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Thoughts and Progress

Design of a Symmetry Controller for Cycling Induced by Electrical Stimulation: Preliminary Results on Post-Acute Stroke Patients

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Abstract: This study deals with the design of a controller for cycling induced by functional electrical stimulation. The controller will be exploitable in the rehabilitation of hemiparetic patients who need to recover motor symmetry. It uses the pulse width as the control variable in the stimulation of the two legs in order to nullify the unbalance between the torques produced at the two crank arms. It was validated by means of isokinetic trials performed both by healthy subjects and stroke patients. The results showed that the controller was able to reach, and then maintain, a symmetrical pedaling. In the future, the controller will be validated on a larger number of stroke patients. **Key Words:** Control systems—Cycling—Electric stimulation therapy—Rehabilitation—Stroke.

In industrialized countries, stroke represents the first cause of long-term disability. Thus, the demand for new rehabilitation treatments able to improve and accelerate motor recovery is increasing. Clinical studies on neuroplasticity support the role of goal-oriented, repetitive movements in the motor relearning process (1). Besides, exercises induced by functional electrical stimulation (FES) provide the patients afferent feedback to the central nervous system (CNS); the term afferent feedback applies to the whole sensorial information reaching the CNS from the periphery. Thus, goal-oriented, repetitive movements induced by FES could furthermore facilitate the reorganization of motor schemes (1).

Riding an ergometer is often used in the lower limbs rehabilitation of stroke patients. Cycling involves most of the leg muscles and can be safely performed from a wheelchair even before gait training is possible. The application of FES via surface electrodes synchronized to the pedaling may enhance the rehabilitation progress. Over the past 20 years, FES cycling has become an established method in the rehabilitation of spinal cord-injured patients (2), while only recently, it has been used in the treatment of stroke patients (3,4). Because of the laterality of the pathology, recovery of motor symmetry is crucial for these patients (5). Therefore, the aim of this study was the design of an automatic controller for FES cycling that is able to adjust the stimulation parameters in real time, in order to compensate a possible unbalance between the contributions of the two legs.

MATERIALS AND METHODS

Design of the symmetry controller and stability analysis

The controller structure consists of two parallel branches (Fig. 1): Each system represents one leg, whose input is the pulse width, PW, used to stimulate the leg and whose output is the torque, T, produced at the crank arm. Thanks to the isokinetic training mode, the left and right active torque, $T_{a,L}$ and $T_{a,R}$, are computed as the difference between the torque generated by stimulation and the one obtained during passive cycling, that is, when the legs are driven only by the motor. Then, for each revolution, $T^*_{a,L}$ and $T^*_{a,R}$ are computed by averaging $T_{a,L}$ and $T_{a,R}$ over the crank angle ranges, in which the muscles of each leg are stimulated. Afterward, comparing $T^*_{a,L}$ and $T^*_{a,R}$, the error signals, e_L and e_R, are defined. The rationale of the controller is to stimulate as much as possible the weaker leg until a maximum value, PW_{max}, and then, if an unbalance is still present, to decrease the PW of the stronger leg, until symmetry is reached. Finally, two integral controllers (IC) use the error signals to update the PWs, according to Eq. 1:

$$PW_i(\mathrm{k}) = PW_i(\mathrm{k}-1) + \mathrm{K_C}\, e_i(\mathrm{k});\ \mathrm{i} \in \{L, R\}, \quad (1)$$

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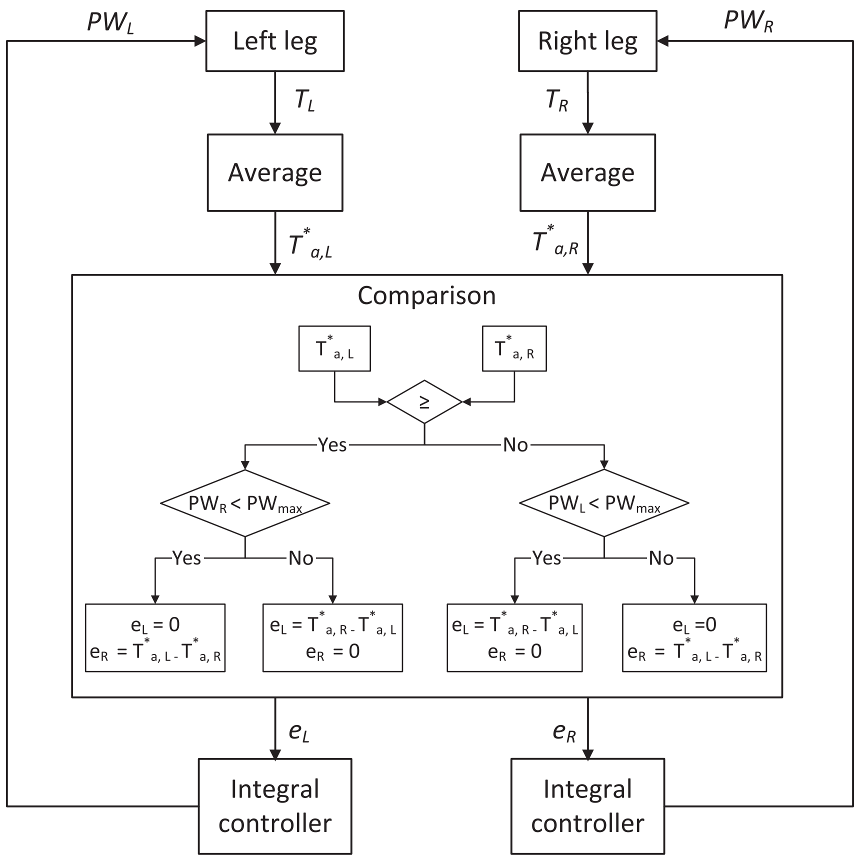

FIG. 1. Block scheme of the symmetry controller. $PW_{L/R}$, left/right pulse width; $T_{L/R}$, left/right torque; $T^*_{a,L/R}$, left/right active torques, averaged in the angular stimulation ranges; $e_{L/R}$, left/right error signals.

where k is the revolution index and K_C is the integral gain.

The *PW*s are updated once per revolution when the crank angle is equal to 0° (conventionally, 0° corresponds to the crank angle in which the left hip is maximally flexed). The ICs are implemented with an integral anti-windup design so that PW_i is constrained between 0 and PW_{max}.

A stability analysis was carried out to tune K_C. Because the two ICs are defined by the same K_C, only one branch of the controller structure was considered. To perform a linear stability analysis, the output (PW_i) was assumed not to be saturated (i.e., $0 < PW_i < PW_{max}$) and the plant was approximated by a discrete-time linear system described by the form $q^{-1}K_p$, with q^{-1} being the delay operator. The gain K_p was fixed to a value of $4.6 \times 10^{-3} \pm 0.05 \times 10^{-3}$ Nm/µs by means of trials performed by three healthy subjects. K_C was tuned according to the pole placement method:

$$H(q^{-1}) = \frac{K_P K_C q^{-1}}{1 + (K_P K_C - 1) q^{-1}}. \qquad (2)$$

To guarantee stability and a well-damped system response, the pole of $H(q^{-1})$ has to be real positive and inside the unit circle. Thus, the upper limit of K_C was set to 200 µs/Nm. However, K_C was fixed to 50 µs/Nm to provide sufficient stability margins in case of fluctuations of K_P and to yield a good trade-off between noise sensitivity and achievable closed-loop bandwidth. Moreover, this permits to adjust the stimulation gently, increasing the convergence time but improving the subject's acceptability.

Experimental setup

The experimental setup includes a current-controlled eight-channel stimulator (RehaStim, Hasomed GmbH, Magdeburg, Germany) and a motorized cycle ergometer. The latter is equipped with a shaft encoder to measure the crank angle and with resistance strain gauges on the crank arms to provide the right and left torques produced during pedaling. The torque's uncertainty is 0.029 Nm (6). A personal computer running Matlab/Simulink (The MathWorks, Natick, MA, USA) under Linux is used for data acquisition and control at a sampling period of 5 ms.

Subjects and protocol

The controller was tested on two healthy subjects (S1, S2) (different from the three healthy subjects involved in the trials to tune K_P) and three post-acute

TABLE 1. *Details on the subjects and the amplitude of currents used in the experimental trials*

Subject	Sex	Age	Ictus origin	Paretic side	Months post-ictus	Currents (mA)			
						LQ	LH	RQ	RH
S1	F	25	—	—	—	30	30	30	30
S2	F	30	—	—	—	30	30	30	30
P1	M	61	Ischemic	Left	2	55	60	55	60
P2	F	30	Hemorrhagic	Left	5	45	55	40	50
P3	M	51	Ischemic	Left	4	45	45	45	45

LQ, left quadriceps; LH, left hamstrings; RQ, right quadriceps; RH, right hamstrings.

stroke patients (P1, P2, P3) collaborative and trained to FES. Details on the subjects are given in Table 1. The protocol was approved by the Ethical Committee of the Valduce Hospital and all the subjects signed a written informed consent.

Quadriceps and hamstrings of both legs were stimulated by means of surface self-adhesive 40×90 mm electrodes (Medical 95 s.n.c., Nova Milanese, Milano, Italy). The stimulation frequency was fixed to 20 Hz, and the currents, reported in Table 1, were set individually on each muscle to a value tolerated by the subject, which produced a visibly good muscular contraction with the maximum value of PW (PW_{max} is between 400 and 500 µs). No ramps were used in the PW profile. Each trial lasted 3 min: 60 s of passive cycling and 120 s of stimulation according to the crank angle ranges reported in Table 2. During the whole trial, the subject was asked not to pedal voluntarily, and the cadence was maintained constant at 25 rpm by the motor inside the ergometer.

Data analysis

To verify the unbalance during the passive phase, the passive unbalance index (PUI) was computed by normalized cross-correlation of the phase-aligned left and right torque profiles at time lag 0. Values of PUI close to 1 indicate a good level of balance between the two legs during passive cycling.

The results of the trials were analyzed in terms of unbalance, computed as follows:

$$\Delta T_a^{(n)} = T_{a,R}^{*(n)} - T_{a,L}^{*(n)} \tag{3}$$

where n is the revolution index.

To evaluate the symmetry achievement, an average value of $\Delta T_a{}^{(n)}$ over five revolutions was considered with its standard deviation, as follows:

$$\Delta T_{a,S}^{(n)} = \frac{1}{5}\sum_n^{n+5} \Delta T_a^{(n)} \pm \sqrt{\frac{1}{4}\sum_n^{n+5}\left(\Delta T_a^{(n)} - \frac{1}{5}\sum_n^{n+5}\Delta T_a^{(n)}\right)^2} \tag{4}$$

Symmetry is assumed to be achieved when $\Delta T_{a,S}{}^{(n)}$ is less than 0.1 Nm. Finally, also, the time needed to reach symmetry, Δt, was calculated.

TABLE 2. *Angular stimulation ranges*

	LQ	LH	RQ	RH
Start angle	0°	100°	180°	280°
Stop angle	100°	220°	280°	40°

0° corresponds to maximum flexion of the left hip.
LQ, left quadriceps; LH, left hamstrings; RQ, right quadriceps; RH, right hamstrings.

RESULTS

Table 3 reports, for all the trials:

- the value of PUI;
- the initial unbalance $\Delta T_a{}^{(1)}$, that is, the unbalance at the first revolution in which the stimulation was switched on;
- the value of $\Delta T_{a,S}$ of the revolution in which symmetry was achieved (when symmetry was not accomplished, the value of $\Delta T_{a,S}$, given in Table 3, was the last computed);

TABLE 3. *Results of the experimental trials*

Subject	PUI	$T_{a,R}{}^{*(1)}$ (Nm)	$T_{a,L}{}^{*(1)}$ (Nm)	$\Delta T_a{}^{(1)}$ (Nm)	$T_{a,R}{}^{*(n)}$ (Nm)	$T_{a,L}{}^{*(n)}$ (Nm)	$\Delta T_{a,S}{}^{(n)}$ (Nm)	Δt (s)
S1	0.987	0.25	0.82	–0.57	1.09	1.09	–0.07 ± 0.08	24
S2	0.988	0.67	0.29	0.38	0.52	0.47	0.03 ± 0.04	23
P1	0.975	0.27	0.03	0.24	0.42	0.42	–0.01 ± 0.08	77
P2	0.992	1.35	0.72	0.63	0.95	0.86	0.10 ± 0.08	54
P3	0.987	0.71	0.06	0.65	0.23	0.11	0.12	—

PUI, passive unbalance index.

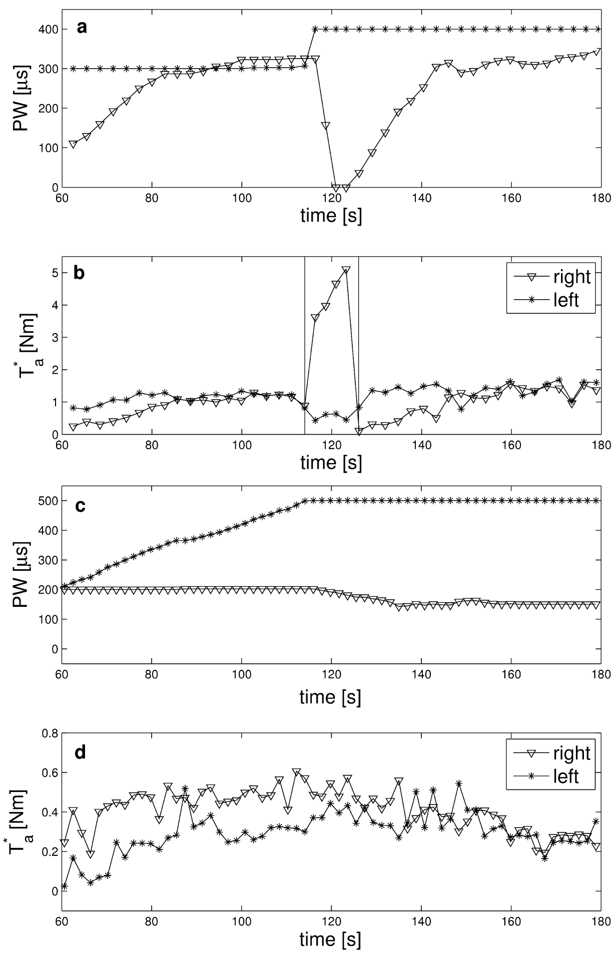

FIG. 2. Panels (a) and (b) show the trial performed by S1; profiles of PW_L and PW_R are reported in panel (a) and those of $T^*_{a,L}$ and $T^*_{a,R}$ in panel (b). Panels (c) and (d) refer to P1; profiles of PW_L and PW_R are reported in panel (c) and those of $T^*_{a,L}$ and $T^*_{a,R}$ in panel (d). Only the phase in which the stimulation was switched on is reported (60–180 s).

- the time Δt needed to reach symmetry, when possible.

In all the trials, the value of PUI was close to 1, indicating that passive cycling was balanced.

Figure 2a,b shows the results obtained by S1. The initial values of PW_L and PW_R were set to 100 and 300 µs, respectively, in order to make the pedaling unbalanced. Thus, at the beginning of the trial, the left leg was stronger and the initial unbalance was –0.57 Nm. Accordingly, the controller increased PW_R, reaching symmetry after 24 s of stimulation. Then, to test the robustness of the controller, the subject was asked to pedal voluntarily only with the right leg: $T^*_{a,R}$ increased and, thus, the controller first increased PW_L to the maximum value (400 µs) and then decreased PW_R to zero, in the attempt of reducing the unbalance. When the subject stopped pedaling voluntarily, $T^*_{a,R}$ suddenly decreased to zero and the controller increased PW_R until regaining symmetry. Similarly, S2, starting from an unbalance of 0.38 Nm, achieved symmetry in 23 s.

Figure 2c,d shows the results of the trial performed by P1. Both initial values of *PW* were fixed at 200 µs and the initial unbalance was 0.24 Nm. The controller nullified the unbalance after 77 s of stimulation. In the trial carried out by P2, symmetry was reached after 54 s of stimulation. Finally, in the trial performed by P3, symmetry was not achieved, but the unbalance was reduced from 0.65 to 0.12 Nm.

DISCUSSION

An innovative control system for FES cycling was developed. Torque sensors were mounted at the crank arms to provide the pedaling unbalance in real time. The controller was validated on healthy subjects and stroke patients. The results show that the controller was able to nullify the unbalance and maintain a symmetrical pedaling. In one trial (P3), the unbalance was only reduced, because of strong impairment, voluntary reactions against stimulation, spasticity, muscle tone increase, etc. From the collected data, the time needed to reach the steady state is less than 80 s. Afterward, the *PW* profiles showed only slight oscillations, as shown in the last 40 s of the trial reported in Fig 2d. Here, a slight decrease of the torques, which may be due to the onset of muscular fatigue, can be observed. Despite that, the controller maintained a symmetrical pedaling. Initial unbalance and the patient's level of impairment seem to have an influence on the time required to reach the steady state, that is, symmetry, when possible, or the most feasible reduction of unbalance. However, more trials are needed to quantitatively evaluate this relation.

As the reader can notice, torques produced by FES are low due to the limited stimulation amplitudes tolerated by the subjects and to the muscle atrophy in stroke patients. However, the major rehabilitative aim is to teach a symmetrical motor strategy, which is strictly correlated with the afferent feedback provided to the CNS by FES cycling therapy, regardless of the amount of produced torque.

For assumption, the gains of the two ICs were fixed to the same value. This setting is suboptimal, particularly for stroke patients, whose performance of the legs can significantly differ. In the future, a procedure to set automatically the controller parameters on each leg of the patient will be developed. Moreover, the controller will update not only the *PW*s but also the amplitude of currents in order to provide a larger control action range to achieve symmetry for strongly impaired patients. Finally, the initial stimulation parameters will be set automatically to make the daily application of the controller possible in clinics. Afterward, the controller will be tested on a larger number of stroke patients and a study to demonstrate its clinical benefits will be performed.

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