MODELLING AND CONTROL OF PARAPLEGIC'S KNEE JOINT (FES-SWINGING)

by

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Abstract

The use of electrical signals to restore the function of paralyzed muscles is called functional electrical stimulation (FES). FES is a promising method to restore mobility to individuals paralyzed due to spinal cord injury (SCI). This thesis is concerned with the development of an accurate paraplegic knee joint model and control of electrically stimulated muscle. The modelling of musculoskeletal of paraplegic's lower limb is significantly challenging due to the complexity of the system. The first aim of this study is to develop a knee joint model capable of relating electrical parameters to dynamic joint torque as well as knee angle for FES application. The knee joint is divided into 3 parts; active muscle properties, passive knee joint properties and lower limb dynamics Hence the model structure comprising optimised equations of motion and fuzzy models to represent the passive viscoelasticity and active muscle properties is formulated. The model thus formulated is optimised using genetic optimization, and validated against experimental data. The results show that the model developed gives an accurate dynamic characterisation of the knee joint. The second aim of this study is to develop FES-induced swinging motion control. A crucial issue of FES is the control of motor function by artificial activation of paralyzed muscles. Major problems that limit the success of current FES control systems are nonlinearity of the musculoskeletal system and rapid change of muscle properties due to fatigue. Fuzzy logic control (FLC) with its ability to handle a complex nonlinear system without mathematical model is used. Two FLC strategies; trajectory based control and cycle-to-cycle control are developed. In the trajectory based control, the controller with less energy consumption is developed to reduce muscle fatigue. The ability of this controller to minimize the fatigue is proved in the experimental work. The discrete-time cycle-to-cycle control strategy is developed without predefined trajectory. This strategy is applicable for controlling FES-induced movement with the ability to reach full knee extension angle and to maintain a steady swinging of the lower limb as desired in the presence of muscle fatigue and spasticity. The performances of the controllers are assessed through simulation study and validated through experimental work.

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CHAPTER 1

INTRODUCTION

1.1 Preamble

The brain is the most complex organ in the human body and is the centre of the nervous system. It produces every human thought, action, feeling and experience of the world. The spinal cord is the pathway for impulses from the brain to the body as well as from the body to the brain. However, the bounty of this pathway could be lost due to spinal cord injury (SCI) and that results in a loss of function especially mobility. The effects of SCI depend on the type of injury and the level of the injury. For example, paraplegics are those individuals with impairment of both lower extremities but have full use of their arms and hands.

Individuals with SCI do not easily regain their function without the help from an artificial device. The improvement or restoration of lost function is one of the major issues in the rehabilitation of people with neurological disorder such as those caused by SCI (Popovic and Sinkjaer, 2000). One technique is called functional electrical stimulation (FES), and can be used in rehabilitation by producing controlled contractions in the paralyzed muscle. The ultimate aim for the rehabilitation of paraplegic patients is to regain their ability to walk. Walking is a cyclical movement (Veltink, 1991) and is divided into 2 phases, the stance phase where one leg is static on the ground, and the swing phase, where one leg is off the ground swinging forwards to make the next step. A prerequisite for FES-assisted walking for paraplegic is to control the swinging motion of the lower limb.

Therefore, in this study the research is mainly on the development of paraplegic's knee joint model and control strategies of FES-assisted swinging motion.

SCI also leads to susceptibility to the inactivity related diseases such as obesity, insulin resistance, type II diabetes, and coronary heart disease increases (Phillips et al., 1998). Physical exercise including the use of electrical stimulation devices, in this group of patients is essential in order to prevent diseases associated with physical inactivity (Kjaer, 2000). Furthermore, exercise is very beneficial to the muscles by strengthening the muscles and increasing their efficiency. The possibilities for exercise in people with SCIs are limited, especially to the paraplegic. One possible solution is by stimulating their paralysed leg muscles through application of FES-assisted exercise such as swinging, cycling and rowing. FES-assisted swinging leg is the easiest exercise that can be done by paraplegic without getting-off from his/her wheelchair.

1.2 Spinal Cord Injury

SCI is defined as damage or trauma to the spinal cord that results in loss of or impaired function and in turn reduced mobility or feeling. A common cause of damage to the spinal cord is trauma. Traumatic SCI may be caused by road traffic accidents, falls, sports injuries and gunshot. Road traffic accidents account for the largest cause of SCIs worldwide (Liverman et al., 2005). SCI can also be caused by so-called 'non-traumatic' SCI or disease such as Transverse Myelitis, Polio, Spina Bifida and Friedreich's Ataxia (http://www.apparelyzed.com, January 2011). Movement in the human body depends on natural electrical currents that flow through nerves connecting the brain with the limbs. At the bottom of the skull, a long bundle of nerves called the spinal cord joins the brain. Along the spine, smaller nerve bundles branch out from the spinal cord to the head, arms, trunk, and legs. The brain and spinal cord constitute the central nervous system (http://fescycling.com, January 2011).

The American Spinal Injury Association (ASIA) defined an international classification based on neurological responses, touch and pinprick sensations tested in each dermatome, and strength of ten key muscles on each side of the body (www.asia-spinalinjury.org, 2010) For example, when a person's spinal cord is severed or completely cut, it is known as complete or ASIA-A. A complete injury means that there is no function below the level of the injury; no sensation and no voluntary movement. When it is incompletely cut it is known as an incomplete making it ASIA-B. An incomplete injury means that there is some functioning below the primary level of the injury. A person with an incomplete injury may be able to move one limb more than another, may be able to feel parts of the body that cannot be moved, or may have more functioning on one side of the body than the other. The higher up the spinal cord is damaged, the less function a person would have.

The spinal cord is composed of 31 segments: 8 cervical (C), 12 thoracic (T), 5 lumbar (L), 5 sacral (S), and 1 coccygeal (Co) (Gondim and Thomas, 2009). The most vulnerable levels of injury are at the fifth through seventh cervical vertebrae, fourth through seventh thoracic vertebrae and tenth thoracic through the second lumbar vertebrae (Vaccaro et al., 1995). The positions of these points are illustrated in Figure 1.1, where each level is designated by the region in which it lies (cervical, thoracic, lumbar or sacral) and the spinal nerve number. SCI is classified according to the person's type of loss of motor and sensory function. The following are the main types of classification (www.ohsu.edu, January 2011):

- **quadriplegia** - involves loss of movement and sensation in all four limbs (arms and legs). It usually occurs as a result of injury at T1 or above. Quadriplegia also affects the chest muscles and injuries at C4 or above, and require a mechanical breathing machine (ventilator).
- **paraplegia** - involves loss of movement and sensation in the lower half of the body (right and left legs). It usually occurs as a result of injuries at T1 or below.
- **triplegia** - involves the loss of movement and sensation in one arm and both legs and usually results from incomplete SCI.
- **hemiplegia** -is a condition in which the limbs on one side of the body have severe weakness.

SCI is also associated with changes within the central nervous system that include problems of muscle activation such as weakness, hyperactive spinal reflexes, and loss of sensory function (Thomas et al., 1997).

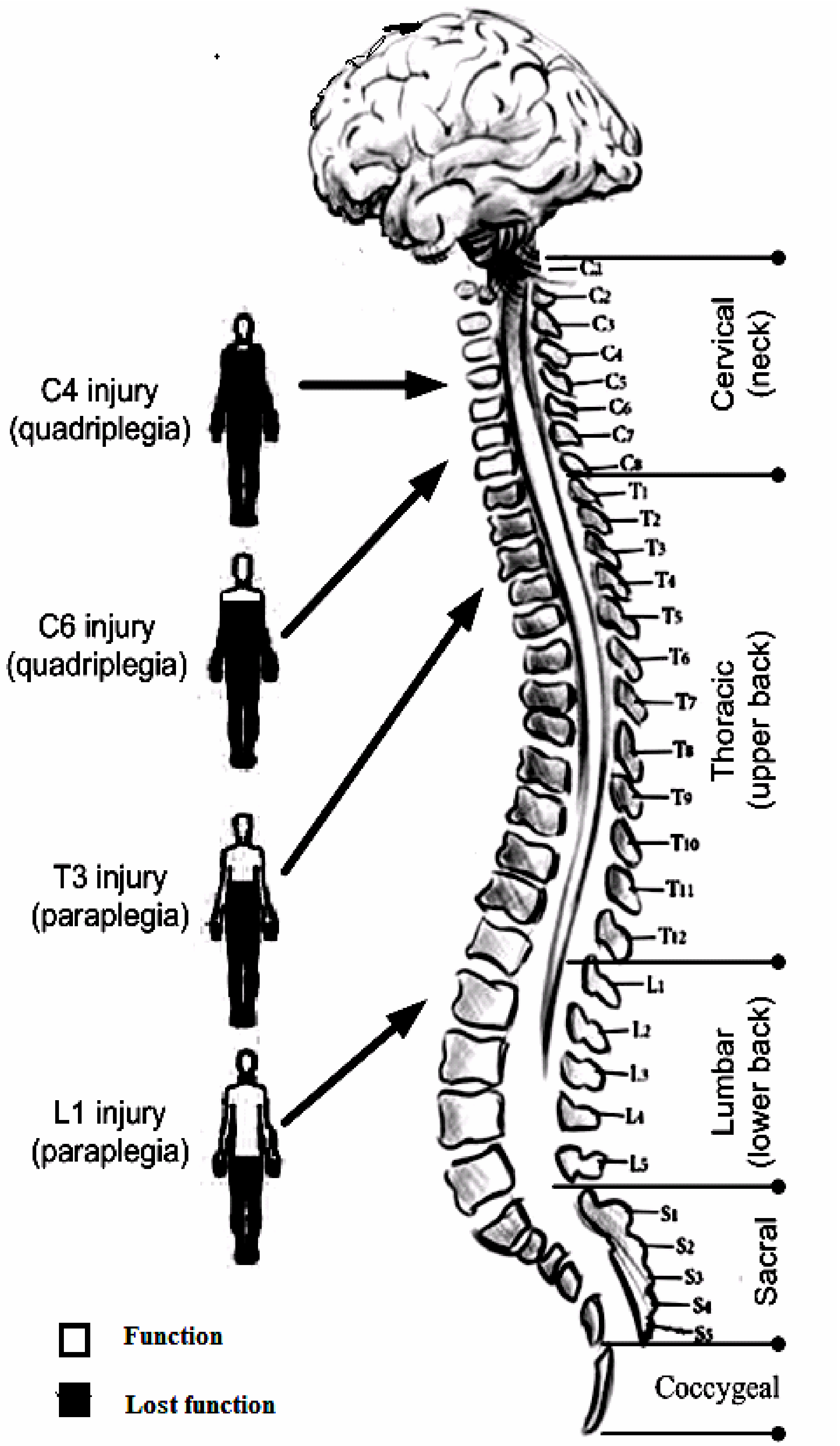

Figure 1.1: The spinal cord and areas affected at various injury levels

1.3 Functional Electrical Stimulation

In the last 50 years, medical scientists and engineers have invented methods and devices based on FES to assist with problems due to loss of mobility. The first developed FES was used as an orthotic system to prevent "foot drop" during hemiplegic walking (Liberson et al., 1961). Its use in SCI subjects with incomplete motor function, where the gait is impaired by weakness of lower limb muscles (Dietz et al., 1981; Conard et al., 1985), was first reported by Bajd et al. (1989). FES has only been available in the UK since 1995, when it has been provided by the Salisbury NHS Trust. Then in 2000, FES was named as a recommended treatment by the Royal College of Physicians for improving ankle dorsiflexion and gait performance (www.salisburyfes.com, January 2011).

FES is a promising way to restore mobility to SCI by sending electrical signals to restore the function of paralyzed muscles. In this technique, low-level electrical current is applied to an individual with disability so as to enhance that person's ability to function and live independently (Kralj and Bajd, 1989). A heart pacemaker is one example of an FES system. Other types of FES may restore lost abilities such as cycling, standing or grasping. It is important to understand that FES is not a cure for SCI, but it is an assistive device (Jacques, 1998). When there is damage only to the central nervous system, the muscle and its nerve supply remain healthy. The reason they do not work is that they are cut off from the command signals coming from the brain as explained in the previous section. FES applied near the muscle or nerve can artificially substitute electrical signals for the missing normal motor signals. The artificial impulses make the muscle contract. The main components of an FES system are the electrodes and the stimulator unit, and when FES is being used to move muscles, current pulses in the electrodes cause the weakened or paralyzed muscles to contract.

The main objective of FES in injuries to the central nervous system is the substitution of the absent bioelectric activity with an appropriately formed series of electric pulses, generated by a stimulator, or the elimination of the hyperactivity in paralysis and spastic paresis (Pasniczek and Kiwerski, 2004). The concept of FES is simple, but its realization is challenging. Two electrodes are essential to close the current circuit. For FES, and particularly when using surface electrodes; the fewer leads the better (Rushton, 1997). Figure 1.2 shows the basic electrical stimulation system.

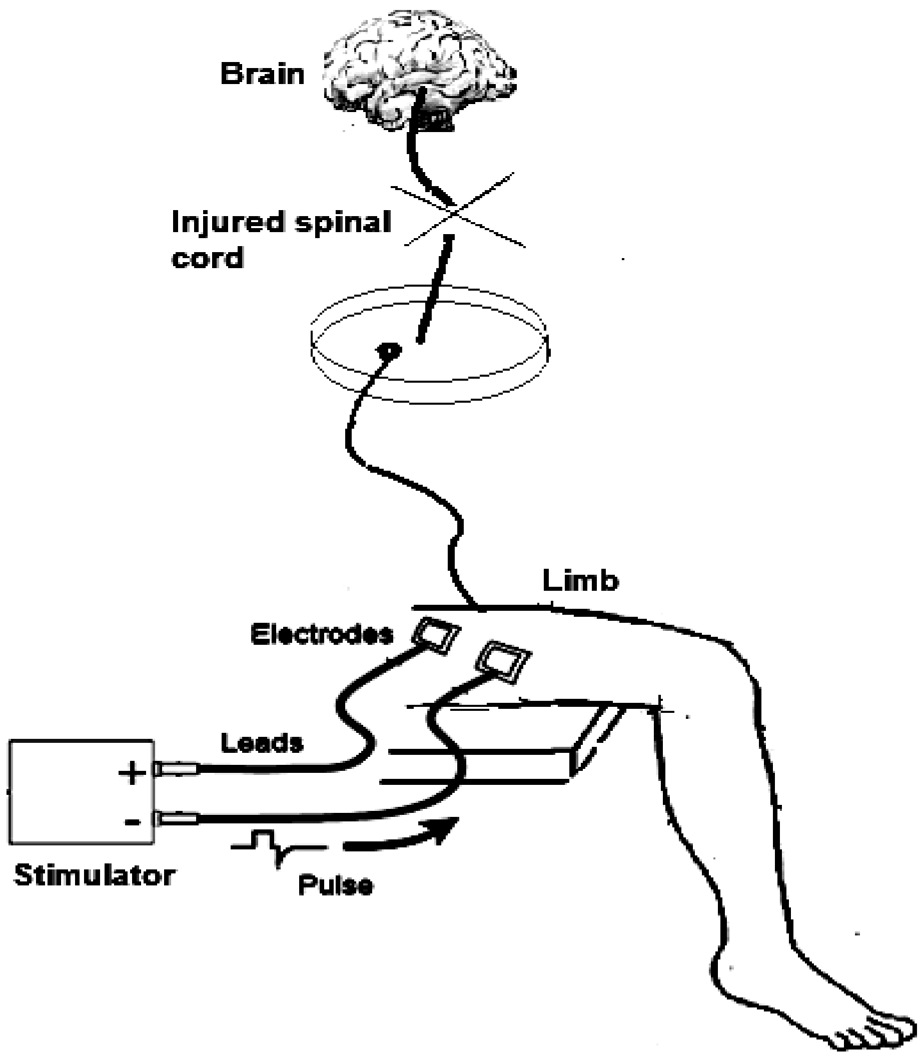

Figure 1.2: A Basic Electrical Stimulation System

Depending on the application, there are three stimulation methods: transcutaneous electrodes, percutaneous electrodes, and implanted electrodes such as epimysial and cuff **(**Popovic and Sinkjer, 2004). Transcutaneous, well known as surface electrodes, are electrodes that are placed on the skin surface and they require daily placement and removal. A percutaneous electrode means that the electrodes placed within a muscle and response to the same stimulus is likely to remain the same through the skin each day. In the past, percutaneous electrodes have failed at a high rate (Shimada et al., 2001). While epimysial is placed on the surface of the muscle and cuff is wrapped around the nerve that innervates the muscle of interest.

In the case of implanted electrodes, both the stimulator and the electrodes are surgically implanted inside the body and need another operation when the electrodes break or move. Hence the surface electrode is the most popular among people with SCI because of its simplicity and not painful method.

FES systems have been developed for restoring function in the upper extremity, lower extremity, bladder and bowel, and respiratory system (Peckham and Knutson, 2005). With regard to the lower extremity, the initial work was for correction of the drop foot. This technique is very mature now (Lyons et al., 2002). Other applications in the lower limbs include knee joint movement (Ferarrin et al., 2001; Jezernik et al., 2004; Previdi et al., 2004), cycling (Hunt et al, 2004; Petrofsky, 2003), standing up (Davoodi and Andrews, 1998; Riener and Fuhr, 1998), keeping body balance (Hunt et al., 2001), and walking (Khang and Zajac, 1989; Kralj and Bajd, 1998; Popovic et al., 199; Huq, 2010). According to a survey on FES in rehabilitation engineering conducted by Zhang and his colleagues (2007), FES has made great progress in the two applications, knee joint movement and cycling as well as many good results achieved from real experiments on subjects. However, most of the work based on activity with regard to the motor task such as standing up, keeping body balance and walking is still at the simulation stage.

1.4 Aims and Objectives of the Research

The development of paraplegic knee-joint model and control strategy for FES-swinging motion are the aims of this research. The developed knee joint model should be capable of relating electrical stimulation and knee angle specifically for FES control development. To achieve these aims, the objectives of this research are formulated as follows:

i. To develop the equations of motion of lower limb model and to optimize the anthropometric inertia parameters for a specific subject based on experimental data.

ii. To develop models of passive joint properties and active muscle properties of the paraplegic knee joint based on experimental data using fuzzy inference system and genetic optimization.

iii. To integrate the equation of motion of the lower limb, the passive joint properties and the active muscle properties to form a complete knee joint model for control development using the MATLAB/Simulink environment.

iv. To develop trajectory based FES-swinging motion control with reduced muscle fatigue within simulation and experimental environments.

v. To develop cycle-to-cycle based FES-swinging motion control, without predefined trajectory, within simulation and experimental environments.

1.5 Thesis outline

The organisation of the thesis reflects the sequence of steps taken in the development of dynamic modelling of knee joint and control schemes for FES-swinging motion. A brief outline of contents of the thesis is given as follows:

Chapter 1: This chapter describes SCI and introduces the concept of FES in paraplegia. Most importantly, the chapter has stated the aims and objectives of this research and the contributions of the thesis as well as the list of the publications arising from this research work.

Chapter 2: In this chapter a brief review relating to the musculoskeletal model such as segmental dynamics, muscle model and passive joint properties is included. The survey of literature in this chapter is intended to highlight the challenges and drawbacks related to the forward dynamic model of human lower limb for simulating FES applications.

Chapter 3: This chapter describes a new approach for modelling non-linear passive joint properties of the paraplegic. Genetic algorithm (GA) based fuzzy model is proposed to represent the passive viscoelasticity of knee joint based on pendulum test. It also describes equations of motion and optimization of anthropometric inertia parameters using GA based on subject-specific anthropometric data and experimental data.

Chapter 4: This chapter describes a new modelling method of active properties of quadriceps muscle using multi objective genetic algorithm (MOGA) with two objectives; to minimize the prediction error to fit the experimental data and the weighting factors of the fuzzy rules. A fuzzy model is used to represent the highly non-linear active properties of the quadriceps muscle.

Chapter 5: This chapter presents the development of FES-induced swinging motion control with reduced energy consumption based on ‘natural’ trajectory approach. Two fuzzy controllers; with and without energy efficiency mechanism are developed using MOGA optimisation. The controllers are assessed in simulation studies and validated through experimental work.

Chapter 6: In this chapter a discrete-time cycle-to-cycle control of FES-induced swinging motion without predefined trajectory is investigated. The control is able to achieve full knee extension angle and to maintain a steady swinging of the lower limb as desired in presence of spasticity and muscle fatigue. The developed controls are assessed in simulation studies and validated through experimental work.

Chapter 7: In this chapter the conclusions and recommendations for future work are presented.

The thesis has two appendices.

Appendix A presents steps of derivation and transformation of the double rod pendulum based on Kane equations for lower limb as discussed in Chapter 3.

Appendix B describes the muscle model from the work of Reiner and Fuhr (1998) for the comparative study as discussed in Chapter 4.

Part of the work presented in the thesis has been published through several papers.

1.6 Thesis Contributions

The original contributions made in this thesis can be outlined as follows:

i. The model of a dynamic system of the lower limb is derived using Kane's equations. The anthropometric inertial parameters such as foot mass, shank mass, moment of inertia about centre of mass (COM) and position of COM along the segmental length of the limb in these equations have been optimised using GA based on subject-specific anthropometric data (height and weight) and experimental data.

ii. A fuzzy model is proposed to estimate the non-linear passive viscoelasticity (combination of viscosity and elasticity) of knee joint properties as a function of knee angle and knee angular velocity using GA optimisation based on experimental data from pendulum test.

iii. A fuzzy model is proposed to estimate the complex active properties of quadriceps muscle using MOGA with two objectives; to minimize the prediction error to fit the experimental data and the weighting factors of the fuzzy rules. The estimated reduced rule of the fuzzy model exhibited good prediction capabilities; it is comparatively less burdened compared to the complex mathematical model.

iv. A novel method of development of FES-induced swinging motion control with reduced energy consumption based on 'natural' trajectory approach has been proposed. In these approaches, fuzzy logic controllers are optimized using genetic optimization with multi objectives. A new control scheme; controller with energy efficiency mechanism is proposed to control paraplegic knee joint with reduced energy consumption The time integral of knee torque is taken as the optimization criterion to design this controller. The control approach has shown up to 10% reduction in energy consumption within simulation studies and has shown to minimize muscle fatigue within experimental work.

v. A new control strategy is proposed for FES-swinging motion without predefined trajectory. The discrete-time fuzzy logic cycle-to-cycle control is applicable to control FES induced movement with the ability to reach full knee extension angle and to maintain steady swinging of the lower limb as desired in the presence of muscle fatigue and spasticity. The capability of fuzzy control in automatic generation of stimulation burst duration is assessed in computer simulations using a developed musculoskeletal model. Then, the performance of the controller has been validated through experimental work.

vi. Finally, the study proposes approaches involving the stimulation of only the quadriceps to obtain steady FES-assisted swinging motion. The control strategies thus developed are able to perform steady swinging motion within simulation and experimental environments.

1.7 List of publications

Technical papers arising from this study, which are either published or accepted for publication are listed in the following subsections.

1.7.1 Journal papers

B.S. K. K. Ibrahim, M.O. Tokhi, M.S. Huq, R. Jailani and S.C. Gharooni (2010). Fuzzy Modelling of Knee Joint with Genetic Optimization. Journal of Applied Bionics and Biomechanics (ABBI) (accepted).

B.S. K. K. Ibrahim, M.O. Tokhi, M.S. Huq and S.C. Gharooni (2010). Discrete-time Cycle-to-Cycle Control of FES-induced Swinging Motion (submitted).

B.S. K. K. Ibrahim, M.O. Tokhi, M.S. Huq and S.C. Gharooni (2010). FES-induced Natural Swinging Motion Control with Less Energy Consumption (submitted).

1.7.2 Conference papers

B.S.K.K. Ibrahim, R. Massoud, M. S. Huq and M. O. Tokhi (2008) Modelling of humanoid and bicycle for FES-sssisted cycling. United Kingdom – Malaysia Engineering Conference 2008, 14-15 July 2008, London (published).

B.S.K.K. Ibrahim, S.C.Gharooni , M.O.Tokhi, R.Massoud (2008) FES assisted cycling with quadriceps stimulation and energy storage, Proceeding of 11th International Conference on Climbing and Walking Robots and the Supporting Technologies for Mobile Machines, 8-10 September, Portugal, World Scientific Publishing Company, Singapore, pp. 1003 – 1010, ISBN: 13 978-981-283-576-5 (published).

B.S.K.K. Ibrahim, R.Massoud, S.C.Gharooni , M.O.Tokhi (2008). Energy-efficient FES cycling with quadriceps stimulation, 13th Annual International Functional Electrical Stimulation Society Conference, 21-25 September, Freiburg, Germany (published).

B.S. K. K. Ibrahim, M.S. Huq, M.O. Tokhi, S.C. Gharooni, R. Jailani, Z. Hussain (2009). Identification of Active Properties of Knee Joint using GA Optimization, Proceeding of International Conference on Biomedical Sciences and Technologies, 28-31 July, Norway, World Academy of Science, Engineering and Technology, 55, 2009, pp. 441 – 446. ISSN: 2070-3724 (published).

B.S. K. K. Ibrahim, M.S. Huq, M.O. Tokhi, S.C. Gharooni, R. Jailani, Z. Hussain (2009). Knee joint model identification using genetic algorithms, Proceding of 12th International Conference on Climbing and Walking Robots and the Supporting Technologies for Mobile and Machines, 9-11 September, Turkey, World Scientific Publishing Company, Singapore, pp. 1145 – 1152. ISBN: 13 978-981-4291-26-2, 10 981-4291-26-9 (published).

B.S. K. K. Ibrahim, M.S. Huq, M.O. Tokhi, S.C. Gharooni, R. Jailani, Z.Hussain (2009). A novel approach to evaluate passive elastic-viscosity and mass for lower limb, Proceedings of 4th IFAC Workshop on Discrete-Event System Design (DESDes2009), 6–8 October, Valencia, Spain (published).

B.S. K. K. Ibrahim, M.O. Tokhi, S.C. Gharooni and M.S. Huq (2010), Development of fuzzy muscle contraction and activation model using multi-objective optimisation, 4th Annual IEEE International Systems Conference, 5-8th April 2010, San Diego, USA. ISBN: 978-1-4244-5883-7 (published).

B.S. K. K. Ibrahim, M.O. Tokhi, M.S. Huq and S.C. Gharooni (2010), Modelling and control of paraplegic knee joint, proceeding of 1st annual conference of the International Functional Electrical Stimulation Society (UK and Ireland Chapter), 15-16th April 2010, University of Salford, United Kingdom (published).

B.S. K. K. Ibrahim, M.O. Tokhi, M.S. Huq and S.C. Gharooni (2010), Muscle Modelling: Comparative Study of Fuzzy and Mathematical Modelling Approaches. Proceedings of CLAWAR 2010: The Thirteenth International Conference on Climbing and Walking Robots and the Support Technologies for Mobile Machines, Nagoya, Japan, 31 August and 1-3 September 2010, World Scientific Publishing Company, Singapore, pp 1247-1254, ISBN: 13-978-981-4327-97-8 (published).

B.S. K. K. Ibrahim, M.O. Tokhi, M.S. Huq and S.C. Gharooni (2010), FES-induced swinging motion control, IFAC Workshop on Intelligent Control Systems 29-30 September - 02October 2010, Sinaia, Romania (published).

CHAPTER 2

A BRIEF REVIEW OF MUSCULOSKELETAL MODELLING

2.1 Introduction

Models of the musculoskeletal system are valuable tools in the study of human movement. Modelling, and hence a simulation study can greatly facilitate to test and tune various FES control strategies. In order to develop a control strategy for the FES to move the leg correctly, an accurate model of the stimulated muscle has to be used. Accurate models can facilitate the design of stimulation patterns and control strategies that will produce the desired force and motion (Perumal et al., 2008).

The major function of muscle is to produce force. Muscle force is not as important as it is in natural systems for FES systems (Jacques, 1998). In classical muscle mechanics, researchers usually examine and model the relationship between stimulation and force in an isolated muscle. In FES, researchers usually examine and model the relation between stimulation and angular displacement of a joint or its joint torque. The distinction is important because the limb itself adds inertia, stiffness, and damping which arise from the bone and joint but also from other muscles crossing the joint (Jacques, 1998). The properties of a joint, rather than a muscle, can dominate the relationship of stimulation to displacement (Allin and Inbar, 1986).

Modelling of joint properties of lower limbs in people with SCI is significantly challenging for researchers due to the complexity of the system. The complexity is due to the combination of complex structural anatomy, complicated movement and dynamics as well as indeterminate muscle function (Ghafari et al., 2009). Depending on the direction in which the neurophysiological and biomechanical processes are represented in the model, one can distinguish direct and inverse dynamic models. Direct dynamic models (also called forward dynamic models) are used to calculate the internal processes in the same order in which they occur in the real system. The inputs in such models are various stimulation parameters or complex stimulation patterns, the outputs are usually joint torques or limb movements.

Contrary to the direct dynamic model, an inverse dynamic model describes the processes underlying FES-induced movements in the opposite direction: the input to the model is a measured or desired movement trajectory, the model then predicts the stimulation pattern, or any other internal quantity, which is necessary to achieve the predefined movement (Wells, 1967). In the control of FES-induced single-joint movements, the use of an inverse dynamic model has been shown to be a promising strategy to control joint angle (Quintern et al. 1989; Veltink et al., 1992a). However, when a redundant system with more muscles than degrees of freedom is considered, a unique solution is no longer possible and additional optimization procedures have to be applied.

On the other hand forward dynamic models of the FES musculoskeletal system have been widely developed such as in (Delp et al., 1990; Hatze, 1980; Pandy et al., 1990; Veltink et al. 1992b). It is easy to compare the results of simulation and experiments in the forward models, as the stimulation input and output are the same for both simulation and experiment. In this chapter, various components or building blocks of the forward dynamic approach for musculoskeletal modelling are briefly surveyed. The main objective of the survey is to review a forward dynamic model and identify its drawbacks so that the modelling approach can be improved. The main components of the forward joint model consist of segmental dynamics, active properties (muscle activation and contraction) and passive properties (elasticity and viscosity) as shown in Figure 2.1.

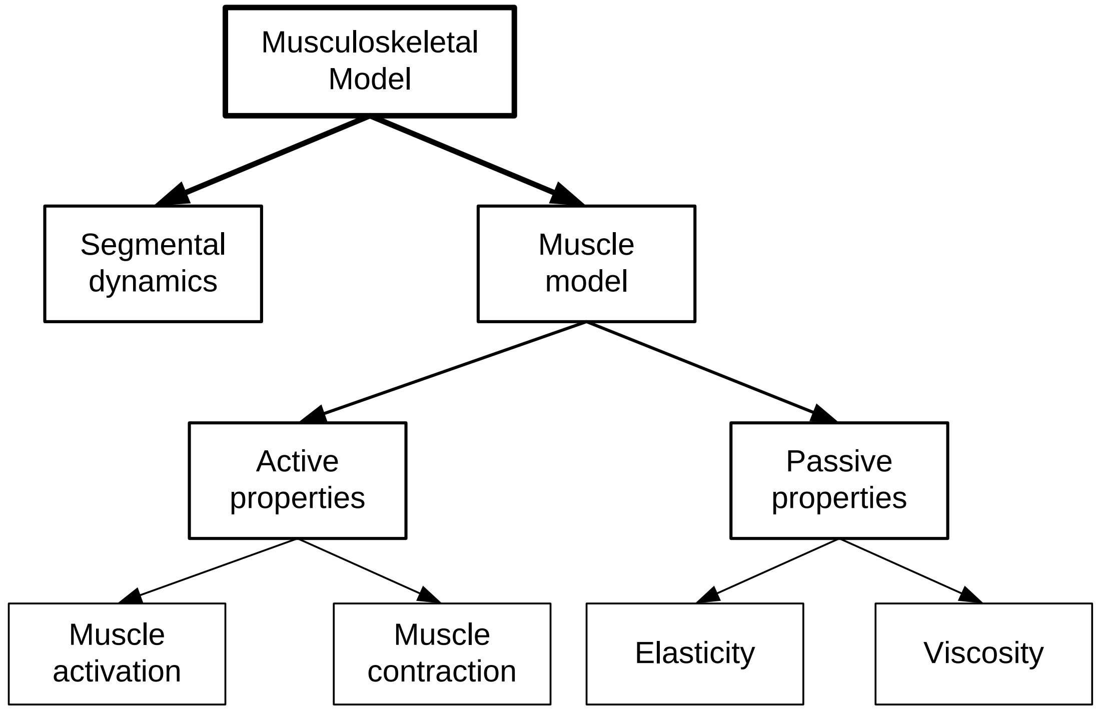

Figure 2.1: Main components of the forward dynamics musculoskeletal model

2.2 Body Segmental Dynamics

The body segmental dynamics are generally modelled as a set of rigid limb segments, whose movements relative to each other are defined by the joint articulations, i.e. skeletal geometry and ligamentous constraints (Huq, 2009). The complexity of a body-segment dynamical model depends on the number and types of body segments, the joints connecting the segments, and the interaction among the segments and the environment (Bronzino, 2006). The degree-of-freedom (DoF) of human body can indicate the different levels of difficulty. The complexity in analysing multi-joint structure is often reduced through reducing the number of DoF to a manageable level.

With regard to leg movement control, most FES applications serve for 1-DoF movement such as knee joint control and ankle joint control (Yoshida and Horch, 1996; Hunt et al., 2001). FES application for 2-DoF movement is also very common, in which the two segments are represented as thigh and shank (Franken et al., 1995; Jonic et al., 1999). For 3-DoF leg movement, the three segments are represented as thigh, shank, and foot (Liu, 2000) in a two-dimensional plane. There are some complex movements that involve more than three DoFs such as walking (locomotion). The work done by Hatze (1980) contains 42 DoFs, Yamaguchi and Zajac (1990) used 8-DoF and Anderson (1999) used a three-dimensional 23-DoF model of the skeleton to simulate one cycle of normal walking. Because these models have complex structures of too many DoFs and muscles, it is hard to control them in reality via FES (Zhang et al., 2007). At present they are only suitable for computer simulation and the experimental test is till far away

Commercial software packages for simulation of mechanical systems such as SDFAST (Symbolic Dynamics Inc., USA), ADAMS (Mechanical Dynamics Inc., USA), and Visual Nastran (MSC Software Corp., USA) have been used to model musculoskeletal systems. These software packages, however, lack the specialized components specific to biological systems, such as musculotendinous force production and musculoskeletal moment arm (Maleki and Fallah, 2006**)**. Most of the visualization software such as Visual Nastran is computationally expensive and lacks mathematical description of the model (Huq, 2009). Moreover, these methods need to gather multiple software packages and seamlessly link them to each other, that is very costly and inconvenient.

The conventional method of using mathematical modelling has been widely used as a simulation tool in biomechanical research specifically to get the dynamical equations. The usual approach of mathematical formulation can be adopted easily to write computer programs using symbolic computer languages such as LISP, PROLOG, and MAPLE and to generate the equations of motion (Winter, 2005). Computer programming languages such as C, FORTRAN, or BASIC can also be used to develop self-formulating programs generic enough to accept model description as an input and provide model response as an output (Huq, 2009). The popular computer software package MATLAB, contains a library of standard vector and matrix manipulation functions. Most of the computational models of the human limb have been developed utilizing a matrix calculation using MATLAB (Barker et al., 2007).

The use of MATLAB has ensured that an easy user interface can be achieved with little detailed knowledge of computer programming (Barker et al., 2007). The ability to use other in-built or custom-written functions within MATLAB is also advantageous for pre- or post-processing of data. Furthermore, MATLAB is very convenient for implementation of complex control strategies by using appropriate toolboxes, and real-time interaction with real world can also be established using the Real-time Workshop toolbox. Therefore by developing the entire model and control in a single software package will help the developer with the costs involved in providing several software packages and the difficulties associated with linking them together.

Huston et al. (1976, 1978) developed a general approach for studying a human body model using equations based on d'Alembert's principle. Onyshko and Winter (1980) developed a seven link planar model. Equations of motion formulated using Lagrangian mechanics, consist of a 7×7 matrix of anthropometric constants and segment angles, a vector of the angular accelerations and vector containing the torques acting on the segments. Hatze (1980) used the traditional Lagrangian approach to define a mathematical model of the total human musculoskeletal system.

Marshall et al. (1985) used a general Newtonian approach to simulate N-segment open chain model of the human body. The model simulated planar movement using data for joint torques and initial absolute angular displacements and velocities for each body segment. These values are used to solve the direct dynamics problem, expressed in the form of n simultaneous linear equations, to yield angular accelerations. Zajac (1989) has developed a planar computer model to investigate paraplegic standing induced by FES.

Kane's equations (Kane and Levinson, 1985) with the combination of computational advantages of d'Alembert's principle and Lagrange's equations have been utilized successfully in modelling of complex systems such as in Langer et al. (1987), Lesser (1992), Komistek et al. (1998), Tisell (2000) and Yamaguchi (2001). Kane's method of dynamics is a sophisticated mathematical technique that allows resolution of a large number of variables through the use of generalized speeds that define the motion in the system. Through the use of generalized speeds one is able to model the human lower extremity as a first-order set of differential equations (Komistek et al., 1998).

2.3 Muscle Models

Different types of muscle model are used for different purposes. The range extends from analytical models which are based on physical properties of the muscle, either at a microscopic or at macroscopic level, to empirical models which are purely mathematical descriptions of the input-output characteristics of the muscle (Huq, 2009). An extensive review of various modelling approaches can be found in Zahalak (1992). Aspects which are particularly relevant to modelling of artificially stimulated muscle are discussed in Durfee (1992). Winter and Stark (1987) compare model structures based on microscopic analysis, macroscopic analysis and purely mathematical models.

The most widely used microscopic model is the cross bridge model, the basic principles of which were developed by Huxley (1974). It aims to describe the muscle characteristics at a microscopic level by modelling the processes within a single muscle fibre. This type of model is in principle useful to describe all characteristics of muscle, as all model parameters are based on physical components, which makes it very popular amongst biologists. However, the microscopic approach makes a description of the entire muscle very difficult as parameters at the level of muscle fibres have to be identified. It also leads quickly to large systems of nonlinear partial differential equations which are difficult to handle. Parameters of a Huxley-type model are difficult to interpret in terms of the macroscopic muscle characteristics. A number of unconventional cross-bridge models have been suggested which make different assumptions than those of Huxley, e.g. Hatze (1974).

The most often cited macroscopic muscle model is based on the description by Hill (1938) with a large amount of experimental data and results.. The Hill-based muscle models are popular as their parameters describe familiar concepts related to the sarcomere structure and they consist of successive equations based on experimental results. This makes them more preferable than those that depend on theories as it is claimed that they are more accurate (Fung, 1981). However, for the identification of parameters of model components special experiments are necessary which may not be applicable in all situations. Winters and Stark (1987) developed the simplest model with the most apparent success of the structure being its ‘task independency’, which has been claimed to be achieved through including nonlinear relations for four fundamental model properties such as contractile element torque-velocity, serial elastic component, parallel element, contractile element active muscle torque-angle. However, this would obviously increase the model complexity.

The most cited and very comprehensive paper that addresses Hill based modelling through exploring muscle-tendon interaction is given by Zajac (1989). Hill-based models calculate the muscle force as the product of three independent experimentally measured factors, namely the force-length property, the force-velocity property and the activation dynamics of the neural input. The model has been simplified by assuming that the muscle activation and muscle contraction dynamics are uncoupled as shown on Figures 2.2. Neural excitation acts through activation dynamics (excitation-contraction coupling) to generate an internal muscle tissue state (muscle activation). Through muscle contraction dynamics this activation energizes the cross-bridges and develops the muscle force. An extensive review of various microscopic modelling approaches can be found in Huq (2009).

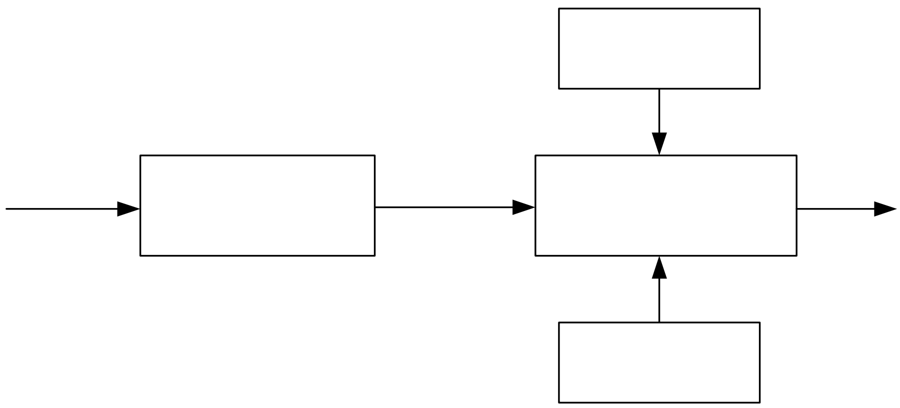

Figure 2.2 : Hill based muscle model

Most models built on analytical bases are not suitable for FES control applications (Massoud, 2007). One way to develop this model for FES control application is to use mathematical models. Thus, empirical model strategies, which aim to describe the input-output characteristics of muscle (often limited to conditions common in FES applications), and whose structure is suitable for the design of stimulation controllers, become useful. As a result, many researchers have developed mathematical models of electrically stimulated muscle based on Hill-type (Ding, 2002; Shue and Crago, 1998), Huxley-type (Zahalak and Ma, 1990), analytical approaches (Bobet and Stein, 1998, Ferrarin and Pedotti, 2000) and physiology approach (Riener and Fuhr, 1998). A review of empirical modelling approaches for muscle is given in Durfee (1992).

The use of mathematical models can significantly enhance the design and evaluation of closed-loop control strategies applied to FES (Riener, 1999). In fact, mathematical models can be used to promote an understanding of the system and they can be used to predict the behaviour of the system (Zahalak, 1992). Accurate models of artificial muscle activation in healthy or paraplegic subjects have been developed but the complexities of the system resulting mathematical representation have a large number of parameters that make the model identification process difficult.

In fact muscle consists of active and passive properties. In this study, a separation is made between the active and the passive muscle properties. So far only the active muscle properties have been considered. The passive muscle characteristics such as viscosity and elasticity are assigned to the joint as considered in (Ferrarin et al., 2001) and are included in the next section.

2.4 Passive Joint Properties

Muscles produce active forces to generate movements, but they also contain collageneous structures which hold the muscle fibres together and resist stretch even when muscle is relaxed (Fung, 1993). The joints themselves exhibit resistance to movement because of the properties of cartilage and the shapes of the contacting articular surfaces. Together all these resistive forces across a joint generate passive moment about the joint (Amankwah et al., 2004). Passive joint moments are among the properties of the musculoskeletal system involved in controlling movement such as locking of the knee at full extension.

Early studies have shown that the joint angle affects passive moments (Agarwal and Gottlieb, 1977, Wright and Johns, 1961), while other studies have shown that passive moments can also vary with the angular velocity of the joint (Duong et al., 2001; Esteki and Mansour, 1996; Hayes and Hatze, 1977). Conventionally, the joint passive resistance is modelled as an elastic element as spring and a viscous element like a rotary damper (Lamb et al., 1991). These two resistances are non-linear, but the viscous resistance is often approached as a linear function of the joint angular velocity (Chizeck et. al., 1999; Mansour and Audu, 1986).

Several groups have investigated passive elastic joint elements and implemented these in biomechanical simulation studies (Mansour and Audu, 1986; Davy and Audu, 1987; Veltink et al., 1992; Riener et al., 1996; Riener and Edrich 1999). These models have illustrated that passive structures contribute significantly to joint torque during complex movements such as walking (Amankwah et al., 2006; Mansour and Audu, 1986; Whittington et al., 2007). However, these studies take only the elasticity of the passive properties into account and, thus ignore the velocity dependence viscosity elements.

Elasticity can be considered as an intrinsic property of the tissue to resist deformation, while viscosity is related to cohesive forces between adjacent layers of tissues. Identifying the parameters of a model that includes viscosity effects is much more challenging since this would also require manipulation of the joint velocity (Vrahas, et al., 1990; Yoon and Mansour, 1982). Indeed, very few studies have modelled both the elastic and viscous components of the passive elements. Therefore, some researchers have modelled the passive viscous moment with a linear damping function (Stein et al., 1996; Ferrarin et al., 2001). However, these important characteristics of passive elements have wrongly been estimated when the nonlinearity nature of viscosity has been neglected. Thus, inaccurate modelling of this will bias the analysis of the joint.

Amankwah et al. (2004) have developed a mathematical equation to describe the passive nonlinear viscoelastic lower-limb joint moments. The developed model was based on Kelvin model with a nonlinear elastic element in parallel with both a linear elastic element and a nonlinear viscous element in series. The elastic moment, described the nonlinear passive elastic moment and was modelled with a double exponential. However, the use of exponential models require accurate kinematic measures since a small deviation in joint angle can cause a large change in the estimated joint moment, and therefore substantially impact the mechanics attributed to passive elements (Silder, 2009). The power function used to model the viscous moment in (Amankwah et al. 2004) does not agree with recent experimental results for knee joints indicating the existence of a linear relationship rather than a non-linear relationship (Nordez et al., 2008).

2.5 Whole Musculoskeletal Models

The whole musculoskeletal model consists of segmental dynamics, active and passive properties of muscle as one complete model suitable for control development. However, most currently built muscle models are not appropriate for control applications, since these models characterize each muscle feature alone, and sometimes there is no connection between the modelled features which may prevent from modelling the whole muscle as one model (Massoud, 2007). The most appropriate complete model that can be used for FES control applications have been developed by Riener et al. (1996) and Ferrarin and Pedotti (2000).

Riener et al. (1996) have developed a model of the human knee based on the physiology approach to predict the shank motion induced by functional neuromuscular stimulation. The model contains activation dynamics with slightly modified version of a discrete-time model proposed by Khang and Zajac (1989), while the contraction dynamics have been derived by scaling the generic Hill-based model. The passive elasticity and viscosity have been modelled nonlinearly through passive pendulum tests. The body segmental dynamics have been modelled as a nonlinear 1-DoF equation of motion to take into account the gravitational moments, moment of inertia and internal moment that act on the knee joint. The model has been found to perform quite well in terms of model prediction tests with passive pendulum test, isometric moment vs. pulsewidth and different stimulation patterns for freely swinging shank. However, this model requires many individualized parameters to be identified and therefore it makes the modelling process more complicated

Ferrarin and Pedotti (2000) have presented a straightforward and a simple approach to model the human knee through identification of the relationship between stimulation parameter, pulsewidth and active knee joint torque produced by the stimulated quadriceps muscles in a non-isometric condition. A mathematical model of a first order transfer function obtained by a least squares error method has been developed. The gravitational and inertial characteristics of the anatomical segments together with the damping and stiffness properties of the knee have been taken into account.

Kinematics data obtained from passive pendulum tests has been used to identify the unknown stiffness and damping parameters. The damping component has been considered to consist of a linear term with a constant viscous coefficient, while the stiffness component has been modelled nonlinearly with an exponential term. Trials of quadriceps stimulation with pulsewidth modulation and predetermined pulse-amplitude at different frequencies have been conducted to obtain the kinematics data. Thus, the obtained data has been used to compute the time course of the active knee torque by substituting pre-identified viscous-elastic components into the equation of dynamics equilibrium for the lower limb, which has then been optimally modelled as a single pole autoregressive with exogenous terms (ARX) model relating pulsewidth (input) and active muscle torque (output). However, the accuracy of this model has been criticized as in Ferrarin et al. (2001) and Massoud (2007).

2.6 Summary

An overview of the various modelling approaches investigated for muscle and musculoskeletal components has been presented. The survey of literature in this chapter is intended to highlight the challenges and problems related to the forward dynamic model of human lower limb for simulating FES applications. In general, a model should be kept as simple as possible, that is, its order and number of parameters should be as low as possible. A too simple a model lacks depth of understanding the system properties and leads to inaccurate representation of system behaviour. Conversely, a too complex a model may lead to an inability to gain sufficient insight into system behaviour due to the tendency to get lost in model details such as parameter identification. In fact, musculoskeletal systems are complex, being inherently higher-order and nonlinear. The traditional way of handling such a system with using a mathematical model has ended up with large equations. One way of overcoming this problem is by identifying new modelling approaches of nonlinear active muscle properties and nonlinear passive joint properties.

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