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Higher-Order Sliding Mode Control of Leg Power in Paraplegic FES-Cycling

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***Abstract*—In this paper, we propose a robust control methodology based on high order sliding mode (HOSM) for control of the leg power in FES-Cycling. A major obstacle to the development of control systems for functional electrical stimulation (FES) has been the highly non-linear, time-varying properties of *neuromusculoskeletal* systems. A useful and powerful control scheme to deal with the uncertainties, nonlinearities, and bounded external disturbances is the sliding mode control (SMC). The main drawback of the classical sliding mode is mostly related to the so-called chattering which is dangerous for FES applications. To avoid chattering, HOSM approaches were proposed. Keeping the main advantages of the original approach, at the same time they totally remove the chattering effect and provide for even higher accuracy in realization. The results of simulation studies and experiments on two paraplegic subjects show the superior performance of the leg power control during different conditions of operation using HOSM control scheme.**

I. Introduction

DURING several years, the beneficial effects of FES exercise have been demonstrated with the improvements in peripheral muscular fitness, central cardiovascular fitness, gas exchange kinetics and aerobic metabolism, reduce atrophy of skeletal muscle, increase lower limb circulation and improve immune system function, increase in bone density, and decrease in spasticity [1]-[3]. A safe and easy application of FES is cycling. An important motivation in FES-cycling is to take physiological advantage of functional electrical stimulation in combination with physiological incentive of cycling as an independent and safe locomotive activity.

Several factors could affect the FES-cycling efficacy. The mechanical configuration of ergometer, such as seat position, seat back angle, and cycling load [4], and stimulation patterns [5] contribute to the efficacy of FES cycling for subjects with paraplegia.

To evaluate the physiological effects of FES-cycling, it is crucial that both cycling cadence and power are well controlled. Daily changes in patients' physical condition, highly non-linear and time-varying properties of stimulated muscle and occasional occurrence of spasticity are major impediment to the development of FES-cycling control system. Moreover, the complexity of the interface between the ergometer and stimulated limb can make the design of a system controller even more complicated. To deal with these problems, several control strategies for closed-loop control of FES-cycling movement have been developed and reported in literature [6]-[7].

In [6], a model-free fuzzy logic fixed-parameter feedback controller was adopted for control of the cycling cadence powered by the stimulated quadriceps and hamstrings of both legs. The experiments on subjects with paraplegia confirm the jerky cycling movement and a sudden change in cranking speed.

Hunt *et al*. [7] used an identified linear model to design a closed-loop power controller using pole placement method. However, pole placement controller design is a technique for LTI systems. The basic idea behind it is the design of state feedback such that all poles of the closed-loop system assume prescribed values. The success of the pole placement design is strongly dependent on the availability of an accurate model of the system under study.

As modeling is a well-known bottleneck, there is a strong demand for robust control design that can take model uncertainty into account, while satisfying the closed-loop stability and performance specifications. The idea of utilizing a method that is robust against the nonlinear and time-varying characteristics of the musculoskeletal system and that does not require a precise model motivates us to develop a control strategy based on sliding mode control as a powerful control scheme to deal with the uncertainties, nonlinearities, and bounded external disturbances. The main drawbacks of classical first-order SMC are principally related to the so-called chattering. Chattering is undesirable because it can excite unmodeled high-frequency plant dynamics.

To solve this problem, we have already employed a robust control methodology which is based on synergistic combination of an adaptive single-neuron controller with sliding mode control (SMC) for control of FES-cycling cadence in paraplegic subjects [8]. To implement the SMC, a model of rider-cycle system should be first presented in a canonical form. Identification of such model is a major impediment to implementing the SMC. To cope with this problem, *in this work*, we employ higher-order sliding mode (HOSM) scheme [9] for control of FES-cycling. The HOSM approach has been actively developed over the last two decades not only for chattering attenuation but also for the robust control of uncertain systems with an arbitrary relative degree, and continuous control functions can be achieved while robustness is retained. In particular, no mathematical

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model of the controlled process is actually needed.

In this paper, we employed a second-order SMC with super-twisting algorithm for control of power in FES-cycling in paraplegic subjects.

II. Higher-order Sliding Mode Control

Consider the following single input non-linear system

$$x^{(n)} = f(\mathbf{x}) + b(\mathbf{x}) \cdot u \tag{1}$$

where $\mathbf{x} = [x \;\; \dot{x} \;\; ... x^{(n-1)}]^T \in \Re^n$ is the state vector of the system, $u \in \Re$ the control input, and the nonlinear dynamics of $f(\mathbf{x})$ and $b(\mathbf{x})$ are not exactly known but they are estimated by their nominal forms of $\hat{f}(\mathbf{x})$ and $\hat{b}(\mathbf{x})$, respectively. The objective of the controller is to design a control law to force the system state vector $\mathbf{x}(t)$ to track a desired state vector $\mathbf{x}_{\mathrm{d}}(t)$ in the presence of model uncertainties and external disturbances. The standard sliding mode way is to define a suitable sliding manifold, $s(\mathbf{x},t)$, onto which the dynamic evolution of the system state must be confined. The manifold choice is developed such that the system trajectories satisfy the performance specifications, when the sliding variable lies on the sliding manifold (i.e., when $s(\mathbf{x},t) = 0$). A time-varying surface $s(\mathbf{x},t)$ is defined in the state space $\Re^n$ by equating the variable $s(\mathbf{x},t)$ defined below, to zero.

$$s(\mathbf{x},t) = (\frac{d}{dt} + \lambda)^{n-1} e(t) \tag{2}$$

where λ is a positive constant and e is the state error. By solving the above equation for the control input using (1), we obtain the expression for u which is called equivalent control. The equivalent control keeps the system states in the sliding surface $s = 0$.

The main idea behind higher order sliding modes is to act on the higher order derivatives of the sliding variables as compared to the 1st-order derivative in standard sliding mode technique. Hence, the *r*th-order sliding mode is determined by the equalities

$$s = \dot{s} = \ddot{s} = ... = s^{(r-1)} = 0 \tag{3}$$

forming an *r*-dimensional condition on the state of the dynamic system.

The control goal of the second-order sliding mode controller is that of steering $s, \dot{s}$ to zero in finite time by defining a proper control $u(t)$. To achieve this, the following conditions are assumed:

1) $$0 < K_m < \left|\frac{\partial \dot{s}}{\partial u}\right| \le K_M \tag{4}$$

2) $$\left|\frac{\partial}{\partial t}\dot{s}(t,x,u) + \frac{\partial}{\partial x}\dot{s}(t,x,u) f(t,x,u)\right| \le C_0 \tag{5}$$

3) $$|s| < s_0 \tag{6}$$

Different kinds of second order sliding mode algorithms can be found in the literature: Twisting, Sampled Twisting, Super-Twisting, Sub-Optimal [9]. In this paper, we used Super-twisting algorithm to implement 2-order SMC. This algorithm has been developed for systems with relative degree 1 to avoid the chattering phenomena. The control law comprises two continuous terms that, again, do not depend upon the first time derivative of the sliding variable. The discontinuity only appears in the control input time derivative. The algorithm is defined by the following control law

$$\begin{aligned} u &= -\lambda \cdot |s|^{\rho} \cdot \mathrm{sgn}(s) + u_1 \\ \dot{u}_1 &= -W.\mathrm{sgn}(s) \end{aligned} \tag{7}$$

and the corresponding sufficient conditions for the finite time convergence to the sliding manifold are

$$\begin{aligned} & W > \frac{C_0}{K_m} \\ & \lambda^2 \ge \frac{4C_0 K_M (W + C_0)}{K_m K_m^2 (W - C_0)} \\ & 0 < \rho \le 0.5 \end{aligned} \tag{8}$$

III. Simulation Studies

A model of rider-cycle system developed in [10] is used here as a virtual patient for evaluating the control strategy. The model consists of two double pendulums, each consisting of a thigh and a shank and attached to a fixed point at the hip (Fig. 1). Both pendulums have two degrees of freedom, which can be expressed as the hip and knee angles, but the endpoints are constrained to move on a circle around the center of the crank.

The muscle model used for simulation of equivalent flexor and extensor muscles are based on the work of Abbas *et al.* [11], which included an input delay, nonlinear recruitment, linear dynamics, and multiplicative nonlinear torque–angle and torque–velocity scaling factors.

A. Trajectory Tracking

Fig. 2(a) shows the results of cadence control using HOSM. It is observed that an accurate tracking control can be achieved using HOSM. Interesting observation is the fast convergence of the control strategy. It is apparent that the HOSM control strategy is able to provide remarkably fast and robust tracking with a smooth control action.

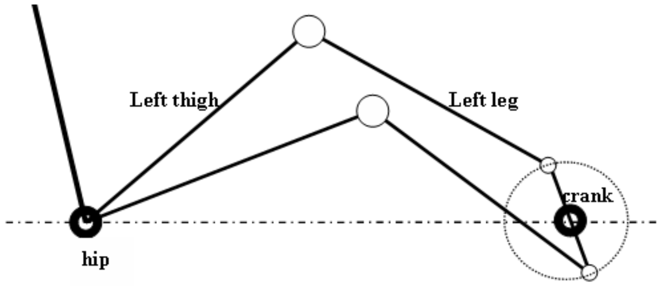

Fig. 1. Model of rider-cycle system. Legs and crank represented by linked rigid bodies.

B. External Disturbance Rejection

To evaluate the ability of proposed control strategy to external disturbance rejection, a constant torque in amount of 7 Nm was subtracted suddenly from the torque generated at the crank for duration of 30 s. Fig. 2 (b) shows that the control strategy is able to compensate the effects of the external disturbances.

C. Fatigue Compensation

Fatigue was included in the virtual patient to evaluate the capability of the control strategy to compensate for it. The effects of muscle fatigue were simulated by an asymptotic decrease in the agonist's (antagonist's) input gain to 50% of its original value over 120 s. Fig. 2(c) demonstrates that the HOSM controller could adjust the stimulation pattern to compensate the effect of muscle fatigue and provide an accurate tracking performance for the cadence of cycling during muscle fatigue.

IV. EXPERIMENTAL EVALUATION

A. Apparatus and System

We adapted a stationary ergometer cycling with an auxiliary electric motor for paraplegic FES cycling. The torque produced at the cranks was measured by a rotary torque sensor (TRD605, Futek Advanced Sensor Technology, Inc, USA) which was mounted on the chassis of the bike and driven by a belt attached to the pulley around the rod of the crank. The sensor also provides a measurement of crank angle using an incremental shaft encoder with a 720 pulse/turn resolution. The cycling cadence was obtained by differentiation of the angle. Thus, instantaneous leg power input can be computed as the product of instantaneous torque and angular speed. Ankle-foot orthoses were designed and were fixed to the pedals to stabilize the ankle joints and constrain the legs to motion in the sagittal plane.

Four muscle groups, i.e. the left and right quadriceps and hamstrings were stimulated by using a computer-based closed-loop FNS system. Pulse width modulation (from 0 to 700 µsec) with balanced bipolar stimulation pulses, at a constant frequency (25 Hz) and constant amplitude was used. The measured crank angle was used to switch on and off the stimulation according to a predefined pattern [6].

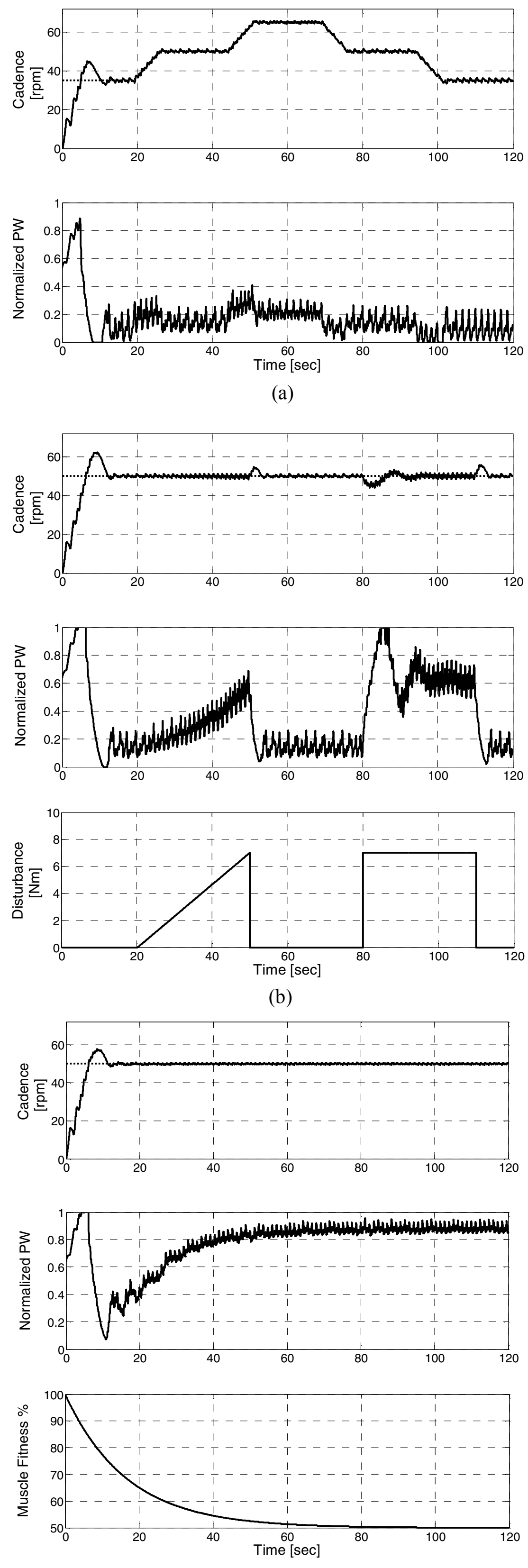

Fig. 2. Simulation results of the cadence control using HOSM control ($W = 0.15, \lambda = 0.3, \rho = 0.5$). (a) Trajectory tracking. (b) Disturbance rejection. (c) Fatigue compensation.

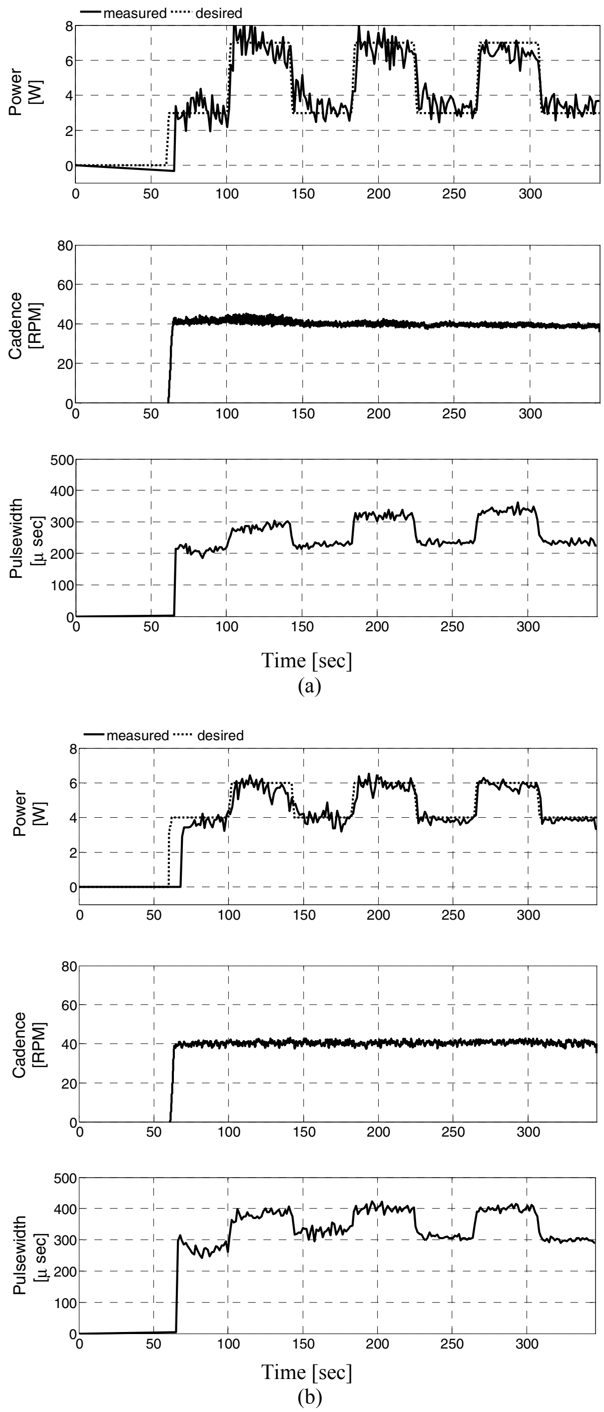

Fig. 3. The results of HOSM control ($W = 0.05, \lambda = 0.2, \rho = 0.5$) of the leg power in FES-Cycling for paraplegic subjects EA (a) and MR (b). .

The computer-based closed-loop FNS system uses Matlab Simulink, Real-Time Workshop, and Real-Time Windows Target under Windows XP for online data acquisition, processing, and controlling.

B. Experimental Results

The experiments were conducted on two complete paraplegics with injury at T7 and T12 levels. Typical results of the leg power control for two paraplegic subjects are shown in Fig. 3. It can be seen that following the initial transient, the controller keeps the leg power close to the desired power trajectory. The convergence of the leg power to the desired trajectory is very fast and the control action is seen to be smooth. Moreover, it is observed that by passing the time, the control signals are trending upward to compensate the effect of muscle fatigue. During this experiment, the cycle cadence is kept constant. The ergometer has the ability to keep the crank velocity constant.

V. Conclusion

In this paper, high-order sliding mode control scheme was used for control of the leg power in paraplegic FES-Cycling. The major motivation of using HOSM control is its robustness against external disturbances and time-varying properties of neuromusculoskeletal system. In particular, it does not require a model for the process being controlled. The controller requires no prior knowledge about the system and no offline identification. These remarkable features make the control scheme to be attractive for FES applications.

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