A Rule-Based Control System for Assisting the Gait by Multichannel Functional Electrical Stimulation: Design and Proof of Concept

Eleonora Vendrame, Dejan B. Popović, Emilia Ambrosini, Lana Popović-Maneski

Abstract—**A rule-based controller to assist stroke survivors' gait through multichannel functional electrical stimulation was developed. The controller was designed for the system comprising a wearable four-channel stimulator and insoles instrumented with pressure sensors and inertial measurement units. The gait segmentation algorithm processes Real-time signals from the sensorized insoles. The heuristically established rules split each gait cycle into five phases. These *If-Then* rules trigger stimulation channels to activate paralyzed thigh and shank muscles in an order that leads to the natural like stance and swing. The stimulator sends electrical pulses via surface electrodes positioned on a hemiplegic patient's paretic leg to activate the quadriceps, the hamstrings, the anterior tibialis, and the gastrocnemius muscles. A graphic user interface was developed to set the stimulation parameters and calibrate the system. The segmentation algorithm was validated on the recordings of 10 stroke patients, and the assistive gait training system was tested on one older adult to prove the concept.**

Index Terms—**Rule-Based Control; Multichannel Functional Electrical Stimulation; Post-Stroke Gait Restoration.**

I. INTRODUCTION

80 percent of stroke survivors develop motor disability and experience problems to walk. Restoring functions after stroke is a complex process mediated by neuroplasticity induced by spontaneous recovery and therapeutic interventions [1]. Early motor training seems essential for successful recovery: motor learning mechanisms may be operative during spontaneous stroke recovery, and by interacting with a rehabilitative exercise, they can be reinforced [2]. Functional Electrical Stimulation (FES) has been used in the rehabilitation of chronic hemiplegia since the 1960s when the first applications for drop-foot correction were patented [3]. During the following years, many improvements were suggested in terms of synchronizing the stimulation with the gait events and the number of stimulated muscles. The first clinically applied FES systems for restoration of locomotion used an open-loop control method, in which stored sequences of muscle activation were associated with the phases of a normal gait cycle [4]. The main drawback of this control was the lack of adaptation of the stimulation to the inevitable walking pattern fluctuations. The microcontroller technology and the enhancement of the gait events detection sensors led to the development of more effective control strategies. The rule-based control (RBC) aims to replace the manual FES-switching function through the automatic detection of the gait phases [5]. This control method is based on *If-Then* expressions: *If* part of the rule corresponds to the real-time recognition of a coded sensor pattern and represents the system*'s state*, *then* part of the rule triggers the functional movement correspondent to the state identified. The real-time recognition of the gait events is usually performed by Gait Phase Detection (GPD) systems that rely on the signals from artificial sensors. Nowadays, wearable sensors can effectively be used for gait segmentation. Foot pressure insoles or footswitches represent the gold standard in gait segmentation since each gait phase can be associated with a specific sensor output [6]. Alternatively, inertial measurement units (IMU) that comprise accelerometers, gyroscopes, and magnetometers are widely used to feed gait phase discrimination algorithms. The signals from footswitches and pressure/force sensors allow identification of the gait events accurately (e.g., heel contact, push-off). Still, they cannot discriminate against the swing phase's sub-phases. This limitation can be eliminated by adding inertial sensors to provide information about the kinematics of the movement [7]. Kojović et al. [8] presented an automatic control for an FES system based on *If-Then* rules designed by mapping sensors, and muscle activated patterns. The sensor system included accelerometers and force-sensing resistors.

We present here a new rule-based controller for a multichannel electronic stimulator to assist gait in stroke survivors.

II. MATERIALS AND METHODS

The experimental setup includes the multichannel MOTIMOVE modular functional electrical stimulation system which allows distributed and asynchronous stimulation (compensated biphasic pulses, 1-100 pulses, 50-1000 µs, up to 170 mA) [9], a set of surface electrodes for four muscle groups and the Gait Teacher, a sensorized insole for the acquisition of ground reaction forces and foot kinematics [10] (Fig. 1). The MOTIMOVE controls current pulses on four stimulation channels and receives six analog inputs from the Gait Teacher (sampling rate 100 Hz). Four input signals come from the insole worn on the paretic side: three ground reaction forces estimated from five pressure

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sensors and the foot's angular rate in the sagittal plane (from the gyroscope) (Fig. 2). Two input signals come from the non-paretic side and are used for the assessment of the gait pattern.

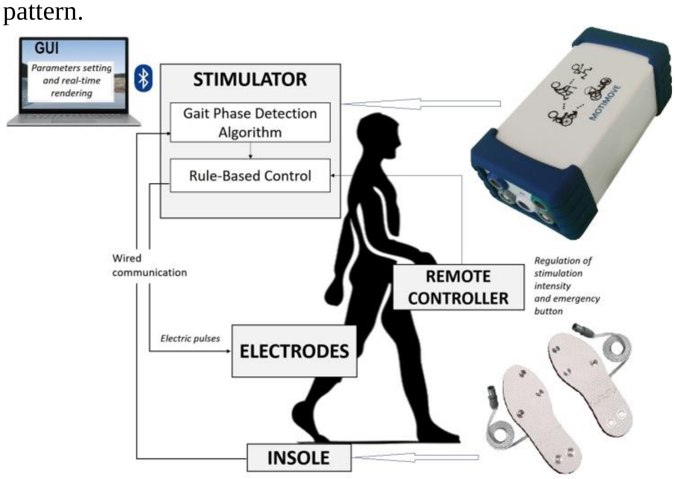

Fig. 1. Setup used for the design and proof of concept of the rule-based controller for FES assisted gait. MOTIMOVE is an up to eight-channel smart electronic stimulator. The Gait Teacher is a set of two instrumented insoles with pressure transducers and inertial measurement units for reproducible, real-time, no hysteresis acquisition of ground force reaction and foot kinematics. The system uses surface electrodes.

A. *Gait Phase Detection Algorithm*

Starting from the signals coming from the Gait Teacher, we implemented a Gait Phase Detection (GPD) algorithm for the real-time detection of gait sub-phases during walking. This algorithm detects five transition events (T1, T2, T3, T4, T5), which define five gait sub-phases corresponding closely to Terminal Swing (TS), Heel Contact (HC), Mid Stance (MS), Push-Off (PO) and Initial Swing (IS), respectively (Fig. 2).

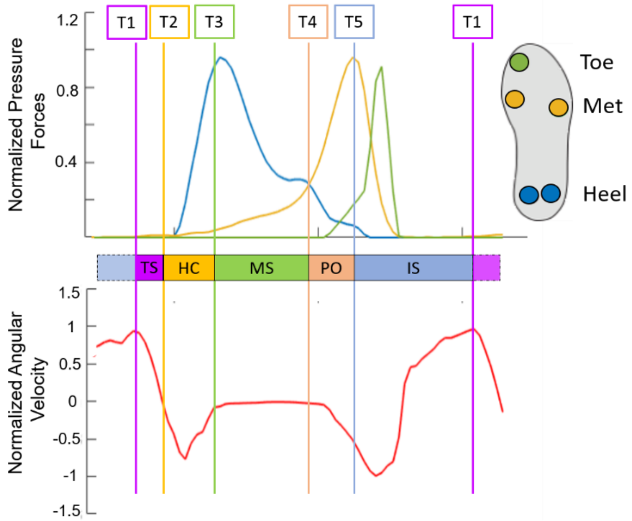

Fig. 2: Gait Teacher recordings from one gait cycle: the upper panel shows the normalized pressure forces on heels, metatarsals, and toe. Outputs from two sensors on the metatarsals (yellow) and two sensors on the heel (blue) are connected in parallel. The bottom panel shows the normalized angular velocity of the foot in the sagittal plane. Transition events T1, T2, T3, T4, and T5 are described in the text (TS-Terminal Swing, HC-Heel Contact, MS-Mid Stance, PO-Push-Off, IS-Initial Swing).

Each transition event is detected by the fulfillment of determined conditions, as reported in Table 1. The algorithm uses the signals coming from the insole (worn on the paretic side) and threshold values that are patient-dependent and are set during the calibration procedure.

TABLE 1. CONDITIONS FOR DETECTING THE TRANSITION EVENTS T1, T2, T3, T4, T5. Ω IS THE ANGULAR RATE IN THE SAGITTAL PLANE, PHEEL, PMET, AND PTOE ARE THE PRESSURES ON HEELS, METATARSALS, AND TOE. PHEEL_THRS, PMET_THRS, AND Ω_THRS ARE THRESHOLD VALUES.

T1	ω = max & *PMet* < PMet_Thrs & *PHeel* < PHeel_Thrs
T2	Zero-crossing of ω from positive to negative values
T3	ω < ω_Thrs & (*PHeel* + *PMet*) > (PHeel_Thrs+ PMet_Thrs)
T4	Intersection of *PHeel* with *PMet*
T5	(*PMet* = max ‖ *PToe* = max ‖ ω = min) & *PHeel* < PHeel_Thrs

B. *Rule-Based Control*

We implemented the GPD algorithm as a state machine, in which each gait phase defines a state, and conditions from Table 1 define transitions between the states. The stimulation starts when the event T4 is detected for the first time. Transitions are allowed only between consecutive states. This constraint leads to the fact that, if the algorithm misses a transition event, the stimulation remains in the current state until the same transition occurs in the following step. To avoid this mechanism, we set a time constraint on each phase duration: if a phase's duration is higher than a set time, the system enters an idle state with no stimulation, waiting for the beginning of the following step (e.g., event T4). This time constraint depends on the subject's walking speed, and the therapist chooses it among pre-selected values during the calibration of the system. The rule-based control of the FES-assistive gait training system is based on *If-Then* expressions, where the *If* statement verifies the current gait sub-phase. In contrast, the *Then* statement activates the corresponding muscular group. Thus, once the *If* the condition of the rule-based control is satisfied (i.e., the GPD algorithm determined the current phase of the system and the phase duration is not higher than the pre-set time constraint), the stimulator can generate and send the electric pulses to the corresponding channels. Each channel is connected to a surface electrode positioned on the bulk of the correspondent muscle. The muscular groups involved in the stimulation are the quadriceps (vastus lateralis and rectus femoris) as knee extensor, the hamstring as knee flexor, the tibialis anterior as dorsal flexor, and the gastrocnemius as a plantar flexor. The stimulation pattern is shown in Fig. 3.

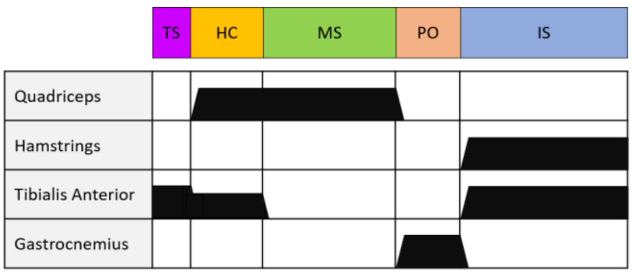

Fig. 3. Basic stimulation pattern for one gait cycle.

C. Graphic User Interface

We developed a Graphic User Interface (GUI) to allow the therapist to easily interact with the assistive gait system (Fig. 4). The GUI can directly communicate with the stimulator, sending and receiving data, and allows setting the system without any external assistance. It allows us to calibrate the insole, remove the offsets from the sensors, and set the amplitude of the stimulation for each muscle depending on the patient's motor and pain threshold. The user can enable the desired channels and select the basic stimulation parameters, such as the frequency and the pulse width of the stimulation. Also, from this interface, the pressure and angular velocity signals' thresholds can be manually set for each patient. Finally, the basic stimulation pattern can be modified according to patient needs. At the end of the calibration session, all the parameters can be saved and stored for the following sessions. The first calibration session lasts about 25 minutes, while the subsequent sessions' setting time is about 5 minutes.

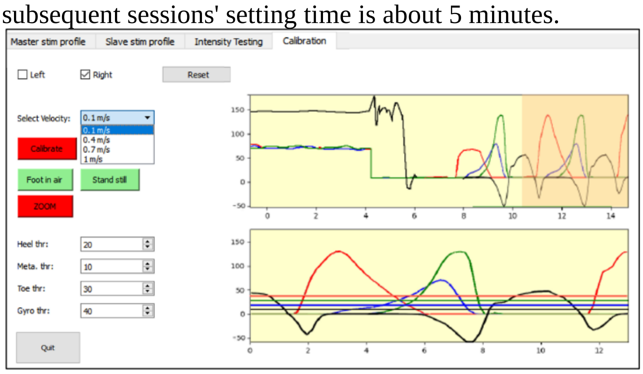

Fig. 4. Insole Calibration Tab of the GUI.

D. Gait Phase Detection Algorithm Validation

The Gait Phase Detection algorithm was validated using the recordings in ten chronic stroke patients to assess their efficiency and robustness. We implemented a second algorithm to perform a precise offline gait segmentation. The outputs of the online and offline algorithms were compared, and the time differences of the same events defined the time errors (Fig. 5). The time error in estimating each phase was computed both for the paretic and non-paretic limb for all the subjects. The analysis was carried out in both intra-subject (across 50 steps) and inter-subject (averaged across subjects). Due to the small sample size, we used a non-parametric statistical test to assess differences between online and offline detection; particularly, the Wilcoxon signed-rank test was employed.

E. Proof of Concept of the FES-Assistive System

Finally, we tested the FES-assistive gait training system on a 70-year-old man without any known motor disability. The gait pattern of older adults is similar to the gait of stroke patients, except that there is no drop-foot, and the symmetry between the legs is high. This test aimed to prove that the stimulation does not prevent normal gait and does not lead to instability. Three different modalities were tested: no stimulation, one single stimulation channel (tibialis anterior only) replicating the typical foot-drop stimulation, and 4-channel stimulation (quadriceps, hamstrings, tibialis anterior, and gastrocnemius). We compared gait characteristics as cadence and gait phases' duration between the different stimulation modalities. Moreover, we performed the analysis of signals oscillations: they were quantified by applying a moving-average low pass filter (20 samples) on the signals for the three tested modalities and computing the difference between the filtered signal and the original ones. Again, we chose a non-parametric statistical test: in this case, we used the Mann-Whitney U test.

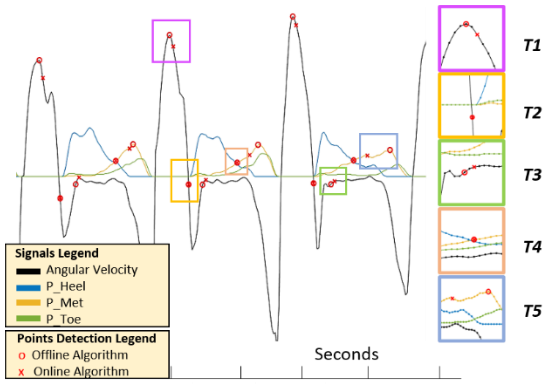

Fig. 5. Example of gait segmentation performed by the offline and the online algorithms on three gait cycles of a stroke patient. On the right, the detected events are zoomed.

III. Results

A. Gait Phase Detection Algorithm Validation

Signals coming from the healthy gait pattern are smoother and more regular than signals acquired from stroke patients. The GPD algorithm was tested on data collected in ten chronic stroke patients. The algorithm detected 100% of the transition events on an average of 50 steps per patient. A higher walking disability characterized three out of ten patients in terms of gait speed and asymmetries between the limbs. The time error distribution computed across each patient for both the limbs showed a low variability in detecting all the phases, except for the most impaired subjects.

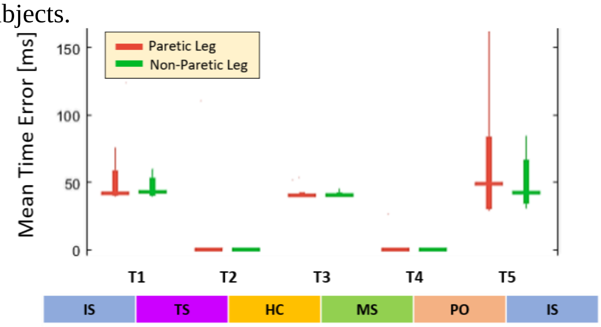

Fig. 6. Meantime error distribution across ten stroke patients in each transition event. The horizontal bar of the box plot represents the median of the population. The bold line extends from the first to the third quartile. The fine line goes from the minimum to the maximum value, excluding the outliers.

For them, the paretic limb was characterized by a large error variability, especially in detecting the beginning of the Terminal Swing (T1) and the beginning of the Initial Swing (T5). The time error was averaged across the steps of each patient to assess the variability among patients. Similarly, this analysis showed a very low variability of the time error for all the phases except for the T5 transition. Moreover, the time error distribution showed the existence of an intrinsic error due to the real-time nature of the algorithm: transitions T1, T3 and T5 had a median delay of 40ms, since the conditions for the detection of these events are based on the derivative of the signals computed over five consecutive points (Fig. 6).

B. FES-Assistive System Proof of Concept

The current amplitude values were set at: 68 mA, 59 mA, 78 mA, and 68 mA for the quadriceps, hamstrings, tibialis anterior and gastrocnemius channels, respectively. The stimulation frequency was set to 40 pulses per second, and the pulse duration was 350µs. First, the cadence did not change significantly between the three stimulation modalities, although a small increase occurred in 4-channel stimulation (No stimulation, 81.1 steps/m; Tibialis Anterior stimulation, 80.1 steps/m; 4-channel stimulation, 84.1 steps/m). We evaluated the angular velocity and heel pressure force signals: in the no stimulation modality, they were characterized by more significant oscillations than in the stimulation modalities. For the angular velocity pattern, we found a statistically significant difference between the modalities with and without stimulation. In contrast, for the heel pressure force, we found a significant difference between all the modalities (Fig. 7). Finally, the phases' duration analysis showed a non-significant difference between the modalities, except for the Terminal Swing and Heel Contact phase duration. The reduction of oscillations probably reduced the Terminal Swing phase duration and increased the duration of the Heel Contact phase.

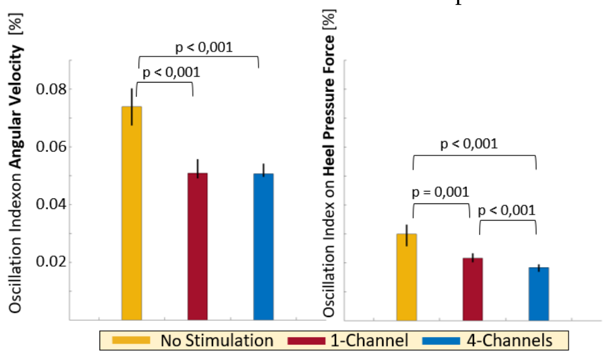

Fig. 7. Oscillation index computed on the angular velocity and the heel pressure force signals across ten gait cycles for the healthy subject. The bar value is the median of the index, and the vertical black line represents the first and third quartile of the distribution. For the statistical analysis, the Mann-Whitney U test was used.

IV. Conclusion

In the framework of FES-assistive systems for stroke patients' rehabilitation, we present a new controller for a multichannel electronic stimulator to assist gait in stroke survivors. The novelty of the proposed control system relies on combining the accurate and reliable gait information provided by the Gait Teacher insole with the modularity of the MOTIMOVE stimulator, which creates an exceptional wearable and easy-to-use solution to assist persons with hemiparesis to recover motor functions after stroke.

The Gait Teacher's signals are online processed by a GPD algorithm to split gait into five phases corresponding closely to: Terminal Swing, Heel Contact, Mid Stance, Push-Off, and Initial Swing. Five transition events, T1-T5, identify these phases. This algorithm was validated on the recordings of ten chronic stroke patients. T2 and T4 are characterized by negligible time delays, while T1, T3, and T5 detections are affected by an intrinsic time delay (40 ms) due to the algorithm's real-time nature. Nevertheless, the GDP algorithm was able to detect all the phases across all the steps of each stroke patient. The validation results were in line with the timing and accuracy requirements that the stimulation demands. The proof of concept performed on a healthy subject showed the stimulation pattern did not counteract the physiological muscle activity and seemed to work in synergy.

Moreover, some oscillations, which characterized the angular velocity signal and the heel pressure force signal, were reduced by the stimulation of the tibialis anterior during the swing phase of the gait cycle and the quadriceps muscle during the Heel Contact phase, respectively. Finally, the system's calibration from the GUI allows partial customization of the system for each subject. The gait parameters sent to the GPD algorithm before the stimulation permitted session to set the proper thresholds of the pressure and angular velocity signals. Moreover, the possibility to store the data of each patient after the first session reduces the setting time to only 5 minutes. The next step is to perform a feasibility study on stroke patients to evaluate the FES-assistive system usability in a clinical environment. Further improvements will involve combining the information coming from both insoles, allowing the implementation of more flexible and efficient rules. Moreover, the automatization of the calibration and the online adaptation of the parameters would reduce the setting time and limit the error due to a manual selection of the signal thresholds.

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V. Acknowledgment

This study was financially supported by the Ministry of Education, Science and Technological Development of the Republic of Serbia, Agreement with the Institute of Technical Sciences SASA, Grant No. 451-03-68/2020-14/200175 and the Grant F-137 from SASA.