact of one foot to the next initial heel contact of the same foot. All data were time normalised to 100% of the stride cycle. The ankle joint angle was resampled at 2000 samples for each gait cycle, so that each point represents 0.05% of the gait cycle. Afterwards, the ankle joint angle is averaged at each sample for all the strides.

V. Preliminary Results

The first evaluation of the proposed method was to test the accuracy of the gait phase detection algorithm. Table II shows the average sub-phase's duration percentages for the non-assisted (for subject 1, 37 strides analyzed; and for subject 2, 20 strides) and assisted (for subject 1, 37 strides analyzed; and for subject 2, 79 strides) scenarios, as well as a comparison with the sub-phase's duration percentages found in the literature [14]. For subject 1, 97.3% of the sub-phases were detected in the correct sequence in the non-assisted scenario, and 96.68% in the assisted one. For subject 2, 100% of the sub-phases were detected in the correct sequence in the non-assisted scenario, and 99.35% in the assisted one. It can be seen that every sub-phase has a significate duration, which is suitable for triggering the system's actuators as defined by the controller rules.

The beginning times percentages for each sub-phase, in proportion of the whole gait cycle, has a maximum relative error of 9.9% for the non-assisted scenario and 8.9% for the assisted one; both maximum errors were detected at the swing phase, when the algorithm is approximating the sub-phases based on the right foot's GRF. Also, the approximation of the terminal swing duration percentage has a significate error. However, this error does not affect the system performance since this sub-phase does not trigger FES actuation, but rather allows the stimulation's continuity.

Note that these results depend on the wearers correct gait progression, consequently a validating system could be implemented, i.e., the GAITRite computerized gait analysis system. It was found that the fuzzy logic is capable of detecting the gait cycle sub-phases with satisfactory rate when walking on a flat surface. However, more experiments should be done for different situations, such as uneven terrain, non-walking tasks, etc.

TABLE II. MEAN SUB-PHASES DURATION PERCENTAGES DURING ONE GAIT CYCLE [%±STANDARD DEVIATION]

	Non-Assisted		***Assisted***		***R.***
SP	S1	S2	S1	S2	
LR	12.79±0.13	10.99±0.15	13.99±0.17	13.60±0.15	10
MSE	9.96±0.14	13.73±0.14	7.85±0.79	8.15±0.31	10
MSL	10.29±0.14	7.67±0.15	14.08±0.3	7.16±0.47	10
TS	19.94±0.11	16.23±0.11	15.73±0.14	19.04±0.14	20
PS	7.58±0.27	9.24±0.12	9.82±0.32	10.67±0.15	10
AIS	9.19±0.12	8.71±0.12	11.27±0.21	7.85±0.17	13
AMS	6.09±0.92	10.12±0.12	11.29±0.26	13.31±0.16	14
ATS	24.23±0.12	23.43±0.16	17.37±0.19	20.34±0.15	13

The preliminary results show that the gait detection algorithm provides reliable sub-phase detection results when applied to healthy subjects. When applying the proposed approach to paretic patients, the fuzzy logic inference rules could be adapted by adding pathology specific FMV definitions. It is worth reminding that only the terminal stance, pre-swing, mid-swing and terminal swing sub-phases are of interest to the controller. Consequently, only the pathologies that could affect these sub-phases should be taken into account, i.e., IC not happening with heel strike but with toe strike instead.

The mean ankle joint angle ranges are presented in Fig. 5 along with their standard deviations. It can be noted that the ranges are bigger throughout all the gait cycle in the assisted scenario, proving its effectiveness to influence the ankle angle. Also, during the terminal swing sub-phase, the angle is influenced by the activation of the FES, proving as well its effectiveness since the toe remains risen during the duration of this sub-phase as opposed to the behavior with the non-assisted scenario. Furthermore, the behavior of the ankle joint's angle in each sub-phase corresponds to the one expected, i.e., having a relative maximum during the terminal stance and terminal swing, and a relative minimum in the sub-phases loading response and initial swing.

VI. CONCLUSION

In this paper, we show the feasibility of achieving detection of the gait cycle sub-phases using FSR sensors and a fuzzy logic based algorithm. Experiments were conducted with two healthy subjects and show that by controlling the AAFO, the powered plantar-flexion movement was enhanced, resulting in a forward foot impulse prior to starting the swing phase. Furthermore, the activation of the FES during the last two sub-phases of the swing phase resulted in a longer dorsiflexed profile of the ankle joint angle trajectory. While studies have highlighted mechanisms of dropped foot impairments and positive effects of AAFO and FES devices independently, few studies have addressed the effect of a hybrid approach applied to patients suffering from foot drop. A better understanding of how AAFO/FES devices could affect the gait performance of individuals with different underlying dropped foot mechanisms will advance the success of the AAFO/FES design.

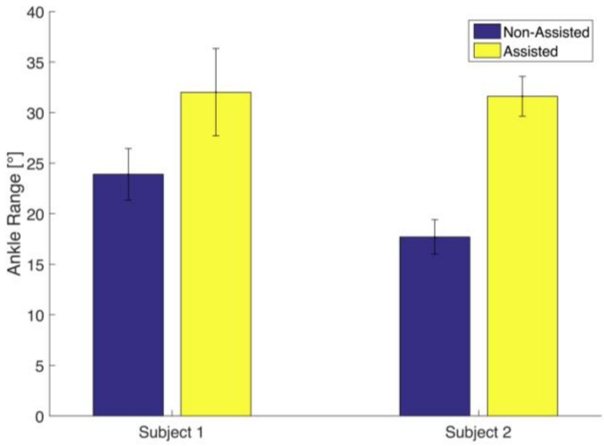

Figure 5. Mean ankle joint angle ranges per subject for the assisted and the non-assisted scenarios. The lines on each bar represents the standard deviation.

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