# SIMULATION AND EXPERIMENTAL DESIGN OF A SYMMETRY CONTROLLER FOR FES CYCLING OPTIMISED ON STROKE PATIENTS

Emilia Ambrosini, Simona Ferrante
*NEARLab, Bioengineering Department, Politecnico di Milano, via Golgi 39, Milano, Italy*

Thomas Schauer
*Technische Universität Berlin, Control Systems Group, Einsteinufer 17, Berlin, Germany*

Alessandra Pedrocchi, Giancarlo Ferrigno
*NEARLab, Bioengineering Department, Politecnico di Milano, via Golgi 39, Milano, Italy*

Keywords: FES cycling, Control systems, Stroke, Rehabilitation.

Abstract: This study deals with the design of a controller for FES cycling able to assure a symmetrical pedalling. This controller is exploitable in the rehabilitation of patients with unilateral neurological disorders which need to recover a symmetrical use of the legs. The controller updates the values of pulse width used in the stimulation of the two legs in order to nullify the unbalance between the torques produced at the cranks and then to maintain a symmetrical pedalling. The controller was tested first in simulation by means of a neuro-musculo skeletal model of a stroke patient which identifies the kinematic and dynamic of cycling induced by FES. After a stability analysis and an optimization of the controller tuning performed in simulation, the controller was tested experimentally on a healthy subject. The results of this trial show that the controller is able to reach a symmetrical pedalling in about 18 s, starting from an unbalance of 0.73 Nm. Furthermore, it is able to maintain a symmetrical task with small oscillations of the PWs. Thus, its employment in the rehabilitation of stroke patients could be crucial in the recovery of motor functions, such as walking, where a cyclical symmetrical motor task is required.

## 1 INTRODUCTION

Cycling induced by means of controlled functional electrical stimulation (FES) of the large leg-actuating muscles is an interesting method in the rehabilitation of patients with neurological or muscular disorders (Scheffler, 2007). We will refer to this artificial movement with the term FES cycling. In order to induce this task, superficial electrodes are placed over the target muscle groups: always over the quadriceps and the hamstrings, sometimes over the gluteus maximum, the plantar-flexors and dorsal-flexors of the ankle. Patients sit on the ergometer and the computer controls the appropriate sequence of stimulation of the muscles in order to obtain the cycling movement. A shaft encoder at the crank measures the crank angle; each muscle is stimulated for a particular range of the crank angle to turn the pedals. Usually, FES cycling system use an active drive mechanism (for example, an electric motor) to maintain a constant cadence. Indeed, iso-kinetic training devices allow a larger number of patients to undertake FES cycling, i.e., those unable to generate and maintain sufficient muscular force to rotate a flywheel or those with low tolerance to FES due to residual sensation.

FES cycling has become an established method in the rehabilitation of patients with Spinal Cord Injury (SCI) (Hunt, 2004; Faghri, 2005). Only recently it was demonstrated the importance of FES cycling on patients with stroke (Ferrante, 2008a). When patients do not have a complete spinal lesion



Ambrosini E., Ferrante S., Schauer T., Pedrocchi A. and Ferrigno G. (2009).
SIMULATION AND EXPERIMENTAL DESIGN OF A SYMMETRY CONTROLLER FOR FES CYCLING OPTIMISED ON STROKE PATIENTS.
In *Proceedings of the International Conference on Biomedical Electronics and Devices*, pages 245-250
DOI: 10.5220/0001548802450250

the employment of FES cycling could become crucial in re-learning the Central Nervous System the proper sequence of activation of the muscles involved in the task (Scheffler, 2007).

Designing a proper FES cycling controller presents several challenges. First, the physiological system to be controlled is strongly non-linear and time variant. Further, there is an high degree of uncertainty and variability in the response properties of the system. In this context, it would be helpful to design a neuro-musculo skeletal model able to identify the response of the subject to FES in order to increase the awareness about the system to control and to test the implemented controllers before performing experimental trials on patients. Another crucial aspect in the development of control strategies for FES cycling is the identification of the rehabilitative objective and, thus, the choice of the controller rationale; both these steps depend on the target pathology. In this contest, another essential step is the choice of the sensors needed on the ergometer in order provide a real-time controlled signal. Up to now only controllers for patients with complete SCI aimed at the maximization of the power output or at the minimization of muscular fatigue were developed (Hunt 2004; Hunt 2006). In those studies, the ergometer was equipped with a torque sensor at the crank and thus the power output could be chosen as the controlled signal (Hunt, 2008).

The aim of the study was the development of a control system for FES cycling optimised for the rehabilitation of patients with stroke. Because of the laterality of the pathology, the recovery of the motor symmetry is crucial in the rehabilitation of these patients. Thus, the controller, starting from a real-time measure of the unbalance between the torques produced at the two cranks, modifies the stimulation parameters of the two lower limbs independently in order to achieve a symmetrical pedalling. The implemented controller was first tested and validated by means of a simulation model and then some trials on healthy subjects were carried out.

## 2 THE DESIGN OF THE SYMMETRY CONTROLLER

The structure of the closed-loop system developed is shown in Figure 1. It includes two parallel branches in order to control two systems contemporaneously. Each system corresponds to a lower limb, whose input (control signal) is the value of pulse width, *PW*, used to stimulate the selected muscles of the leg and whose output (controlled signal) is the torque, *T*, produced at the crank. For each leg, the muscular groups included in the stimulation strategy are the quadriceps and the hamstrings, which are the muscles which mostly contribute to the pedalling (Ferrante, 2008b). The range of stimulation of each muscle in respect to the crank angle is set following (Ferrante, 2008a).

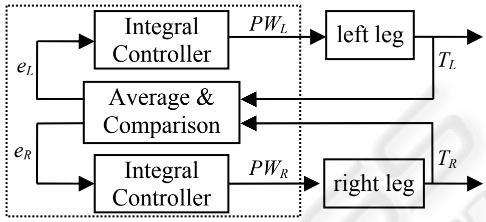


Figure 1: Block scheme of the symmetry controller; the dotted line includes the whole controller.

The *PW* delivered to each leg is defined by a pure integral controller (see Figure 1). The definition of the reference signal of each controller is not unique. Indeed, there are two possibilities to nullify the difference between the $T_R$ and $T_L$, which are the *T* produced at the right and left crank, respectively:

- increasing the *PW* of the weaker leg;
- decreasing the *PW* of the stronger leg.

In the design of the controller, it is chosen to stimulate as much as possible the weaker leg till the maximum value ($PW_{max}$). Then, if an unbalance is still present, the *PW* of the stronger leg is decreased. In order to take into account the unbalance only due to the stimulation, the two active *T*, $T_{a,L}$ and $T_{a,R}$, are computed for each leg. $T_{a,\,L/R}$ is the difference between the *T* produced when the stimulation is ON and the one obtained during passive cycling, i.e, when legs are driven by the motor at a constant speed. Then, $\overline{\boldsymbol{T}_{\boldsymbol{a,L/R}}}$ is computed averaging $T_{a,L/R}$ over the crank angle range during which the muscles of the left/right leg are stimulated. Finally, the error signals, $e_L$ and $e_R$, are defined by comparing $\overline{\boldsymbol{T}_{\boldsymbol{a,\,L}}}$ and $\overline{\boldsymbol{T}_{\boldsymbol{a,\,R}}}$, as shown by the flow diagram reported in Figure 2.

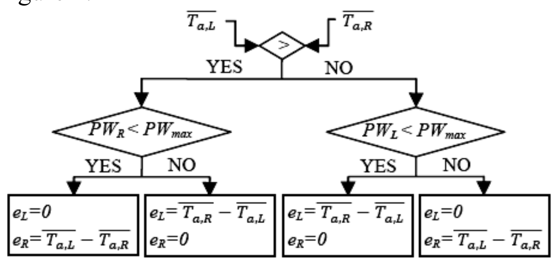


Figure 2: The flow diagram which describes how the error signals are computed.

The two integral controllers, shown in Figure 1, are parameterised only be the integral gain $K_C$, as described by equation (1)

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

The values of $PW_i$ are updated every revolution.

$K_C$ was set at the same value for both the legs and was defined according to the stability analysis performed by means of the simulation model. The integral controllers were developed with an integral anti-windup design so that the $PW_i$ is constrained between 0 and $PW_{max}$. The controller was implemented in Matlab.

# 3 STIMULATION PROTOCOL

A FES cycling protocol was defined both to perform the stability analysis and to test the working of the controller in simulation and in experimental trials. Each trial lasted 3 minutes: the initial 60 s were characterised by passive cycling while, during the last 120 s, FES started with the symmetry controller switched on. The patients was ask not to pedal voluntary at all. An electric motor maintained the angular velocity at a constant value of 20 rpm, during the whole trial.

# 4 SIMULATION

## 4.1 The Neuro-muscolo Skeletal Model for Cycling

To test the symmetry controller developed, a neuro-muscolo skeletal model which simulates a stroke patient pedalling by means of FES was designed.

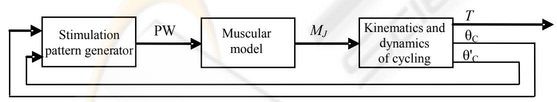


Figure 3: Block structure of the simulation model.

The block structure of the simulator is reported in Figure 3 and consists of three main parts:

1) *Stimulation Pattern Generator*, which defines the *PW*, the frequency and the crank angular ranges in which the stimulation has to be delivered to each muscle involved in the stimulation strategy.

2) *Muscular Model*, which calculates the joint moments, $M_J$, produced by muscular contractions. The model is inspired to a previous work (Riener, 1998). The muscle groups included in the model are: mono-articular hip extensors, hamstrings, biceps femoris-short head, rectus femoris and vasti. The maximum isometric forces of all the muscles of the right leg were set at the half of those of the muscular groups of the left leg. This permits to reproduce the muscular model of a stroke patient, with the right side impaired. The fatigue occurrence shows a decrease of the muscle activation to about 50% of its nominal value over 100 s of stimulation with a *PW* of 400 µs, comparable to (Abbas, 1995).

3) *Kinematics and Dynamics of Cycling*, which computes the crank angle, $\theta_C$, the cadence, $\theta'_C$ and the *T* produced by each leg at the cranks, starting from the $M_J$. The mechanical structure consists of a planar five-bar linkage (Figure 4). All the five links (B1 to B5) are assumed to be rigid and are connected by planar joints (J1 to J7), which correspond to the hips, the knees, the ankles and the crank shaft. The ankle joints coincide with the pedals and any rotation around these joints is forbidden; the positions of the two hip joints coincide and they are fixed as well as the crank axis. Thus, the entire system, has only one degree of freedom and can be fully characterised by $\theta_C$. The kinematics and dynamics were implemented using the Open Dynamics Engine (ODE). The complete model was developed in Matlab/Simulink.

More details on the simulator can be found in (Ambrosini, 2008).

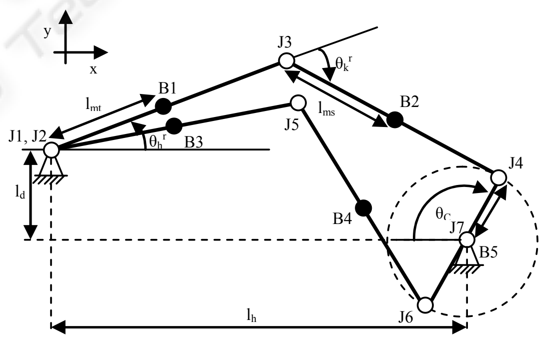


Figure 4: The five-bar linkage model. B1: right thigh, B2: right shank, B3: left thigh, B4: left shank, B5: crank arms; J1: right hip, J2: left hip, J3: right knee, J4: knee left, J5: right ankle, J6: left ankle, J7: crank shaft.

## 4.2 The Stability Analysis of the Controller

The stability analysis of the controller was carried out to tune $K_C$. Only one branch of the block scheme reported in Figure 1 was included in the analysis. The simplified system analysed is shown in Figure 5; it refers to the left leg.

In order to analyse only the region of linearity, it was assumed that the controller output ($PW_L$) was not saturated and the system was approximated by a linear transfer function $P(q^{-1})$, computed as follows:

$$P(q^{-1}) = K_P\, q^{-1} \quad (2)$$

where the only parameter is the gain $K_P$. Its value was estimated by means of the simulation model, setting the value of $PW$ at 400 µs, i.e., the value of $PW_{max}$, and calculating the consequent value of $\overline{T_{a,L}}$, produced. The value of $K_P$ was computed dividing this value of $\overline{T_{a,L}}$ by the difference between 400 µs and the threshold value of $PW$ over which the stimulated muscles start to produce an increase in the torque, i.e., 100 µs. The value of $K_P$ obtained was 0.0075 Nm/µs. It was chosen to analyse the stability of the system referred to the left leg because, in the model, the left side was the healthy one and, thus, its value of $K_P$ was the bigger.

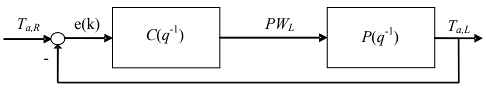


Figure 5: Simplified version of the closed-loop system. $P(q^{-1})$ and $C(q^{-1})$ represent the transfer function of the system and of the controller, respectively.

The transfer function of the integral controller $C(q^{-1})$ was computed as follows:

$$C(q^{-1}) = \frac{K_C}{1 - q^{-1}} \quad (3)$$

From equations (2) and (3), it was possible to calculate the closed-loop transfer function as:

$$H(q^{-1}) = \frac{K_P K_C q^{-1}}{1 + (K_P K_C - 1) q^{-1}} \quad (4)$$

The stability of a closed-loop system is guaranteed if the poles of $H(q^{-1})$ are inside the unit circle. Therefore, our system is stable if:

$$|\,1 - K_P K_C\,| < 1 \quad (5)$$

From equation (5), it resulted that the maximum value of $K_C$ to remain in the stability region is 260 µs/Nm. Furthermore, to avoid oscillations of the system output, the pole of the system should be real positive. Thus, the maximum value of the controller gain was fixed at 130 µs/Nm.

To verify the results of the simplified stability analysis, the controller was tested in simulation with different values of $K_C$. The results of the trials are reported in Figure 6; only the first 60 s in which the stimulation was on are shown. According to the stability analysis, the system was stable if $K_C$ was 100 µs/Nm (panels (a)-(d)), stable with some oscillations if $K_C$ was 150 µs/Nm (panels (b)-(e)), and unstable with $K_C$ of 300 µs/Nm (panels (c)-(f)).

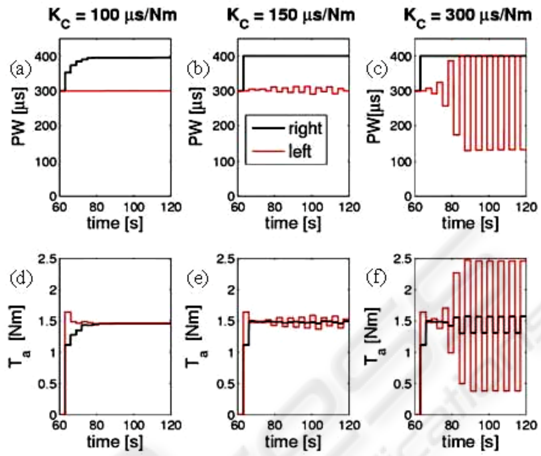


Figure 6: Panels (a)-(b)-(c): values of $PW_R$ and $PW_L$. Panels (d)-(e)-(f): values of $\overline{T_{a,R}}$ and $\overline{T_{a,L}}$. The trials were carried out with different values of $K_C$: 100 µs/Nm (panels (a)-(d)), 150 µs/Nm (panels (b)-(e)), and 300 µs/Nm (panels (c)-(f)). Only the first 60 s in which the stimulation was on are reported.

Finally, it would be better to fix $K_C$ at a value lower than 130 µs/Nm to be sure that the closed-loop system is stable without oscillations. From equation (5) follows that stability will not be lost for any decrease in the value of $K_P$ during the cycling session, e.g. caused by muscular fatigue.

## 4.3 Results

Figure 7 reports the results of a simulation trial characterised by the protocol described in Section 3. The initial values of $PW_L$ and $PW_R$ were the same and fixed at 300 µs; the value of $K_C$ of both the integral controllers was set at 50 µs/Nm to update the $PW$ gradually. As shown in panel (b), at the beginning the value of $\overline{T_{a,R}}$ was lower than the one of $\overline{T_{a,L}}$, because the right was the impaired side in the model. Thus, the $PW_R$ increased until the difference between $\overline{T_{a,L}}$ and $\overline{T_{a,R}}$ became zero (at about 90 s). Between the 90 s and the 130 s, the controller maintained the symmetry. It is possible to notice a slow decrease of both the $\overline{T_{a,L/R}}$ due to the occurrence of the muscular fatigue. Then (130 s-140 s), in order to test the robustness of the controller, a positive constant value of 4 Nm was added to the $\overline{T_{a,R}}$. Accordingly, because the value of $\overline{T_{a,R}}$ became higher than the one of $\overline{T_{a,L}}$, the controller tried to nullify the difference, first increasing the $PW_L$ till the maximum value and then decreasing the $PW_R$. When the disturbance ceased, the $\overline{T_{a,R}}$ suddenly became lower than the $\overline{T_{a,L}}$ and, therefore, the $PW_R$ increased again till the maximum value and then the $PW_L$ decreased till the symmetry was reached, at about 170 s.

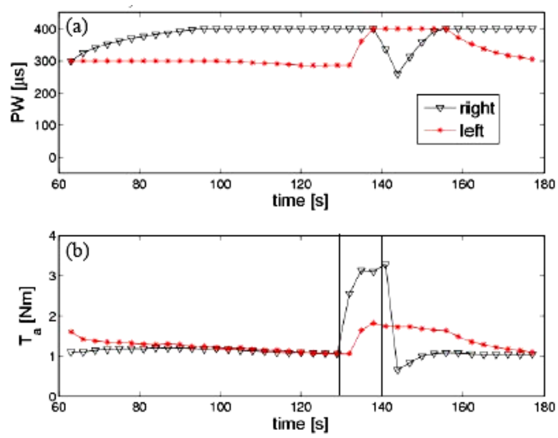


Figure 7: Panel (a): values of $PW_R$ and $PW_L$. Panel (b): values of $\overline{T_{a,R}}$ and $\overline{T_{a,L}}$. The black lines indicate the period in which a positive constant value of 4 Nm was added to the $\overline{T_{a,R}}$. Only the phase in which the stimulation was on is reported.

# 5 EXPERIMENTAL TRIALS

## 5.1 The Experimental Setup

The experimental setup developed includes an 8-channel stimulator (Rehastim[TM], Hasomed GmbH, Germany) and a motorised cycle-ergometer (THERA-live[TM], Medica Medizintechnik GmbH, Germany) equipped by resistance strain gauge sensors able to measure the torques at the left and right crank arm. These signals are transmitted from the ergometer to a desktop PC via wireless, providing a measure of the unbalance between the two legs during cycling. More details on the setup can be found in (Ferrante, 2008b).

In all the trials, an ON-OFF PW profile was used. The stimulation currents were set at a value, tolerated by the subject, which produces a good muscular contraction. The stimulation frequency was fixed at 20 Hz and all the signals were sampled at 200 Hz.

## 5.2 Results

Figure 8 shows the results of the controller test carried out on an able-bodied subject. The subject was a female (24 years old, 166 cm and 52 kg). The stimulation currents used were 30 mA for all the 4 muscles. The value of $K_C$ of both the integral controllers was set at 50 µs/Nm. Moreover, in this trial, the values of $PW$ were fixed at 300 µs and 100 µs for the left and the right leg, respectively, in order to induce an unbalance between the two sides in a healthy subject.

Figure 8 shows that, at the beginning, there was a slight unbalance which the controller tried to compensate increasing the $PW_R$.

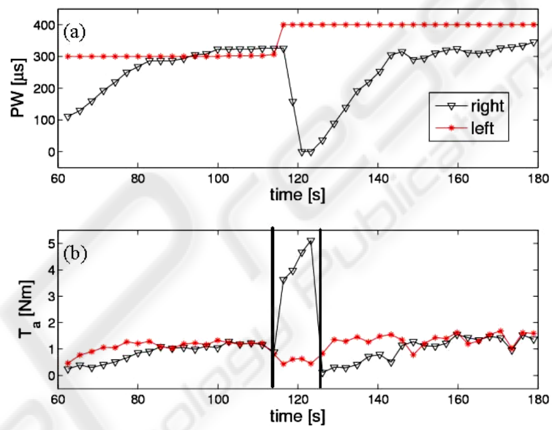


Figure 8: Panel (a): values of $PW_R$ and $PW_L$. Panel (b): values of $\overline{T_{a,R}}$ and $\overline{T_{a,R}}$. The black lines indicate the period in which the subject pedalled voluntary only with the right leg. Only the phase in which the stimulation was on is reported.

After 85 s, the controller achieved a symmetrical movement. This symmetry was maintained till the subject began to pedal voluntarily only with the right leg (115 s-125 s) producing a high increment in the $\overline{T_{a,R}}$ (panel (b)). Thus, the controller increased the $PW_L$ and then, when the saturation value was reached, the controller reduced the value of $PW_R$ (panel (a)). This controller action was not sufficient to gain symmetry because the subject was pedalling voluntarily. When the subject stopped to pedal voluntarily with the right leg, $\overline{T_{a,R}}$ suddenly decreased to zero, which corresponds to the mean value during passive cycling. Indeed, the leg was not stimulated at all. Thus, the $PW_R$ started to increase and the symmetry was re-gained in about 25 s and maintained till the end of the trial.

## 6 DISCUSSIONS AND CONCLUSIONS

The present study deals with the design and the testing of a novel closed-loop controller for FES cycling, able to gain and then to maintain the symmetry of the pedalling in stroke patients. This controller could be useful in the rehabilitation of these patients, who need to re-learn symmetrical tasks in order to recover basic motor functions, such as walking.

Furthermore, a neuro-muscolo skeletal model to simulate cycling induced by FES in stroke patients was developed. This simulator aided in the tuning of the controller parameters and in the validation of the controller before testing it experimentally.

Finally, first trials on healthy subjects were carried out. Starting from a measurement in real-time of the unbalance between the torques produced by each leg at the cranks, the controller was able to reach and then to maintain a symmetrical pedalling, modifying the stimulation parameters of the two lower limbs, independently. For example, the initial unbalance of 0.38 Nm was nullified by the controller in about 18 s as shown in the results reported in Figure 8. The results of this trial showed also that the controller maintained the symmetry of the pedalling by means of small oscillations of the values of *PW*, till an external contribution occurred. When the subject started to pedal voluntarily only with the right leg, an unbalance between the two legs was introduced again and the controller answered properly, without showing an unstable behaviour. When the subject ceased to pedal voluntarily, the unbalance of the pedalling was about 0.73 Nm and the controller re-gained the symmetry in about 18 s. This trial showed clearly that the system is not linear; indeed, even if the unbalance doubles, the time needed to reach symmetry is the same.

The automatic control system developed shows a reliable behaviour. Thus, the next step will be the testing of the controller on stroke patients to demonstrate if this system could be actually useful in the rehabilitation of these patients, accelerating and improving the motor recovery of the lower limbs.

## ACKNOWLEDGEMENTS

This work was supported by the Italian Institute of Technology (IIT). It was also partly funded through grant by the German Federal Ministry of Education and Research (BMBF) within the project RehaRobES (FKZ 01EZ0766).

## REFERENCES

Sheffler, L.R., Chae, J. (2007) ‘Neuromuscular electrical stimulation in neurorehabilitation’, *Muscle & Nerve*, vol. 35, no. 5, pp. 562-590.

Hunt, K.J., Stone, B., Negard, N.O., Schauer, T., Fraser, M.H., Cathcart, A.J., Ferrario, C., Ward, S.A., Grant, S. (2004) ‘Control strategies for integration of electric motor assist and functional electrical stimulation in paraplegic cycling: utility for exercise testing and mobile cycling’, *IEEE Transactions on Neural and Rehabilitation Systems Engineering*, vol. 12, no. 1, pp. 88-101.

Trumbower, R.D., Faghri, P.D. (2005) ‘Kinematic analyses of semireclined leg cycling in able-bodied and spinal cord injured individuals’, *Spinal Cord*, vol. 43, no. 9, pp. 543-549.

Ferrante, S., Pedrocchi, A., Ferrigno, G., Molteni, F. (2008a) ‘Cycling induced by functional electrical stimulation improves the muscular strength and the motor controls of individuals with post-acute stroke’, *European Journal of Physical and Rehabilitation Medicine*, vol. 44, no.2, pp. 159-167.

Hunt, K.J., Ferrario, C., Grant, S., Stone, B., Mclean, A.N., Fraser, M. H., Allan, D.B. (2006) ‘Comparison of stimulation patterns for FES-cycling using measures of oxygen cost and stimulation cost’, *Medical Engineering & Physics*, vol. 28, no. 7, pp. 710–718.

Ferrante, S., Comolli, L., Pedrocchi, A., Bocciolone, M., Ferrigno, G., Molteni F. (2008b) ‘Optimization of FES cycling neuroprosthesis on stroke patients by means of the left and right crank measurement’, *Biodevices – Conference Proceedings*, International Conference on Biomedical Electronics and Devices, Madeira, Portugal, pp. 206-211.

Riener, R., Fuhr, T. (1998) ‘Patient-driven control of FES-supported standing up: a simulation study’, *IEEE Transactions on Rehabilitation Engineering*, vol. 6, no. 2, pp. 113-124.

Abbas, J., Chizack, H. (1995) ‘Neural network control of functional neuromuscular stimulation systems: Computer simulation studies’, *IEEE Transactions on Biomedical Engineering*, vol. 42, no. 11, pp. 1117-1127.

Ambrosini, E., Schauer, T., Ferrante, S., Pedrocchi, A., Ferrigno, G (2008) ‘Neuro-muscolo skeletal model for cycling induced by functional electrical stimulation in individuals with stroke’, accepted by 9th International Workshop on Research and Education in Mechatronics, Bergamo, Italy.