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A Novel Approach in Development of Dynamic Muscle Model for Paraplegic with Functional Electrical Stimulation

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Abstract: This paper presents the development of paraplegic muscle model with Adaptive Neuro-Fuzzy Inference System (ANFIS). A series of experiments using Functional Electrical Stimulation (FES) with different stimulation frequencies, pulse width and pulse duration to investigate the impact on muscle output torque are conducted. The data that is obtained is used to develop the paraplegic muscle model. 500 training data and 300 testing data set are used in the development of muscle model. The muscle model thus developed is validated with clinical data from one paraplegic subject and in comparison with two other muscle models from previous researchers. The ANFIS muscle model is found to be the most accurate muscle model representing paraplegic muscle model. The established model is then used to predict the behaviour of the underlying system and will be used in the future for the design and evaluation of various control strategies.

Keywords: Functional electrical stimulation, muscle model, ANFIS, paraplegic, spinal cord injury, muscle torque..

INTRODUCTION

Paraplegic is impairment in motor and/or sensory function of the lower extremities. It is usually the result of spinal cord injury (SCI) which affects the neural elements of the spinal canal. In United Kingdom (UK), incidents of SCI are 10 to 15 per million people per annum, amounting to 600 to 900 new cases per year (Swain and Grundy, 2002). Sisto et al. (2008) reported that more than 200,000 people in the United States (US) suffer from SCI and each year 10,000 new cases occur.

Functional electrical stimulation (FES) is a means of producing contractions in muscles, paralysed due to central nervous system lesions, by means of electrical stimulation. However, to find suitable electrical stimulation values to be applied to the paraplegic is not an easy task since experiments are sometimes burdensome for the participating subjects and are generally time consuming. In general, one can gain more insight into the system and save a lot of time by computer modelling and simulation. Specifically, the performance of control strategies, system stability and other can be studied using computer simulation without involving the actual system provided the model accurately represents the real data. However, an important step required before the implementation of the movement synthesis and associated control strategy is the development of muscle model. Therefore, this paper presents the development of dynamic muscle model to represent the actual muscle behaviour of paraplegic and it will be used in computer simulation before FES can be applied practically.

Many muscle models have been developed. The earliest and most popular is the Hill model. This model is composed of three elements in which two are arranged in series, an elastic element E_S which represents the mechanical isometric response of the muscle and contractile element E_C which represents the active force generating capacity derived from chemical free energy stores. The third element is elastic element E_P joined in parallel with the other two series elements, and is added to account for the resistance of passive muscle stretch. These three elements are defined in terms of force-length property and force-velocity property (Vignes, 2004).

Since the introduction of Hill model, various modifications have been made to increase the muscle model accuracy. Zajac et al. (1986) introduced the tendon connection and accounted for the muscle fiber pennation angle in his model. Ferrarin and Pedotti (2000) developed a model that is capable of relating electrical stimulus to dynamic joint torque. The optimal model is described by a simple one pole transfer function that relates the stimulus pulse width and active muscle torque that was identified by means of parametric approach that considered the family of autoregressive with exogenous input (ARX) models and using least squares method on the error between the real data and the output of the model. More complex models have been developed by researchers (Riener, 1996; Riener, 1998 and Ferrarin, 2001) to increase the model accuracy, describing the physiologically based interpretation that capture activities under microscopic and macroscopic level such as muscle fatigue, calcium dynamics and cross bridge interaction. They (Riener, 1996; Riener, 1998 and Ferrarin, 2001) introduced a muscle model composed of three parts, activation dynamics, contraction dynamics and body segmental dynamics. Activation dynamics provide the activation needed by the muscle to generate force. It is computed as a function of pulse width and frequency with first order relation and includes the effect of muscle fatigue by introducing the fitness function and a linear second order calcium dynamics (Riener and Fuhr, 1998).

Makssoud et al. (2004) developed a muscle model composed of two parts, activation model and mechanical model. The activation model depends on the parameter of the stimulation intensity, pulse width and frequency whereas the mechanical model deals with the mechanical behaviour. The model developed is based on physiological operation condition through the implementation of macroscopic muscle model designed by Hexley (1957) who provided an explanation of the interaction of cross bridge phenomena and thus can be linked to the microscopic muscle model introduced by Hill (1938). The drawback of Makssoud et al. (2004) muscle model is that the important component of physiological based muscle model such as muscle fatigue and calcium dynamics are not accounted. Therefore, this research will develop a new muscle model from a series of experimental data, then the model developed will be compared and validated with the experimental data and two other known muscle models developed by Ferrarin (2001) and Riener (1996, 1998).

MATERIALS AND METHODS

Subjects: There are 8 paraplegics involved in data collection while 1 male paraplegic subject age 45 years old participated for model validation. The injury level of model validation subject is T2/T3 and subject having the lesion for 29 years. Ethical approval for the experiments was given by the ethic committee of the University of Sheffield.

Apparatus and setup: Throughout the experiments, paraplegic subject is placed in a semi-upright sitting position (45º to 60º) with the thigh hanging using thigh support on a frame to avoid any constraint on the leg movement. Velcro straps are used to stabilise the subject's upper trunk, waist and thigh. The isometric force output of the quadriceps muscle is recorded via a force transducer (PCE-FM200, PCE Group Company, Deutschland) placed aligned with the anterior aspect of the leg, about 5cm proximal to the lateral malleolus. The position of the leg is recorded instantaneously using Matlab software through analogue to digital converter (ADC) card and serial connection. The force and torque are also recorded simultaneously.

Data collection: Electrical stimulation is delivered via two MultiStick™ gel surface electrodes (Pals platinum, Axelgaard Mfg. Comp, USA, 50mm x 90mm). The cathode is positioned over the upper thigh, covering the motor point of rectus femoris and vastus lateralis. The anode is placed over the lower aspect of thigh, just above patella. Prior to each test, the electrodes are tested for suitable placement on the muscle by moving the electrode about the skin over the motor point, looking for the maximum muscular contraction using identical stimulation signal for the entire trials. A RehaStim Pro 8 channels (Hasomed GmbH, Germany) stimulator receives stimulation pulses generated in Matlab software through USB connection for application to the muscle.

More than 600 simulation pulses with simulation frequencies, pulse widths and pulse durations varying from 10Hz to 50Hz, 200µsec to 400µsec and 1sec to 5sec respectively are used to develop the muscle model. 731 data points are obtained from the experiments, of which 500 data points are used as training data set while 300 data points some overlapping with training data, are used as testing data set.

Adaptive Neuro-Fuzzy Inference System (ANFIS): High performance FES control systems depend greatly on the accessibility of an accurate muscle model. However, muscle model is known as a highly complex, time varying and nonlinear dynamic system. Therefore, it is necessary to develop the muscle model using an approach that can cope with the complexity and uncertainty of the model. Adaptive Neuro-Fuzzy Inference System (ANFIS) is well known by its ability to undertake this of problems.

ANFIS architecture consists of five layers with the output of the nodes in each respective layer represented by $O_{i,l}$ where i is the ith node of layer l (Denai, 2007; Mahfouf, 2004).

Layer 1: The membership function layer. The output of any node in this layer gives the membership degree of an input (crisp);

$$O_{1,i} = \mu_{A_i}(x), \qquad i = 1,2 \ or \qquad (1)$$
$$O_{1,i} = \mu_{B_{i-2}}(y), \qquad i = 3,4$$

where x (or y) is the input to the node and A_i (or B_{i-2}) is the fuzzy set associated with this node such as the generalized bell function

$$\mu_{Ai}(x) = \frac{1}{1+\left[\left((x-c_i)/a_i\right)^2\right]^{b_i}} \qquad (2)$$

where $\{a_i, b_i, c_i\}$ is the parameter set referred to as premise parameters.

Layer 2: Multiplication layer. Every node here multiplies the inputs of membership degrees and produces the firing strength of the rule or the degree in which the corresponding rule is fired;

$$O_{2,i} = w_i = \mu_{A_i}(x) \times \mu_{B_i}(y), \quad i = 1,2 \qquad (3)$$

Layer 3: Normalization layer. It calculates a ratio of the particular rule-firing degree to the sum of all rule degrees;

$$O_{3,i} = \bar{w}_i = \frac{w_i}{w_1+w_2}, \; i = 1,2 \qquad (4)$$

Layer 4: This layer applies Sugeno's processing rule and is therefore an output calculating one;

$$O_{4,i} = \bar{w}_i f_i = \bar{w}_i(p_i x + q_i y + r_i) \qquad (5)$$

where $\{p_i, q_i, r_i\}$ is the consequent parameters

Layer 5: Calculates the overall output as the sum of all incoming signals;

$$O_{5,i} = \sum_i \bar{w}_i f_i = \frac{\sum_i w_i}{\sum_i w_i} \qquad (6)$$

Hence, the artificial neural network (ANN) architecture has been built so that it can operate exactly like a Sugeno-type fuzzy system.

The network architecture for muscle model consists of four inputs and one output. The parameters of choice as inputs must have an influence on the desired output. For this model, stimulation frequency, pulse width, pulse duration and stimulation time were selected to be the model inputs. These inputs and output are used to induce an ANFIS muscle model. The time is included since the muscle torque has a significant influence on the stimulation time since muscle fatigue occurs when the stimulation time increases.

RESULTS

The performance of the ANFIS muscle model obtained after training was evaluated using the testing data set and validated with experimental results and two other muscle models developed previously by other researchers. 500 data points are used for training and 300 data points for testing purposes. Figure 1 shows the training data used in this paper. Then, the muscle model is validated with experimental data from a subject and with muscle models developed by Reiner and Ferrarin. The muscle model developed by Riener (1996, 1998) and Ferrarin (2001) has been used considering all the parameters calculated based on the same subject.

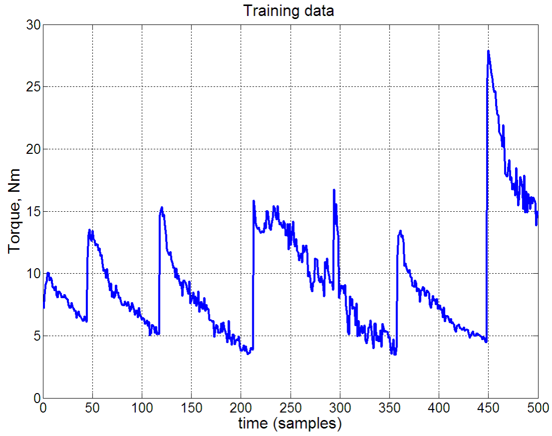

Figure 1: Training data set

2500 iterations were used to train the network. It is found that the network converged around 220th iterations. Figure 2 shows the convergence curve during the training of the muscle model network. After 250th iterations, there were no further improvement on the training error and it is assumed that the network already conversed to global optimum.

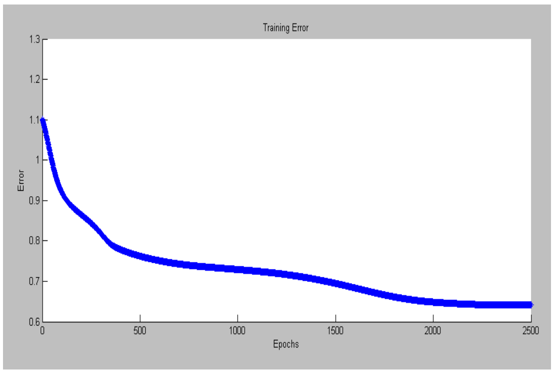

Figure 2: Convergence curve for ANFIS

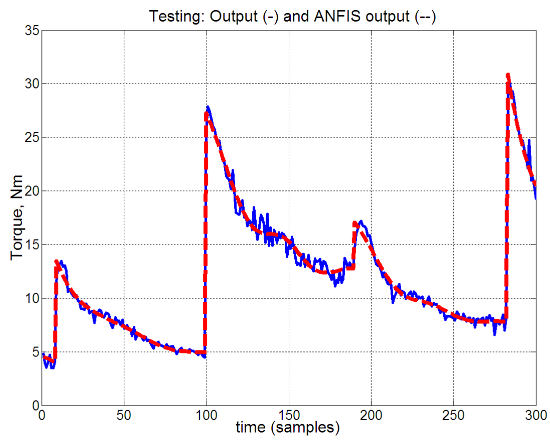

Figure 3: Output from the testing data set

After the training process, 300 data points were used to test the network. Figure 3 shows the output data from ANFIS muscle model and actual data from the experiment with the testing data. It is found that the output from ANFIS followed the actual data accurately. The variation of the experimental data is believed to come from the recording instruments; ANFIS output followed the mean of the output data variations. Figure 4 shows the prediction error of the ANFIS muscle model from the testing data set. 99% of the error lies between ±2 Nm, thus the actual muscle model represented very accurately.

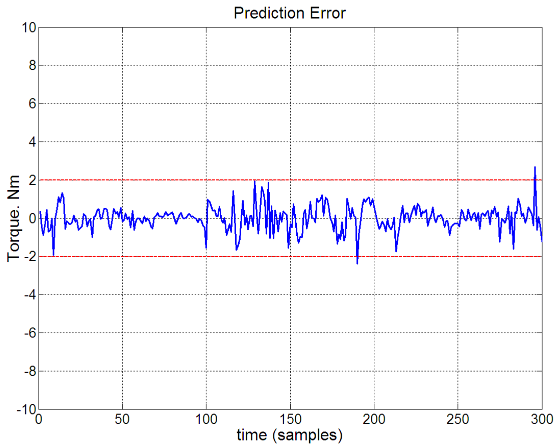

Figure 4: Prediction error from the testing data set

After ANFIS muscle model is tested, the validation of the network with experimental data obtained from a paraplegic subject and two other muscle models that have been developed using characteristics of the same subject was carried out. One male paraplegic subject aged 45 years participated to validate this model. The injury level of the subject was T2/T3 and the subject has had the lesion for 29 years. In the validation process, the stimulation frequency, pulse width, pulse duration and current of 30Hz, 250μsec, 3 seconds and 40mA respectively were used. Figure 5 shows the muscle torque produced by the stimulation parameters described.

The ANFIS muscle model was found to be most accurate model representing the actual muscle model while Riener's and Ferarrin's muscle models were found to produce lower muscle torque but slowly decreasing by end of stimulation to embody the passive elements of the leg. The damping in Riener's muscle model is found to be quite low at the end of the stimulation making the leg swing higher and longer than expected and this behaviour is not represented in the actual paraplegic leg since paraplegic leg has a shorter leg's muscle. Ferrarin's muscle model has decreased the muscle torque gradually after end of stimulation. This will make the leg slowly settle and have the second longest leg swing after Riener's muscle model.

The ANFIS muscle model follows the actual data from experiment except for one second after stimulation provided to the muscle. The ANFIS muscle model response rose immediately to 14Nm while the actual data gradually rose after two seconds. This will result the leg to extend more than actual in the first second of stimulation. However, the leg extended is about 10º more compared to the actual angle which is still within acceptable range. Furthermore, during this validation process, ANFIS muscle model followed 90% of experiment data, making it a more accurate model compared to the other two muscle models developed previously.

It is found that disadvantages of Riener and Ferrarin's muscle model to be implemented in FES application are these muscle models consist of parameters that have to obtain from the subject involved. These parameters are difficult to measure and need special instruments. Moreover, these muscle models have been developed from normal subjects and few paraplegics make these muscle model did not fully represent paraplegic muscle model.

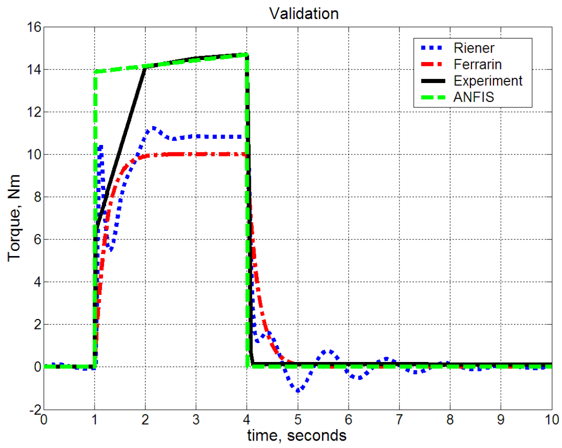

Figure 5: Response of ANFIS, Riener and Ferrarin muscle models.

CONCLUSION

An ANFIS muscle model has been developed using training and testing data set validated with experimental results and two other muscle models developed previously by other researchers. The ANFIS has been used to model and hence predict the behaviour of paraplegic muscle incorporated with FES. The performances of the three models, at their respective most optimally tuned set of parameters, has been evaluated, and the ANFIS muscle model has been found to be the most suitable and accurate model for use to determine the muscle torque.

In order to improve the accuracy of the ANFIS muscle model further, data samples from more paraplegics, will be collected and used.

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