行政院國家科學委員會補助專題研究計畫成果報告

下半身癱瘓病患功能性電刺激踩踏電動輔助復健輪之設計與控制

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計畫主持人：章良渭
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執行單位：國立台灣大學工學院醫學工程學研究所

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下半身癱瘓病患功能性電刺激踩踏電動輔助復健輪之設計與控制

Design and Control of an Electrical Assist Wheelchair Pedaled by Paraplegics through Functional Electrical Stimulation

計畫編號：NSC 89-2614-E-002-009

執行期限：89 年 8 月 1 日至 90 年 7 月 31 日

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中文摘要

本研究期間為二年，旨在完成功能性電刺激踩踏電動輔助復健輪椅之設計與控制，針對下肢肌肉神經傳導障礙的患者進行電刺激，前輪又加一個輔助電動馬達，做為上坡及病患肌肉疲乏時的輔助推動力，除了能達到週期性的踩踏復健運動，也能增加復健功能之多樣性與趣味性。系統控制將依據復健目標調整復健果效控制，再整合輪椅性能控制，同時兼顧復健與行動之安全與舒適。另外，因此建立之完整理論架構提供充分之理論基礎，將可幫助做有系統的進一步研發之用。

第一年研究提出一結合電刺激肌肉關節模型與四連桿機構之數學模型來模擬下肢癱瘓病患電刺激輔助踩踏運動，其中肌肉關節模型可預測電刺激所引發之關節力矩，四連桿機構可根據此力矩來模擬電刺激踩車運動。基於上述之數學模型，我們施加一由系統模型推演之 PID 控制器來完成病人踩踏運動之速度控制。我們不但要求控制系統能掌控踩踏速度的動態響應，也將透過此一數學模型的理論分析與模擬結果之評估，達成整体系統之設計目標。

關鍵詞：功能性電刺激；踩踏運動；下肢癱瘓病患。

Abstract

The objective of this two-year project is to design and control of an electrical assist wheelchair FES pedaling system,. The FES is used to stimulate the paralyzed lower limbs to achieve periodic cycling exercise and a disk motor incorporated into the front wheel will assist the ambulation when the wheelchair is in a up-hill situation, the muscle fatigue sign is on, or the patient simply gets tired. The system in turn will make the patients' rehabilitation process more versatile and interesting. The system control will integrate the wheelchair performance control and the rehabilitation outcome control according to the goals so that the wheelchair will be able to serve for both rehabilitation and ambulatory purposes. This research will also establish a solid theoretical foundation, which allows for a systematic approach to the further development.

A mathematical model, which contains a Hill-type muscle-joint model and a four-bar linkage, is established in this first-year research to simulate FES-assisted pedaling movements for paraplegics. Our muscle-joint model is a modified three-factor model and is used to predict joint moment induced by functional electrical stimulation (FES). And the five-bar linkage is used to simulate pedaling movement as a result of the FES induced moment. Based on this model, we developed a PID control for patient pedaling speed control. It is required that the control should perform the desired system dynamic response. It is also our intention that the system model can be utilized in system analysis and evaluation to achieve a

systematic overall system design.

Keywords: Functional Electrical Stimulation (FES); Pedaling Movement, Paraplegics.

Introduction

The functional electrical stimulation (FES) has been applied to paralyzed muscles for restoring muscular function. The added benefits are the improvement of cardiopulmonary function [1] and muscular endurance [2]. The applications of the FES were carried out in the areas of standing, walking and cycling for paraplegics. The FES assisted stationary cycling has been a choice exercise, which enables the lower extremities of the patients to pedal and achieve physiotherapeutic goals. In our research, the cycling mobility is being added to the system and therefore a motorized tricycle is being designed such that the FES assisted pedaling movements provides paraplegics physiological benefits and the front disk motor assists the tricycle driving power for mobility purposes. The design of the control system involves two parts, namely, the FES pedaling movement control and the cycling speed control. The FES pedaling movement control will be first designed, the strategy of which can be speed control, force control or hybrid control such that the patients have the best total rehabilitation results. The controlled pedaling movements will then be incorporated into the disk motor control to design the cycling speed control accordingly for the tricycle motion. The tricycle will be custom-made to suit the individual. An individually identifiable system model of the FES induced pedaling movements is a keystone to the design and control of the tricycle system.

The system model should include the pedaling dynamics and the models of FES muscles. Hull *et al.* (1985) took the bicycle pedaling system as a five-bar linkage and then discussed the joint reaction forces and moments from motion analysis [3]. Fang (1999) also adopted a five-bar linkage to present preliminary results on the simulation of the FES assisted pedaling movements [4]. Durfee *et al.* (1994) modified Hill-type muscle model and showed the capability of their model in predicting FES-induced force and torque [5]. A non-invasively identifiable muscle model for the FES application was thereby developed by Chizeck *et al.* (1999) [6].

The control of the FES stationary cycling exercise was performed using non-model based control via clinical experiments directly. Petrofsky *et al.* and Chen *et al.* applied PD control and fuzzy logic control respectively to deal with the feedback pedaling speed control [7-8]. There was no system modeling mentioned in their papers and the reported results and discussions were limited to the findings of their experiments. The design and control of the system has been analyzed and discussed more thoroughly by Schutte *et al.* [9] using a dynamic musculoskeletal model, which includes a four-bar linkage model and a Hill-type muscle model. However, the model requires an extensive amount of musculotendon parameters and their values are not readily identifiable from the patients.

The objective of this first-year research is to establish a continuous-time nonlinear dynamic model for the FES pedaling control system. The model enables the analysis and design of the pedaling movements in order that the biomechanical system can be developed such that the patient gains the maximum rehabilitation benefit. A part of the parameters in the model are drawn from patient's anthropometric data. The patient will be requested to perform muscle testing from which the rest of the system parameters can then be identified

Mathematical Model

A mathematical model was developed to simulate FES-assisted pedaling movements for paraplegics. A muscle-joint model represents the production of joint moments caused by the FES induced muscle contraction. Pedaling dynamics describe how the joint moments affect the pedaling movements. This model will serve as a research tool for design and control purposes.

Muscle-Joint Model

Decomposing muscle behavior into three factors, namely, activation, force-length, and force-velocity, has been widely accepted in Hill-type muscle models. For pedaling applications, it is more convenient to use moment-angle and moment–angular velocity instead of force-length and force-velocity since the joint angle and joint velocity can be obtained from sensors like goniometers. This paper presents a modified Hill-type FES muscle-joint model (Fig. 1).

Activation dynamics are combining the recruitment level and activation dynamic response. The activation output was formulated [9] as

Activation output

=(Activation level)(Activation dynamic response)

i.e., $$O=\left(\frac{1}{1+Ae^{-BI}}\right)\left(te^{-t/\lambda}\right), \tag{1}$$

I is a current input, O is an activation output and A, B and λ are the model parameters. The moment-angle factor was defined as the maximum isometric joint moment in relation to the joint angles. The model was linearized with respect to the joint angle (θ_{j}), i.e.,

$$M_{\max} = C + D\theta_{\mathrm{j}} \tag{2}$$

where C and D are model parameter. The total joint position related moment (M_{jp}) was represented as the product of the activation output and maximum isometric joint moment, i.e.,

$$M_{\mathrm{jp}} = O \cdot M_{\max} \tag{3}$$

The total joint moment (M_{j}) induced by the FES was modeled as the product of the total joint position related moment and the joint velocity related moment (M_{jv}), i.e.,

$$M_{\mathrm{j}} = M_{\mathrm{jp}} M_{\mathrm{jv}} \tag{4}$$

where the joint velocity related moment (M_{jv}) were linearly formulated with respect to the angular velocity ($\dot{\theta}_{\mathrm{j}}$) [6] as

$$M_{\mathrm{jv}} = 1 + E\dot{\theta}_{\mathrm{j}} \tag{5}$$

E is a model parameter.

By taking the knee joint for example, Eq. (4) represents the knee joint moment M_{knee} induced by the stimulated quadriceps provided that the stimulated current, knee joint angle, and knee joint angular velocity are known. The model parameters will be identified by a series of muscle testing, e.g., isometric and isokinetic muscle testing.

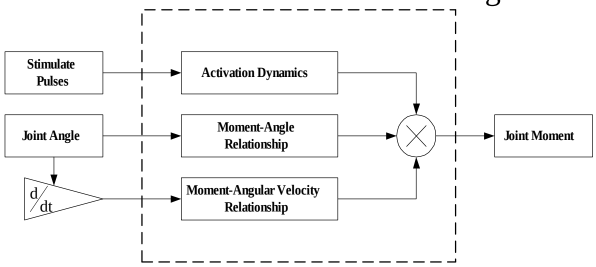

Fig. 1 The three-factor muscle-joint model used to predict the FES-induced moment

Pedaling Dynamics

We adopted a five-bar linkage being composed of right thigh, right shank, left thigh, left shank, and the crank in the sagittal plane (Fig. 2) to simulate the pedaling movements of a patient [4]. Each joint was treated as a simple hinge and the hip joint was considered to be a fixed hinge for the pedaling analysis. We neglected friction in each joint and assumed the foot and the shank to be one rigid body to eliminate the system complexity.

The equation of motion for the pedaling dynamics was written as Eq. (6). $\ddot{\theta}_{\mathrm{c}}$ is the angular acceleration of the crank. U is an inverse of a moment of inertia equivalent function of the pedaling linkage geometry. V is an acceleration function including centripetal, Coriolis, and gravitational accelerations. M is the resultant moment at each joint.

$$\ddot{\theta}_c = U_{hip}^{right} M_{hip}^{right} + U_{hip}^{left} M_{hip}^{left} + U_{knee}^{right} M_{knee}^{right} + U_{knee}^{left} M_{knee}^{left} + U_{ankle}^{right} M_{ankle}^{right} + U_{ankle}^{left} M_{ankle}^{left} + V \quad (6)$$

In our FES application, we chose the quadriceps of a paraplegic patient to be stimulated and only knee joint moment is generated by FES. From a reduced Eq. (7), forward dynamics were performed and the pedaling movements were predicted when the knee joint moment was given.

$$\ddot{\theta}_c = U_{knee}^{right} M_{knee}^{right} + U_{knee}^{left} M_{knee}^{left} + V \quad (7)$$

Since this closed chain linkage has one degree of freedom, the knee joint kinematics were obtained by knowing the crank kinematics and sent to our FES muscle-joint model for feedback purposes.

Fig. 2 A four-bar linkage system was used to simulate pedaling movements (sagittal view).

Control System

A model-based PID control was also designed in this research, which calculates required stimulation current for producing the M_{knee} to propel the pedaling system in a desired speed (Fig. 3). We defined the system error $e = \dot{\theta}_r - \dot{\theta}_c$ where $\dot{\theta}_r$ is a desired speed of the crank. The control law was designed by utilizing the system model and is given as follows,

$$M_{knee} = K_p e(t) + K_i \int_0^t e(t)\,dt + \left(K_d - \frac{1}{U_{knee}}\right)\dot{e}(t) - \frac{V}{U_{knee}} \quad (8)$$

The system error dynamic was derived as

$$K_d \dot{e}(t) + K_p e(t) + K_i \int_0^t e(t)\,dt = 0 \quad (9)$$

where K_p, K_i, and K_d are the gains of the PID control. The system dynamic response can be regulated by adjusting the system damping ratio ζ and the system natural frequency ω_n. However, the pedaling system is sometimes limited in the joint moment supply and the above control design should be further studied. The system current input can then be derived using the inverse of the FES muscle-joint model, i.e.,

$$I_{n+1} = ln\left(\frac{(\lambda/e) M_{max} M_{jv}}{(M_{knee} - M_{res})} - 1/A\right) \Big/ (-B) \quad (10)$$

where

n = pulse number in a pedaling cycle

$$M_{res} = (C + D\theta)(1 + E\dot{\theta}) \left\{ \sum_{i=1}^{n} \left(\frac{1}{1 + Ae^{-BI_i}}\right)(t - t_i)\left(e^{-(t-t_i)/a}\right) H(t - t_i) \right\}$$

$H(t - t_i)$ = Unit step function

$t_i = \frac{i-1}{f}$

f = stimulation frequency

Note that Eq. (10) was derived based on the following Eq. (12) assuming time constant of the muscle impulse response as Eq. (13).

$$M_{knee} - M_{res} = (C + D\theta)(1 + E\dot{\theta})\left(\frac{1}{1 + Ae^{-BI_{n+1}}}\right)(t - t_{n+1})\left(e^{-(t-t_{n+1})/a}\right) \quad (12)$$

$$a = t - t_{n+1} \quad (13)$$

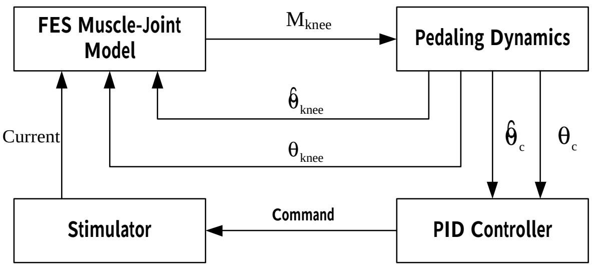

Fig. 3 A block diagram of the pedaling control system

Conclusions

The system model and control were developed to pave a road for the analytical design and control of the pedaling system. This study was adopting an identifiable model in order to develop custom-designed devices for the patients. Preliminary results have shown that the overall system design can be systematically achieved with the system model. In the second year, the project is to identify the system parameters, validate the system model and conduct clinical trials. There will be four experiments. The first is to identify parameters contained in the system model through the FES muscle testing on a dynamometer. The next is to validate the system model by stationary FES pedaling according to the designed stimulation pattern. Thirdly, the patient will be requested to pedal with our optimal seat configuration in order to assess the personalized design and control of the FES-assisted pedaling system. The last is to accomplish speed control of a motorized tricycle as the patient is performing FES-assisted pedaling at the same time so that we can evaluate the practicability of the whole wheelchair system.

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