31st Annual International Conference of the IEEE EMBS
Minneapolis, Minnesota, USA, September 2-6, 2009

An FES Cycling Control System Based on CPG

Peng-Feng Li, Zeng-Guang Hou, Feng Zhang, Min Tan, Hong-Bo Wang, Yi Hong, Jun-Wei Zhang

Abstract— **This paper presents a scientific strategy for cycling induced by the functional electrical stimulation. In order to simulate the FES-cycling movement produced by human body, a neuro-musculo-skeletal model containing 16 segments and 186 muscles is developed, which can simulate human movements precisely. This model contains mathematical model of electrically stimulated skeletal muscles. Having known the kinematics and dynamics of the model, we design an FES-cycling control system based on the central pattern generator (CPG), which can produce rhythm stimulus to produce desired torque and generate rhythm cycling movements. And an approach to control multiple muscles is proposed. In the end of this paper, the simulation results are provided.**

I. INTRODUCTION

To restore the function of the paralyzed extremities, functional electrical stimulation (FES) has been used since 1960s. There are mainly three applications of functional electrical stimulation for rehabilitation of patients with spinal cord injury (SCI): standing, walking and cycling. FES-cycling has been mostly studied since it is easier and safer for patients [1]-[4]. In this paper, we focus on cycling induced by FES.

In this paper, a three-dimensional neuron-musculo-skeletal model of human body with 16 segments and 186 muscles is developed. The lower extremities of the model which contain 7 segments including crank and 86 muscles, is used for FES-cycling simulation. The muscles are divided into 7 groups in each side. This is meaningful, because one pair of electrodes may activate more than one muscle or even may activate several muscle groups [1], depending on its position and stimulation intensity. The mathematical model of electrically stimulated skeletal muscles is a nonlinear Hill type model [1][5]-[7], which is controlled by the pulse width (or amplitude) and pulse frequency of the electrical stimulus.

An FES-cycling control system based on the central pattern generator (CPG) is developed. An approach has been proposed to control multiple muscles and minimize the stimulation input. Simulation results are given in section III.

II. METHODS

The block diagram of simulation of FES-cycling control system are shown in Fig. 1.

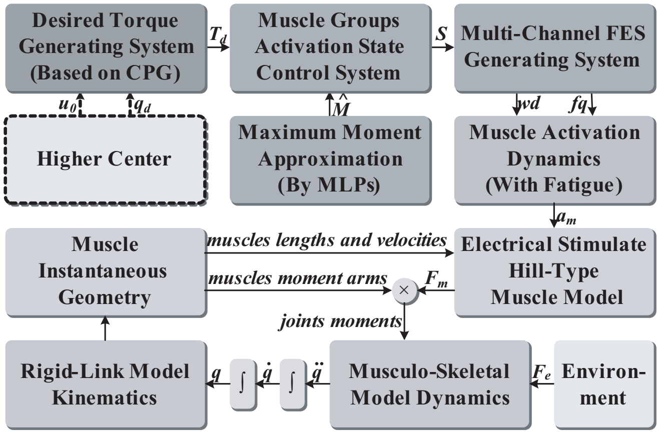

Fig. 1. Block diagram of the FES-cycling system.

A. Neuro-Musculo-Skeletal Model

The neuron-musculo-skeletal model has 16 rigid links and 41 degrees of freedom. The muscular model consists of 186 muscles, 86 for lower extremities and 100 for upper body. The muscles model are based on line segment model [8], which uses several points and lines to represent muscle. Fig. 2 shows the line segment model of biceps brachii lone head.

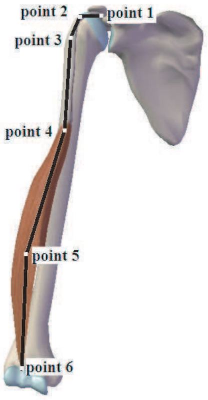

Fig. 2. Muscle line segment model.

The inertial properties of each body segment and parameters of muscles, including mass and moment of inertia, are from literatures [9]-[15]. Fig. 3 shows the musculo-skeletal model we developed in MATLAB^{TM}.

B. Kinematics and Dynamics

When cycling, the hip joint is restricted to one degree of freedom, the ankle joint is fixed with a plastic ankle

Peng-Feng Li, Zeng-Guang Hou, Feng Zhang and Min Tan are with the the Key Laboratory of Complex Systems and Intelligence Science, Institute of Automation, Chinese Academy of Sciences, P.O. Box 2728, Beijing 100190, China. E-mails: lipengfeng007@163.com; hou@compsys.ia.ac.cn; zfeng1203@yahoo.com.cn; tan@compsys.ia.ac.cn.

Hong-Bo Wang is with the School of Mechanical Engineering, Yanshan University, Hebei 066004, China. E-mail: wzfamilyhb@gmail.com.

Yi Hong and Jun-Wei Zhang are with Department of SCI Surgery, China Rehabilitation Research Center, Beijing 100068, China. E-mails: hongyihhyy@yahoo.com.cn; jwnihaoma@yahoo.com.cn.

Peng-feng Li and Feng Zhang are also with the Graduated School of Chinese Academy of Science, 19 Yuquan Road, Shijingshan, Beijing, China.

978-1-4244-3296-7/09/$25.00 ©2009 IEEE

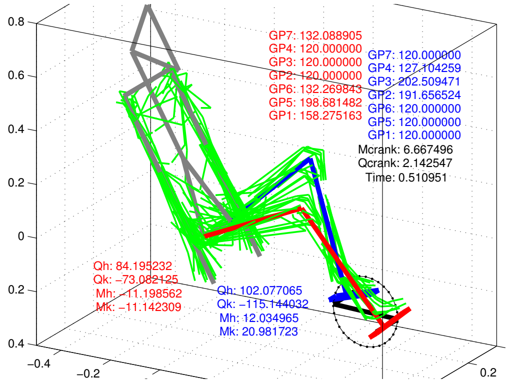

Fig. 3. Musculo-skeletal model.

foot orthosis for protection. The FES cycling model has one degree of freedom, which is represented by crank angle θ. The joint angles of left and right legs are symmetrical and the phase difference is π.

Dynamics of FES cycling model is gotten based on virtual work principle and Euler-Lagrange formulation.

C. Model of Electrically Stimulated Skeletal Muscle

The model of electrically stimulated skeletal muscle is based on the Hill type muscle model. Muscle tension can be represented as [5][6][16][18]

$$F_m = F_{max} L\left(l_m\right) V\left(l_m, v_m\right) a_m \tag{1}$$

where F_{max} is the maximum isometric muscle force; l_m is the muscle length; v_m is the contractile velocity; L is the force length relationship; V is the force velocity relationship; a_m is muscle activation state.

When muscle is stimulated by electrodes, muscle activation state a_m can be expressed as [19]

$$a_m = a\left(w_m f_m\right) \tag{2}$$

where w_m is the stimulation pulse width factor and f_m is the frequency factor.

The action potential process is a linear second-order Ca^{2+} dynamics with time constant T_{Ca} and damping coefficient ζ. The fatigue process which represented by a_{fit}, is a first-order difference equation with fatigue time constant T_{fat} and recovery time constant T_{rec} [2][19]. The muscle activation factor a_m has a time delay T_{del} which accounts for the actin potential propagation [19].

D. Torque Generate System Based on CPG

Central pattern generator is used to generate rhythm control signal. CPG can be expressed as [2][16][17][18]

$$\tau_0^y \dot{y}_0 + y_0 = -\delta \max\left(y_1, 0\right) - \beta z_0 + u_0 + feed \tag{3a}$$

$$\tau_0^z \dot{z}_0 + z_0 = \max\left(y_0, 0\right) \tag{3b}$$

$$\tau_1^y \dot{y}_1 + y_1 = -\delta \max\left(y_0, 0\right) - \beta z_1 + u_0 - feed \tag{3c}$$

$$\tau_1^z \dot{z}_1 + z_1 = \max\left(y_1, 0\right) \tag{3d}$$

$$C_{out} = \max\left(y_0, 0\right) - \max\left(y_1, 0\right) \tag{3e}$$

where y_i is the inner state of CPG; z_i is the fatigue state; δ is the connecting coefficients; β is the weight of fatigue state; u_0 is the non-specific input; $feed$ can be expressed as

$$feed\left(\theta, \dot{\theta}\right) = k_p\left(\theta_d - \theta\right) + k_d\left(\dot{\theta}_d - \dot{\theta}\right) + k_i. \tag{4}$$

The desired quasi-torque is the CPG output with inverse dynamics

$$T_d = k_c C_{out} + ID\left(\theta, 0, 0\right). \tag{5}$$

E. Multi-Muscle Control Strategy

According to (1) and (2), let

$$s_{i,j} = w_{i,j} f_{i,j}$$

so the force of the jth muscle in the ith muscle group is

$$\begin{aligned} F_{i,j}^M &= F_{i,j}^{Max} L_{i,j}\left(\theta\right) V_{i,j}\left(\theta, \dot{\theta}\right) a_{i,j}\left(s_{i,j}\right) \\ &= F_{i,j}\left(\theta, \dot{\theta}\right) a\left(s_{i,j}\right). \end{aligned} \tag{6}$$

The moment of crank generated by the jth muscle in the ith muscle group can be obtained by

$$\begin{aligned} M_{i,j} &= M_{i,j}^{Hip} \frac{dq_h}{d\theta} + M_{i,j}^{Knee} \frac{dq_k}{d\theta} \\ &= F_{i,j}^M \times \left(R_{i,j}^{Hip} \frac{dq_h}{d\theta} + R_{i,j}^{Knee} \frac{dq_k}{d\theta}\right) \\ &= M_{i,j}^{Max}\left(\theta, \dot{\theta}\right) a\left(s_{i,j}\right) \end{aligned} \tag{7}$$

where $R_{i,j}^{Hip}$, $R_{i,j}^{Knee}$ are the moment arms about hip and knee joints of the jth muscle in the ith muscle group.

Muscles of the same group have the same stimulation intensity, so their activations are the same $a\left(s_{i,j}\right) = a\left(s_i\right)$. Thus total moment of the crank can be obtained by

$$\begin{aligned} M_{crank} &= \sum_{i=1}^{N_{Group}} \sum_{j=1}^{N_{Muscle}^{Group,i}} M_{i,j} \\ &= \sum_{i=1}^{N_{Group}} M_i^{Max}\left(\theta, \dot{\theta}\right) a\left(s_i\right) \end{aligned} \tag{8}$$

where $M_i^{Max}\left(\theta, \dot{\theta}\right)$ is the maximum crank moment generated by the ith muscle group, when this muscle group is in the highest activation state. $M_i^{Max}\left(\theta, \dot{\theta}\right)$ can hardly be obtained analytically. So multilayer perceptrons (MLPs) are used to approximate each $M_i^{Max}\left(\theta, \dot{\theta}\right)$. MLPs are trained offline by extreme learning machine algorithm.

Since T_{fat} and T_{rec} are much larger than T_{Ca} and T_{del}, the activation state can be expressed as 3^{rd} order ordinary differential equation without considering muscle fatigue for simplicity. Let

$$\begin{aligned} y &= \hat{M}_1^{Max} a_1 + \hat{M}_2^{Max} a_2 + \cdots + \hat{M}_{14}^{Max} a_{14} \\ u &= \hat{M}_1^{Max} s_1 + \hat{M}_2^{Max} s_2 + \cdots + \hat{M}_{14}^{Max} s_{14} \end{aligned}$$

so torque generating system can be expressed as

$$c_3 \dddot{y} + c_2 \ddot{y} + c_1 \dot{y} + y = u \tag{9}$$

where c_3, c_2 and c_1 are coefficients related to T_{Ca} and T_{del}.

A composite control system with feedback and feedforward control as shown in Fig. 4 has been designed to control torque generating system. A tracking differentiator TD has been used to approximate T_d

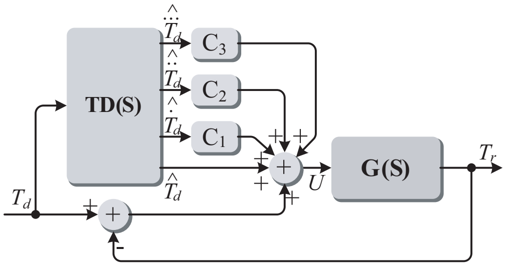

Fig. 4. Composite control system with tracking differentiator.

$$TD(S) = \frac{1}{(T_{track}S+1)^4} \tag{10}$$

where T_{track} is the tracking time constant and should be as small as enough. In this simulation $T_{track} = 0.001$. Input signal to the system is given by

$$u = (T_d - T_r) + \left(c_3\hat{\dddot{T}}_d + c_2\hat{\ddot{T}}_d + c_1\hat{\dot{T}}_d + \hat{T}_d\right). \tag{11}$$

With the simulation parameters listed in Tab. I, the close-loop transfer function can be obtained

$$G_c(S) = 2.5\times10^4\frac{\left(S+4\times10^7\right)}{(S+1000)^4} \cdot \frac{(S+58.04)\left(S^2+32.11S+861.5\right)}{(S+60.19)\left(S^2+29.81S+830.7\right)}$$

thus

$$\lim_{S\to0} G_c(S) = 1.000032540108684.$$

F. Multi-Channel FES Simulation Optimization Strategy

In (9), muscle group activation input can be expressed as

$$u = \hat{M}_1^{Max}s_1 + \hat{M}_2^{Max}s_2 + \cdots + \hat{M}_{14}^{Max}s_{14}$$

where s_i is the FES stimulation intensity; $\hat{M}_i^{Max}$ is the maximum crank moment that ith muscle group can generate.

Let

$$\hat{M} = [\hat{M}_1^{Max}, \hat{M}_2^{Max}, \cdots, \hat{M}_{14}^{Max}]^T$$
$$S = [s_1, s_2, \cdots, s_{14}]^T$$

where $s_i \in [0, s_{max}]$, $s_{max} = 0.9$ in this simulation.

Based on the energy and muscle fatigue minimization principle, the stimulation intensities distribution problem becomes a constrained least square optimization problem

$$\min_S \left\{S^TS \mid \hat{M}^TS = u\right\}.$$

Finally, it can be obtained that

$$S = \frac{2u\hat{M}}{\hat{M}^T\hat{M}}.$$

If $s_i \le 0$, $s_i = 0$ and if $s_i \ge s_{max}$, $s_i = s_{max}$. Redo the optimization process until all s_i have been gotten. Then FES stimulation pulse width factor can be obtained by [19]

$$w_i = \frac{s_i}{f_i}$$

where $f_i = 0.9$ when stimulation pulse frequency is 100Hz in this simulation.

III. SIMULATION RESULTS

The simulation is conducted in MATLAB^{TM} environment. The adaptive Runge-Kutta-Fehlberg integration method is used to solve differential equations. Parameters for the links and coordinates of muscles segments are from [9]-[15]. Parameters for the muscle activation state are listed in Tab. I. Parameters of CPG are listed in Tab. II.

TABLE I

PARAMETERS FOR MUSCLES [19].

T_{Ca}	T_{rec}	T_{del}	α	β	ζ
s	s	s			
0.04	30	0.025	0.1	0.6	1.0

TABLE II

PARAMETERS FOR CENTRAL PATTERN GENERATOR.

τ_i^y	τ_i^z	δ	β	u_0	k_p	k_d	k_i	k_c
0.2	0.2	2.0	2.5	5.0	0.5	0.1	0.6	1.0

The result are shown from Fig. 5 to Fig. 10. Results of the right leg are similar to the left.

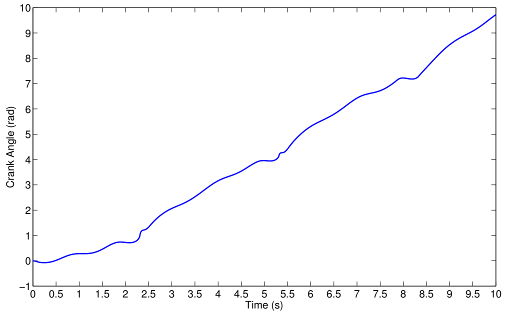

Fig. 5. Crank angle.

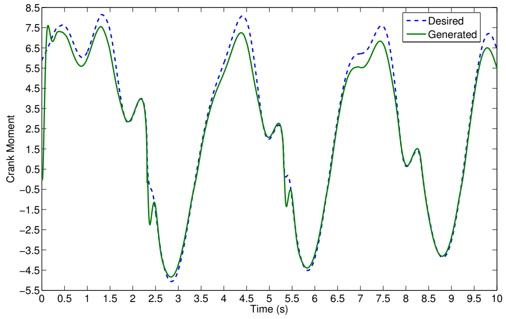

Fig. 6. Desired and generated crank moment.

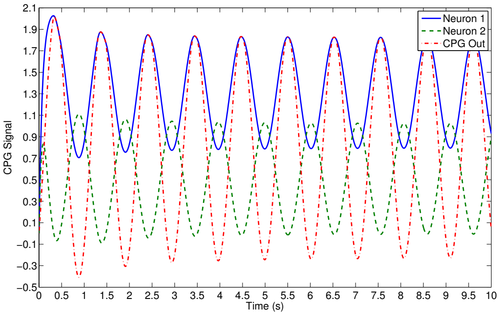

Fig. 7. CPG oscillator signals and CPG output.

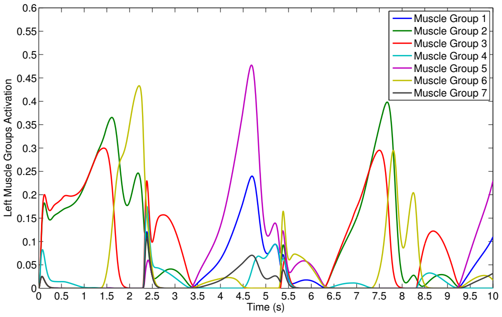

Fig. 9. Activation state of muscle groups (left).

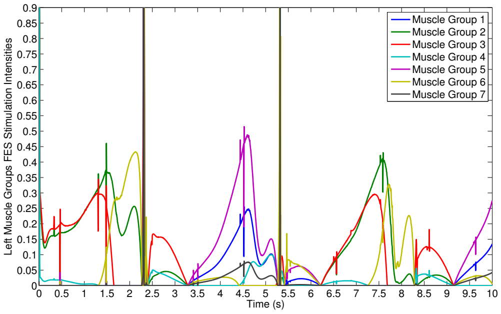

Fig. 8. FES stimulation intensities of muscle groups (left).

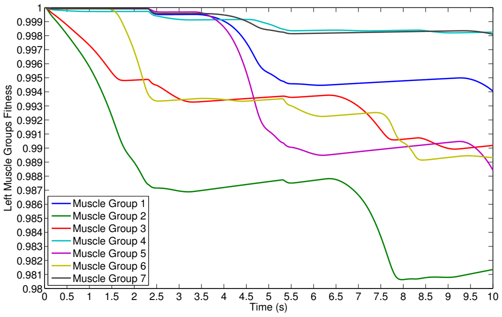

Fig. 10. Fitness of muscle groups (left).

IV. CONCLUSION AND FUTURE WORK

In this paper, an FES-cycling system is developed based on a three-dimensional musculo-skeletal model. There is much work deserved in the future. The FES cycling control system must be further improved. The feedback signal provided to central pattern generator should be improved. And parameters of CPG need to be optimized.

REFERENCES

[1] J. J. Chen, N. Y. Yu, D. G. Huang, B. T. Ann, G. C. Chang, "Applying fuzzy logic to control cycling movement induced by functional electrical stimulation", *IEEE Trans. Rehabilitation Engineering*, 1997, vol.5, no.2, pp. 158-168.

[2] C. S. Kim, G. M. Eom, K. Hase, G. Khang, G. R. Tack, J. H. Yi, J. H. Jun, "Stimulation pattern-free control of FES cycling: simulation study", *IEEE Trans. Systems, Man, And Cybernetics—Part C: Application And Reviews*, 2008, vol.38, no.1, pp. 125-134.

[3] K. J. Hunt, B. S. Stone, N. O. Negård, T. Schauer, M. H. Fraser, A. J. Cathcart, C. Ferrario, S. A. Ward, S.Grant, "Control strategies for integration of electric motor assist and functional electrical stimulation in paraplegic cycling: utility for exercise testing and mobile cycling", *IEEE Trans. Neural Systems and Rehabilitation Engineering*, 2004, vol.12, no.1, pp. 89-101.

[4] M. Gföhler, P. Lugner, "Dynamic simulation of FES-cycling: influence of individual parameters", *IEEE Trans. Neural Systems and Rehabilitation Engineering*, 2004, vol.12, no.4, pp. 398-405.

[5] H. Hatze, "A myocybernetic control model of skeletal muscle", *Biol. Cybernetics*, 1977, vol.25, pp. 103-119.

[6] H. Hatze, "A general myocybernatic control model of skeletal muscle", *Biol. Cybernetics*, 1978, vol.28, pp. 143-157.

[7] S. J. Dorgan, M. J. O'Malley, "A nonlinear mathematical model of electrically stimulated skeletal muscle", *IEEE Trans. Rehabilitation Engineering*, 1997, vol.5, no.2, pp. 179-194.

[8] K. Hase, N. Yamazaki, "Development of three-dimensional whole body musculoskeletal model for various motion analysis", *JSME International Journal Series C*, 1997, vol.49, no.1, pp. 25-32.

[9] E. P. Hanavan, "A mathematical model of the human body", *Final Report for Aerospace Medical Research Laboratories Wright-Patterson Air Force Base, Ohio*, 1964, Report No. AMRL-TR-64-102.

[10] M. P. T. Silva, J. A. C. Ambrósio, M. S. Pereira, "Biomechanical model with joint resistance for impact simulation", *Multibody System Dynamics*, 1997, vol.1, pp. 65-84.

[11] W. Maurel, D. Thalmann, P. Hoffmeyer, P. Beylot, P. Gingins, P. Kalra, N. M. Thalmann, "A biomechanical musculoskeletal model of human upper limb for dynamic simulation", *Proc. of the Eurographics workshop on Computer Animation and Simulation '96, Poitiers, France*, 1996, pp. 121-136.

[12] C. K. Cheng, H. H. Chen, C. S. Chen, C. L. Lee, C. Y. Chen, "Segmental inertial properties of Chinese adults determined from magnetic resonance imaging", *Clinical Biomechanics*, 2000, vol.15, pp. 559-566.

[13] M. P. T. Silva, J. A. C. Ambrósio, "Sensitivity of the results produced by the inverse dynamic analysis of a human stride to perturbed input data", *Gait and Posture*, 2004, vol.19, pp. 35-49.

[14] K. R. S. Holzbaur, W. M. Murray, G. E. Gold, S. L. Delp, "Upper limb muscle volumes in adults subjects", *J. Biomechanics*, 2007, vol.40, pp. 742-749.

[15] M. D. K. Horsman, "The Twente lower extremity model: consistent dynamic simulation of human locomotors apparatus", *Thesis, University of Twente*, 2007.

[16] N. Ogihara, N. Yamazaki, "Generation of human bipedal locomotion by a bio-mimetic neuro-musculo-skeletal model", *Biological Cybernetics*, 2001, vol.84, no.1, pp. 1-11.

[17] G. Taga, "A model of the neuro-musculo-skeletal system for human locomotion: I. emergence of basic gait", *Biological Cybernetics*, 1995, vol.73, no.2, pp. 97-111.

[18] K. Hase, N. Yamazaki, "Computer simulation study of human locomotion with a three-dimensional entire-body neuro-musculo-skeletal model: I. acquisition of normal walking", *JSME International Journal Series C*, 2002, vol.45, no.4, pp. 1040-1050.

[19] R. Riener, T. Fuhr, "Patient-driven control of FES-supported standing up: a simulation study", *IEEE Trans. Rehabilitation Engineering*, 1998, vol.6, no.2, pp. 113-124.