Synergetic Control Strategy for a Hybrid FES-Exoskeleton System: A Simulation Study

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Abstract. This paper proposes a novel control strategy for a hybrid assistive system that combines functional electrical stimulation (FES) with a powered lower extremity exoskeleton to provide adaptive torque assistance for paraplegic patients. The control architecture is based on central pattern generator (CPG) that acts as a feature extractor, and a proportional-derivative (PD) controller that adaptively regulates the exoskeleton's assistive torque. Currently our work focuses on controlling the swing of shank, so two muscle groups (Vasti and Hamstrings) are stimulated to produce active torque for knee joint. However some drawbacks such as rapid muscle fatigue and uneven muscle response severely limit FES-aided systems. We use a powered extremity exoskeleton to adaptively compensate the torque loss. The CPG model keeps its output in phase with the measured angle trajectory to predict total torque with a torque estimator. The PD controller regulates the gain that determines how much of estimated total torque should feed back to the musculoskeletal system. The control system is implemented in MATLAB/SIMULINK.

Keywords: Synergetic control, functional electrical stimulation (FES), exoskeleton, central pattern generator (CPG), PD control.

1 Introduction

Functional electrical stimulation (FES) has been demonstrated as an effective way to help paralyzed patients such as spinal cord injured (SCI) restore legged mobility. However considering the musculoskeletal system's nonlinearity and time variability, controlling FES to assist SCI individuals to move in a natural manner is a complicated problem. There are two significant drawbacks that severely hindered the use of FES [1]. The first is rapid muscle fatigue caused by continuous stimulating muscles and the second is poor controllability resulting in inadequate joint torque to produce reliable limb movement and body support.

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Some hybrid systems have been developed to solve these problems. A controlled-brake orthosis combined with FES aids hip and knee flexion with a spring assist, and uses sensors and modulated friction brakes for the feedback control of joint angles [1-2, 4]. However this device can only provide passive torque for lower limb joints, and the control structure is based on finite state machine that cannot adaptively deal with the synergy between the orthosis and legs.

Exoskeletons, such as Lokomat (Hocoma, AG, Volketswil, Switzerland) [3], has been widely used in gait rehabilitation. However, the patients' muscles are not excited actively in such a pure exoskeleton system, i.e. the movement is generated passively. Therefore it is a natural idea to use a powered exoskeleton as a torque compensator for FES. Some previous work has been done so [5]. However in this work, the FES is merely a subsidiary approach to aid the hip extension with affecting the system when flexion torques are needed. The stimulator basically relies on an on-off controller. We hope the FES technique can be a main actuator to enhance SCI patients' active motion and the loss torque can be adaptively compensated by powered exoskeletons. Therefore we propose a synergetic control structure that can achieve this goal.

The idea originates from human-robot synchrony developed by Ronsse et al [6], which uses adaptive Hopf oscillators and a torque estimator to predict the torque during the elbow's rhythmic flexion-extension movement. Thus robot can detect the users' motion intention and provide realtime torque assistance needed by the user. The whole system in this work does not explicitly model the human and elbow dynamics and regulates the feedback gain by hand. In this paper we propose an alternative method for a hybrid FES-exoskeleton system with automatical movement assistance for the knee joint. For our purpose, the joint angle needs to be measured to entrain the artificial oscillator. A Matsuoka model [7] is used as adaptive oscillators. Besides, we use a model-based feedforward controller combined with PD to control FES-induced joint movements. Some research groups have investigated feedback and adaptive controllers for FES [8], from which our work draws useful ideas.

2 Methods

The simulation study in this paper is for the practical use of hybrid FES-exoskeleton systems. Multiple joints control deals with greater challenges than single joint control, such as motion coupling between joints. Our goal is to present a novel control strategy for the hybrid system so we take the knee joint movement for instance. Unlike the human-robot synchrony method, a reference trajectory is necessary for the simulation. Here we set a sinusoidal angle trajectory for the shank to move.

2.1 The Musculoskeletal Model

To control the movement of knee joint, we adopted a musculoskeletal model based on Hill-type muscle model [9-10] and generic dynamics functions. The Hill-type model describes the activation and contraction properties of muscles. The

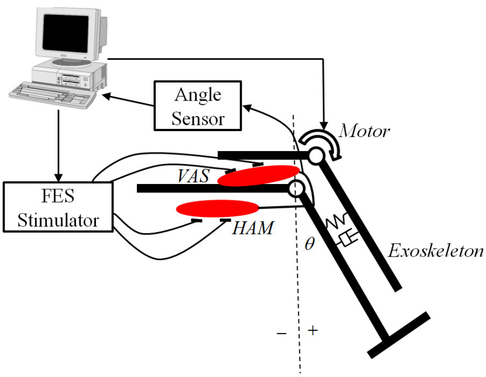

Fig. 1. Simplified model of the hybrid FES-exoskeleton system

muscles can be seen as biological actuators via FES. The whole model consists of three parts (Fig. 2): activation dynamics, contraction dynamics and segmental dynamics, in which muscle fatigue and passive characteristics (viscosity and elasticity) are modeled separately.

The activation dynamics relate FES input to muscle activation. Generally speaking, muscle force can be controlled by adjusting pulse width *PW* and pulse frequency *f*. To minimize the influence of muscle fatigue, we set the pulse frequency at 20Hz and modulate pulse width. The recruitment function is determined by three factors [10]: a threshold pulse width *Thres* [s], a saturation pulse width *Sat* [s], and a scaling factor *Sf* [Nm]. The muscle activation Act is given as follows:

$$Act = \begin{cases} 0 & PW \leqslant Thres \\ \frac{Sf \cdot (PW - Thres)}{Sat - Thres} & Thres < PW < Sat \\ Sf & PW \geqslant Sat \end{cases} \tag{1}$$

Muscle fatigue is one of the most important factors we should consider. We use a generic fitness function *fit(t)* to describe the effect of muscle fatigue:

$$\frac{dfit}{dt} = \frac{(fit_{min} - fit)a(t)\lambda}{T_{fat}} + \frac{(1 - fit)(1 - a\lambda)}{T_{rec}} \tag{2}$$

Where T_{fat} is the time constant for fatigue, and T_{rec} for recovery. λ stands for the effect of stimulation frequency on muscle fatigue. We set f at a certain value so λ is a constant here.

The moment-angle relation is given by [10]:

$$T_{ma} = exp[-(\frac{\theta + \pi/2 - K_1}{K_2})^2] \tag{3}$$

where K_1 and K_2 are two parameters that decide where the maximum and minimum angles locate.

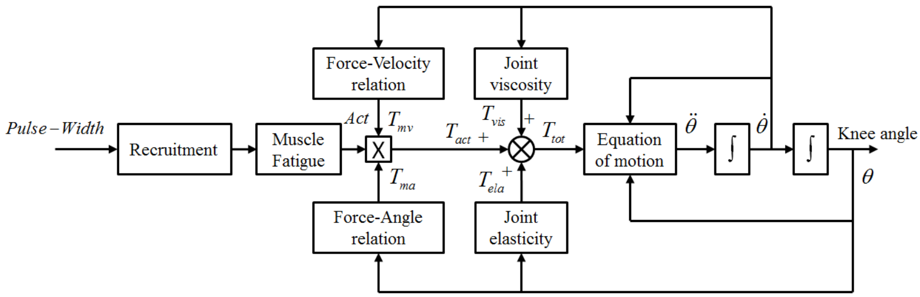

Fig. 2. The musculoskeletal model of the knee joint

The moment-angular velocity can be expressed as follows:

$$T_{mv} = 1 - K_3\dot{\theta}(t) \tag{4}$$

the parameter K_3 equals the knee-moment arm over the maximum contraction velocity. Then we can get the active muscle moment via multiplying activation by the moment-angle and moment-velocity parameters. The total moment exerting on the knee is obtained by adding active, passive and gravitational moments.

The passive viscous of the knee joint can be modeled proportional to the angular joint velocity [9]:

$$T_{vis} = k_{vis}\dot{\theta} \tag{5}$$

with the damping coefficient k_{vis}.

The passive elastic joint moment is related to three joint angles of lower limbs, but we assume that the knee joint oscillates like the scheme described in Fig. 1. Thus the angles of ankle and hip will be constants. The passive elastic joint moment is given by [9]:

$$T_{ela} = exp(3.2872-0.04940\theta)-exp(-1.4261+0.02540\theta)+exp(2.5-0.250\theta)+1.0 \tag{6}$$

For our purpose, we merely consider one-segment skeletal dynamics. The Lagrange's equation is used to solve this problem. Thus the dynamics function is given by:

$$T_{tot} = (\frac{1}{8}ml^2 + \frac{1}{2}J)\ddot{\theta} + \frac{1}{2}mglsin\theta \tag{7}$$

where T_{tot} is the total torque exerting on the knee joint, m is the skeleton mass and J the moment of inertia, and g is the gravitational constant.

2.2 The Control Architecture

In this paper, we propose a synergetic control strategy for the hybrid FES-exoskeleton system based on initial work on human-robot synchrony, which uses adaptive oscillators to provide flexible torque assistance. However in our study,

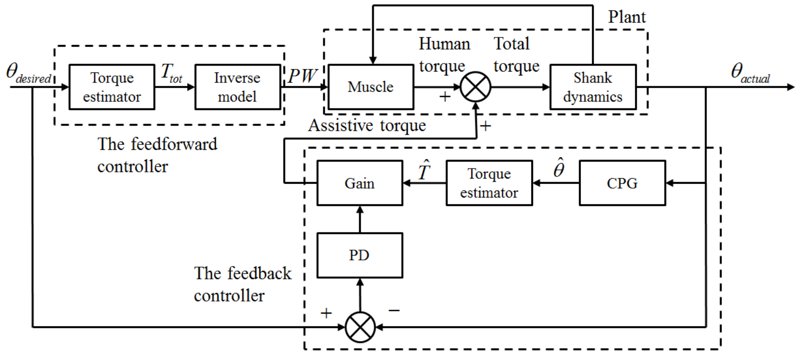

Fig. 3. Block diagram of the whole system. Each box is detailed in the paper

FES will replace healthy people's active motion and an inverse model is used as a feedforward controller to modulate stimulation intensity (pulse width) that muscles need. We choose the Matsuoka model as a feature extractor, which is a widely-used CPG algorithm. The estimated total torque can be predicted via a torque estimator and the assistive torque produced by exoskeletons is part of that. In the model the torque estimator is calculated by an inverse dynamical function of a single joint, which has been expressed in (7). A proportional-derivative (PD) controller is introduced to adaptively control the gain. Each part of the control architecture will be detailed in the paper.

The Feedforward Controller. To perform a desired joint angle trajectory, a feedforward controller based on the inversion of the direct model is an effective method. Firstly the total torque required can be computed by (7) and the passive torque is neglected in the inverse model to obtain an ideal pulse width output. The activation can be given by [10]:

$$Act = T_{tot} \cdot exp[(\frac{\theta + \pi/2 - K_1}{K_2})^2] \cdot (1 - K_3\dot{\theta(t)})^{-1} \tag{8}$$

Then the desired pulse width can be computed by the inverse recruitment function. We use the linear part to calculate:

$$PW = \frac{Act(Sat - Thres)}{Sf} + Thres \tag{9}$$

The CPG Model. A CPG has been demonstrated to exist in animals' central nervous system to regulate locomotion without sensory feedback or brain input. The most significant property of the CPG is entrainment that can synchronize the oscillation of feedback signals. The CPG can produce a robust pattern with appropriate feedback signals even in unpredictable situations. It has been widely used in controlling the locomotion of humanoid robots. There are several mathematical tools to describe CPG and we choose a model proposed by Matsuoka based on mutual inhibiting neurons. It can be given by the following equations:

$$\begin{cases} \tau_1 \dot{x_1} + x_1 = -a_{12} y_2 + s_1 - b_1 x_1' - \theta^{+}_{actual} \\ \tau_2 \dot{x_2} + x_2 = -a_{21} y_1 + s_2 - b_2 x_2' - \theta^{-}_{actual} \\ T_1 \dot{x_1}' + x_1' = y_1 \\ T_2 \dot{x_2}' + x_2' = y_2 \\ y_i = max(x_i, 0), i = 1, 2 \\ \hat{\theta} = x_1 - x_2 \end{cases} \tag{10}$$

Where x_i represents the inner state of the ith neuron, x_i' represents the degree of the adaptation, and y_i is the output (i=1, 2). Here τ_i and T_i are time constants, s_i an impulse rate of the tonic input, b_i the parameter that determines the steady-state firing rate, a_{12} and a_{21} the weight of inhibitory connection between the neurons, θ_{actual} the actual angle, and $\hat{\theta}$ the estimated angle. We use a symmetric CPG model. The parameters are set as follows: τ_1=τ_2=0.224, a_{12}=a_{21}=2.0, s_1=s_2=2.0, b_1=b_2=2.5, T_1=T_2=0.280.

The Gain Regulation. The total torque needed for knee joint can be estimated via appropriately tuning the parameters of the CPG. The assistive torque produced by the exoskeleton is a fraction of the total estimated torque, which is determined by the gain. In the human-robot synchrony experiment [6], the equivalent gain is tuned artificially, reflecting the collaboration between the subject and the assistance device. However, it is not a convenient way for practical use in hybrid FES-exoskeleton systems and we hope the assistance device can automatically compensate torque loss for the subject. Therefore we introduce a regulation rule for the gain to achieve this goal. In the control rule, a PD controller is used to minimize the tracking error between the reference knee joint angle and the actual one, which makes the system become closed-loop architecture. The adaptive gain can be computed by the following rule:

$$\mathscr{G} = \kappa(1 + K_p e + K_d \dot{e}) \tag{11}$$

where κ represents how much effort the exoskeleton should perform, and e represents the tracking error, K_p and K_d are parameters of the PD controller. We set K_p at 2.0 and K_d at 0.12 by trial and error. If there is no muscle fatigue, the torque produced by FES will be stable and we can set κ as a constant, still resulting in a quite small error. Considering the effect of muscle fatigue, the active torque will drop with time, so the error will increase and can be used to adaptively tune the gain. For practical consideration, we set κ no higher than 0.5.

3 Results

In this paper, we mainly focus on controlling single joint movements. To test the synergetic control strategy we developed for hybrid FES-exoskeleton

Table 1. Muscle-Specific Parameters

Muscle group	λ	fit_{min}	T_{fat}[s]	T_{rec}[s]
Hamstrings	0.424	0.2	25	30
Vasti	0.424	0	18	30

Table 2. Parameters of the Musculoskeletal Model

Sf[Nm]	$Thres$[μs]	Sat[μs]	K_1[deg]	K_2[deg]	K_3[s/rad]	k_{vis}[Nms/rad]	m[kg]	l[m]	J[kgm^2]
15	100	500	50	65	0.04	1.0	3.5	0.45	0.0476

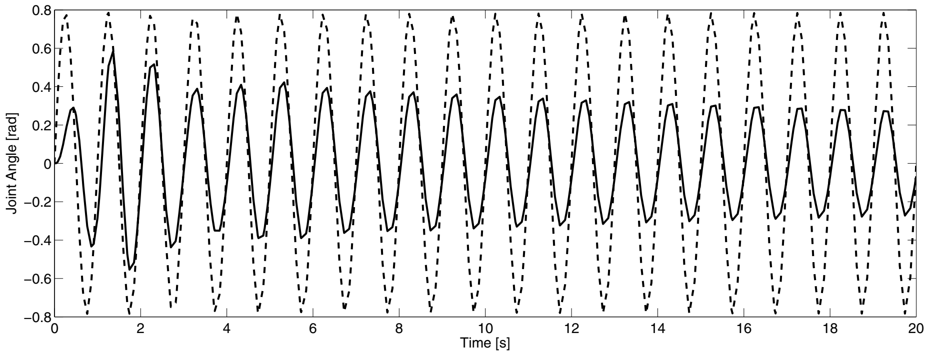

Fig. 4. Comparison between the actual joint trajectory (solid line) and the reference angle (dashed line) without any assistive torque

systems, a reference knee angle trajectory should be given in advance. For example, $\theta_{desired} = \frac{\pi}{4}sin(2\pi t)$, which means the shank oscillates at 1Hz and moves within a relatively large range. All data processing and modeling were realized in MATLAB/SIMULINK. The following tables show the relevant parameters of the musculoskeletal model. They correspond to real physiological parameters based on simplified models [9], [10].

For validating the effect of muscle fatigue, the κ should be zero at first. In the absence of exoskeletons' assistive torque, a comparison between the reference and actual knee joint angle is shown in Fig. 4. There is a significant attenuation with time because of muscle fatigue. The actual angle reached merely 50% of the originally desired trajectory after 10 seconds stimulation, which means assistive torque is highly necessary. Then we tuned κ respectively at 0.3 and 0.4 to make the assistance device work. The κ should not be tuned too high because the error in the beginning is large and then overshoot will be great. Actually even though κ is not very large, the overshoot is still unacceptable in some sense. For practical purposes, we should make the stimulation increase with time from a small value, so the hybrid system will transform to the stable state in a smooth and steady

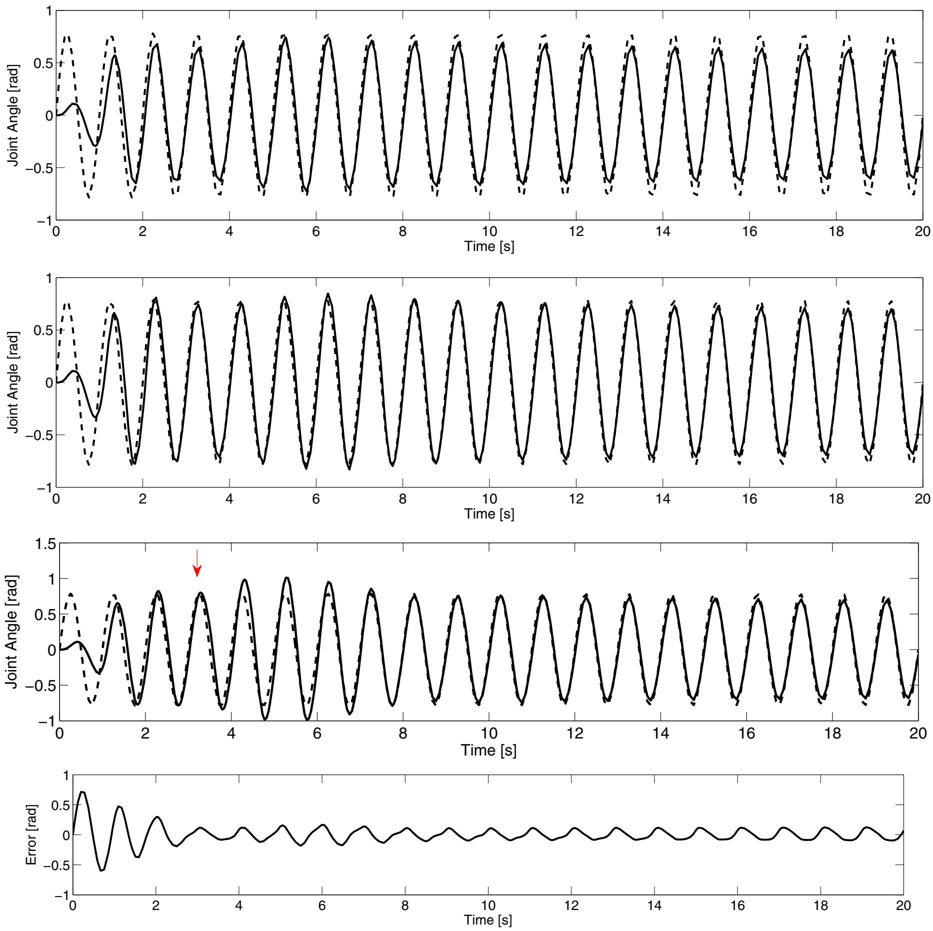

Fig. 5. Comparison between the actual joint trajectory (solid line) and the reference angle (dashed line) by using the synergetic control strategy. The κ is 0.3 for the top figure and 0.4 for the second and third. The third figure shows the joint angle after the whole system is interrupted by a torque pulse of 2Nm on the third second. The bottom figure shows the error when κ equals 0.4.

manner. Fig. 5 shows the results by using the synergetic control strategy for the hybrid system. When κ is 0.4, the exoskeleton can compensate the torque loss caused by muscle fatigue and make the knee joint move as a relatively ideal trajectory.

However if the torque loss caused by muscle fatigue is too large, there will be a slight reduction of the actual joint angle with time. The reason is that we limit the effort performed by the exoskeleton by κ. Thus in practical use, the subjects should have a rest after a relatively long time training. Besides, the CPG model extracts the pattern of joint angle with superior performance in

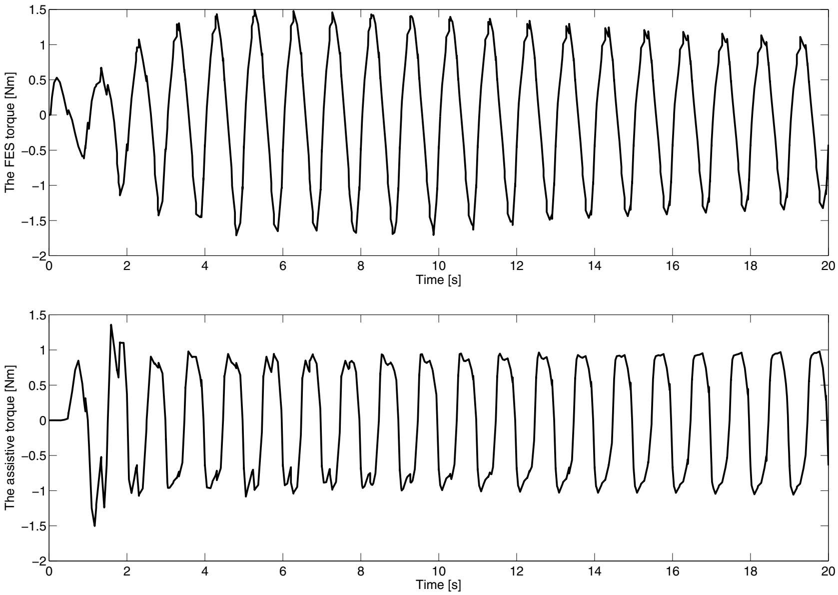

Fig. 6. The torque produced by FES (upper) and the exoskeleton (lower).

robustness. It has been testified that CPG can produce a robust motion pattern with proper feedback signals even in an unpredictable situation. Therefore, even though the real joint angle does not keep stable all the time, the output of the CPG model for torque estimator is nearly invariable owing to its entrainment property, so the relative stability of assistive torque provided by the exoskeleton can be guaranteed. We add a disturbance torque pulse into the hybrid system and the result shows basically little change for the actual angle. Fig. 6 shows the torque produced respectively by FES and the assistance device. We can see the torque generated by the exoskeleton increases to overcome the torque attenuation. The results show that the hybrid FES-exoskeleton system can be an effective way for SCI patients' rehabilitation.

We merely use a rhythmic signal to test the proposed method. Actually in daily life, the movements of lower limbs are usually rhythmical, so the control strategy presented in this paper will have practical use in rehabilitation engineering. A lot of previous work mainly focused on controlling FES to produce a desired trajectory for lower limb joints. The simulation results show that the exoskeleton as an assistance device can be used as an alternative way to make up some intrinsic drawbacks of FES.

4 Conclusion

In this paper, we have proposed a novel control strategy for hybrid FES-exoskeleton systems. A mathematical model of the Hill-type musculoskeletal system is employed to test this method. The inverse model of the stimulated muscle is adopted as a feedforward controller to generate modulated pulse width signals for FES. The CPG based on Matsuoka model acts like a feature extractor to track actual motion patterns of knee joint. Finally a PD controller is used for dynamically regulating feedback torque. The simulation results show that the control structure is efficient to regulate rhythmic movements of the knee joint. This method will bring benefit to ameliorating FES-aided movements for SCI patients.

In the future, this method can be extended to control 2-DOF hybrid systems including the hip joint. However motion coupling between two joints will be a hard problem to deal with, which needs more complicated CPG networks and optimization algorithms to design the gain regulation rules. Besides, our work proposed in this paper implemented mathematical simulations based on simplified physiological models. For practical use, real experiments based on the control strategy need to be conducted to demonstrate its actual efficiency. Our future work will focus on experimental setup and execution of the swing of shank. A hybrid system that combines FES with a knee exoskeleton will be developed, and in the early stage, experiments for this will be done on healthy subjects. After that, we will consider clinical experiments and application involving paraplegic patients.

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