REGULATION OF FES PARAMETERS FOR REDUCED MUSCLE FATIGUE IN LONG TERM SPINAL CORD INJURED INDIVIDUALS

by

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Abstract

Spinal cord injury (SCI) is a medical condition that occurs as a result of trauma, sickness or brain injury and as a result the connection between the brain and some part of the body is lost. This will result in tremendous changes in the daily life of the individual affected. Secondary complications arising from SCI are of very importance as well since some of them if left unattended, will result in life treating conditions. Functional electrical stimulation (FES) is one of the methods developed by researchers in order to get paralysed limbs functional with the help of electrical stimulation once again but the issue of muscle fatigue is limiting the efficiency of this method.

In this thesis the primary and secondary matters arising from SCI are discussed and a CAD-based VN4D humanoid model integrated with Matlab/ Simulink, for use as a platform for analysis, design and evaluation of the developed strategies and approaches in this research is developed. Muscle models using different methodologies are developed in order to represent the reaction of the paralysed muscle to electrical stimulation and their output torque once moved. It was decided that the best model to represent the muscle is the proposed ANFIS model which was then integrated into the VN4D model. One of the main limitations of FES which is rapid muscle fatigue is studied in depth and the stimulation parameters which are having the greatest impact on the muscle fatigue are identified. It was determined that frequency modulation results in faster muscle fatigue therefore the controlled parameter is set to be pulse width after an experiment on 15 SCI individuals.

Different control methodologies, including PID, fuzzy, adaptive neuro-fuzzy and iterative learning control (ILC) to move the simulated paralyzed model previously build using VN4D are explained and has been suggested that fuzzy and PID resulted in fatigue happening later once compared with adaptive and iterative learning control. This is made possible by the study of the trend of changes in pulse width and the amount of energy induced to the muscles. While all four control methods are showing great results once compared to a reference trajectory, adaptive and ILC controllers

have induced greater amount of energy to the muscles resulting in faster fatigue. Two practical FES exercise activities including FES leg extension exercise and FES rowing are designed and controlled in a feedback control setting. It is realised that by employing mechanical facilitators in an FES activity a smoother performance is derived and less fatigue is induced in the muscles resulting in the individual being able to exercise for a longer period of time without damaging any tissue or bone.

In summary, this work resulted in development of a new generalised accurate muscle model which was then used to display that how the issue of premature fatigue is affecting the performance of an FES exercise machine and how the control parameters and the choice of control strategy will result in different amount of energy induced in the muscles which in return results in fatigue showing in the earlier cycles of the movement. Further experiments are carried out on a group of spinal cord injured individuals and two FES exercise facilities, FES leg extension exercise machine and FES rowing machine, are further developed.

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Love to my grand mother, Malakeh, my grand father, Abbas, my aunts, uncle and cousins all around the globe!

DEDICATION

This thesis is dedicated to

My Parents,

Professor M.T.Joghtaei

&

Shiva Bagheriyan

And

in loving memory of my passed away grandparents

Akbar

&

Fatima

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Chapter 1

Introduction

1.1 Overview

Human life style is extensively dependent on the individuals' ability to move around voluntarily. To be able to manoeuvre willingly the body parts need to be in constant contact with the brain to receive commands or to send feedbacks. When due to an unfortunate event such as an accident, sickness or brain injury the connection between the brain and a part of the body is lost that part of the body will become paralyzed. Apart from the paralysis itself there are secondary complications arising from this issue that can be life-threatening in most cases and need extensive attention. Spinal cord injury (SCI) is one of the major unfortunate incidents in which the connection between the brain and body segments is lost due to an injury to the spinal cord.

Unfortunately apart from a few Scandinavian countries, there is not much of an up to date registry on occurrence of SCI so the number of spinal cord individuals globally is not known. In recent years in the UK charities such as Aspire together with national health services (NHS) and Salisbury district hospital started to compile some statistics on the number of SCIs in the UK. The same is happening in most of the developed countries all around the world but there is still quite a way to for an up to date global registry to be available. The benefit of having such an up to date registry is to plan to service the affected individuals resulting in better financial planning and better overall care and treatment. In 2007 it was estimated that there are between 40,000 and 50,000 individuals living with SCI in the UK and the number is increasing by more than 1000 new case each year (Hutchins, 2007). This number has been revised by Aspire (2011) to 1200 new cases per year recently. The same pattern can be seen in USA with the number of new SCI cases rising from 10,000 in 1995 to 12,000 per year in 2009 (Stover, 1995 and NSCISC, 2009).

This rise in the number of affected individuals due to let alone SCI demands an up to date program not to only address the survival and traditional rehabilitation issues

but to strengthen the muscles, provide mobility, effective exercises and recreational activities, and last but not least, maximizing adaptation of the individual to the environment. In the last 15 to 20 years many efforts have been dedicated to make the spinal cord injured individual mobile or to provide them with alternative exercise regimes by actively recruiting their muscles unlike old fashioned passive treatments and extensive progresses have been made so far. However, the newly recruited methods are mostly adaptive to those who have been recently injured and did not lose the majority of their motor unit activity as well as the capacity of their muscles. These new methods such as functional electrical stimulation (FES) are showing approving results on recently injured subjects while the same cannot be said for those subjects with unattended long term injuries. Therefore, providing the individual with a long term history of unattended injury with recreational strengthening exercise activity should be one of the main priorities in the field of rehabilitation. Moreover, the subjects with recent history of the injury who need to gain some confidence to themselves and build up their weakened muscles before going towards more complicated FES treatments, can benefit extensively from such designed muscle strengthening programs.

1.2 Spinal cord injury

Spinal cord injury is the damage to the spinal cord which normally results in the loss of voluntary control of the affected individual's limb below the level of injury. SCI often occurs unexpectedly and most frequently because of trauma such as a motor vehicle accident, a fall, diving into lakes, etc. The earliest description of spinal cord injury was found in the Edwin Smith Surgical Papyrus, written in about 2500 B C (Inman, 1999). Since then addressing the issue of SCI have found itself many dedicated scientist and researchers however due to the lack of advancements in the field of rehabilitation technologies it was only till 30 years ago that researchers where able to deal with complication arising from SCI in a modern way. The complication after injury does not limit to the lack of movement but secondary complication have proved to more life threatening. Before World War II many of the spinal cord injured individuals ended up dead soon after the injury not because of the injury itself but due to secondary complication.

SCI is classified into “complete”, which means a total loss of function below the neurological level and “incomplete” injury in which a patient may have some motor or sensory function below the affected area. The most common types of SCI are tetraplegia (quadriplegia) and paraplegia. Paraplegia is impairment in motor and/or sensory function of the lower extremities and is more common than tetraplegia.

SCI primarily causes loss of sensation, loss of movement (physical inactivity) and loss of muscle function (weakness and paralysis). Some of the secondary causes of SCI can be summarised as breathing issues, irregular heartbeat, spasm, pressure sores and loss of bladder control just to mention a few. Of course some of these complications does relate to the level of injury. Studies (Mizrahi, 1997, Janssen et al., 2002, Guttmann, 1976) have shown that the sooner affected individual attends an up to date rehabilitation program the faster some of the secondary complications would be eliminated or at least be limited.

SCI changes the way an affected individual lives dramatically in seconds however it will take an extended amount of time for SCI subject to regain some of their lost abilities therefore an effective exercise regime and efficient exercise facility is in demand. The exercise facility should be cosmetically appealing to the subject, user friendly, cost effective and last but not least the subjects should be able to feel and visualise their progress.

1.3 Functional electrical stimulation

Functional electrical stimulation (FES) has been widely used as an effective treatment strategy for muscular training in persons with SCI by activating nerves innervating affected extremities (Hooker et al. 1992). It is a substitute for absent bioelectric activity with appropriate electric pulses generated by a stimulator, or the elimination of the hyperactivity in paralysis and spastic paresis (Pasniczek and Kiwerski, 2004). In a FES facility the pattern of electrical pulses are developed in the controller section. These pulses are then generated in the stimulator and delivered to the target muscles using electrodes. The first application of the technology was intended to permanently replace lost neuromuscular function. FES has been tried on the disabled population by evoking

involuntary muscle contraction in which one of the first application where standing and walking (Thrasher and Popovic, 2008).

FES is not a cure and does not repair or regenerate the damaged nerves but can be used as a limited functional recovery tool. FES has been used in individuals with SCI to induce function and improve the mobility, prevent the atrophy, and increase the quality of life in these individuals (Sisto et al., 2008). The applications of the FES can be seen in lower and upper extremities which include standing, walking, cycling, rowing, ambulation, bowel movement control and grasping (ref needed). The benefits of the FES are enormous yet due to the complications involved in adapting it universally there is still a long road ahead. One of the limitations of electrical stimulation for restoration of functional movements, which is mainly a muscle physiology matter, is the issue of the rapid fatigue. It is a known fact that muscle fatigues far more rapidly when artificially stimulated than when excited by the central nervous system. As a result, successful implementation of FES paradigms for rehabilitation has been greatly limited by premature fatigue (Karu et al., 1995).

1.4 FES assisted exercise

Apart from the paralysis of the limbs below the level of injury, spinal cord injury will result in other degenerative pathological conditions such as reduced cardio respiratory fitness, poor blood circulation, loss of bone mineral density and muscle atrophy (Bauman et al., 1999, Belanger et al., 2000). Scelsi (2001) states that the studies since 1991 have well documented the skeletal muscle changes in paraplegics right after SCI, pointing out some of the primary and secondary conflicts a spinal cord injured individual might face and with regard to the issue of muscle atrophy argues that the main mechanism responsible for the skeletal muscle atrophy in paraplegics is thought to be disuse. It should be bared in mind that due to the injury of the spinal cord and lack of the movement the muscle fibres begin to change their normal functional properties straight away after the injury. Also the interest is dedicated to it is necessary to evaluate the conditions of the motor unit and the muscle fibre reversibility and potentialities after the trauma. Physical training, FES, aerobic exercise trainers and bio-mechanic orthoses are some of the rehabilitative options available to paraplegics in order to prevent some

clinical complications and to modify muscle atrophy. FES provokes changes of biomechanical, morphological and contractile properties of muscle fibres and improves the muscle capillarity (Rochester and Barron, 1995).

When the lower motor neuron system is intact, paraplegics and quadriplegics can perform substantial exercise through the use of computerized functional electrical stimulation. Many of the effects of physical inactivity experienced after SCI may be reversed by FES intervention (Kristjan et al., 1991). Functional electrical stimulation is proven to be able to tackle, primary and secondary complication arising from spinal cord injury throughout FES standing, walking, cycling, rowing, endurance training and knee extension (Davis, 1991, Davoodi et al., 2001, Andrews et al., 2002, Eom, 2006, Sisto et al., 2008). Other researchers including Arnold et al, (1992), Phillips et al., (1989) and Petrofsky et al., (1984) report an increase in thigh girth following FES training.

FES leg extension is a method in which it can provide the subject with endurance training as well as to help blood circulation in the lower extremities by working out the quadriceps muscles and help to tackle the complication arising from SCI. Janssen et al. (1998), argue that FES-knee extension training assists limiting the abnormalities of the knee joint is SCI while after 4 weeks of knee extension exersice Ragnarsson et al. (1988) observe an increase in resistance from 25 N to 37 N in the subjects they were testing on. Normally before starting any kind of FES exercise muscle strengthening is suggested. A FES-knee extension training program can be an alternative solution compare to the traditional rehabilitation program.

Sometimes the engineers lose sight of what a rehabilitation perspective for the spinal cord injured individual is and get lost in the complications of the complex engineering or physical matter. Of those researches dedicated to addressing the issue of primary and secondary complications after the spinal cord injury many have been devoted to those individuals injured very recently and does not address the complications those individuals with longer post injury time are facing in order to make them ready to be able to use the newest state of the art rehabilitation facilities and programs. This issue can be sensed developing countries in which the paraplegic did not have the chance to go undergo immediate treatment right after the injury and now suffers from extensive complications resulting from their injuries.

Moreover, in the preparation stage of the subject to go on a rehabilitation program such as an FES based activity the efforts have been focused on getting the

subject to certain level in a specific amount of time not to prepare them to overcome the issue of muscle fatigue through choosing the suitable rehabilitation routine and the use of right stimulation pattern and technique. In order for the spinal cord injured individual to benefit the most from an FES activity, the subject must be able exercise up to the point that the fatigue is believed to be of the same level of central fatigue. At this point the limbs have received the appropriate amount of oxygen and the fatigue induced will result in rebuilding and shaping the trained muscles. Therefore, making sure that premature fatigue is not resulting in shortening the exercise time is of very importance. A suitable and easy to implement control approach together with extra mechanical support will result in reducing the primary and the secondary complications arising from SCI.

1.5 Research aim and objective

The primarily aim of this research is to assess different levels of fatigue induced in the muscles due to the choice of controlled stimulation parameters as well as the type of control methodology. In order to achieve this successfully the following research objectives have been set out:

1. To developed a generalized humanoid model using VN4D software in order to present the subject as accurate as possible for the purpose of this research.

2. To optimize the passive properties of the subjects' lower extremities obtained from performing practical relevant tests and incorporating them into the final design of the humanoid model.

3. To investigate the issues surrounding muscle fatigue and address the effect that stimulation parameters have on stimulated muscles.

4. To develop a generalized accurate muscle model based on the experimental data to represent the muscle physiology of the tested subjects.

5. To develop different known strategies to control the humanoid model in order to compare their results with regards to the number of cycles they force the

humanoid model to complete. The amount of energy these controllers each induce in the system is also examined.

6. To examine the effect of using mechanical support in 2 practical FES applications, leg extension and rowing, on the number of cycles an individual can perform and also to study the level of energy induced in the stimulated muscles during these 2 exercises.

1.6 Thesis outline

Based on the proposed aim and subsequent objectives of the research, this thesis is divided into eight chapters.

Chapter 1: This chapter briefly introduces spinal cord injury and the complications arising from it, functional electrical stimulation and the exercises developed using this concept, and also provides and insight into the aims and objective of the presented research.

Chapter 2: This chapter summarises the literature relevant to this thesis with regards to the anatomy of nervous system, definition of SCI, its causes, how it is classified, the problems associated with it, FES activities available to spinal cord injured individuals and their advantages and disadvantages. The gap in the literature is highlighted and a guideline to the element of this thesis with regards to the novelty is given.

Chapter 3: In this chapter humanoid modelling using VN4D software and the integration of this software with Matlab/ Simulink is explored. In order to create a precise model, passive and active parameters of the knee joint are identified and integrated into the physical model.

Chapter 4: Describes the microstructure of muscles in general and the typical way of muscle contracts. Muscle modelling and stablished muscle models such as Hill's, Huxley's, Zajac's, Ferrarin's and Reiner's are described. The option of having a knee joint model which is developed using 2 different methods, ANN and NARMAX-OLS-

ERR, instead of a muscle model to represent the musculoskeletal joint system is further examined. Finally, an stable generalized fatigue predictable ANFIS based muscle model to be used for the purpose of modeling in FES activities is further developed.

Chapter 5: In this chapter muscle fatigue, premature fatigue and natural fatigue is defined and the factors affecting the fatigue in electrically activated muscles are discussed. A study is carried out to analyse the effect of stimulation parameters, frequency and pulse width, on the issue of muscle fatigue.

Chapter 6: This chapter describes some of the known control approaches available, PID, fuzzy, adaptive neuro-fuzzy and iterative learning control, to control the simulated system and examines the performances of these controllers by comparing their outputs to a reference trajectory. The number of cycles each of these controllers were able to complete and the amount of energy they each induced in the system are reported and analysed.

Chapter 7: In this chapter the effects of some of the previously built controllers are practically evaluated on SCI subjects with regard to the trend of changes in pulse width and the energy induction to the stimulated muscles. Two FES exercise activities including an FES leg extension and an FES rowing are designed, developed and controlled. The effect of employing mechanical facilitators in a FES exercise activity is studied. The same methodology expressed in chapter 6 is used to analyse the effect of fatigue in stimulated muscles.

Chapter 8: This chapter summarizes the main conclusions of the work and makes recommendations for further work.

1.7 Publication

1. M. Joghtaei, M.O. Tokhi, D. Kahani, P. Mireshghi, M. Samani, A.M. Arabloo, The combined effect of incorporating mechanical apparatuses into an FES exercise systems together with suitable control approach in order to reduce muscle fatigue, Biomedical Engineering Online, 2013, Submitted.

2. M. Joghtaei, M.O.Tokhi, M.Arabloo, R. Massoud, The effect of pulse width modulation in the quality of the controlled human limb in functional electrical stimulation based task, Journal of Disability and Rehabilitation, 2013, Submitted.

3. M. Bijanzadeh, D. Kahani, E. Daneshi Kohan, M. Joghtaei, S. C. Gharooni, M. O. Tokhi, NARMAX_OLS Representation of a Semi-active Dynamic Leg Joint Model for a Paraplegic Subject Using Functional Electrical Stimulation, Proceedings of the International Conference of Systems Biology and Bioengineering, London, UK, 3-5 July 2013.

5. R. Jailani, M.O. Tokhi1, S.C. Gharooni1 and M. Jogtaei, Finite State Control of FES-Assisted Walking with Spring Brake Orthosis, UKSim 13th International Conference on Modelling and Simulation.

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Chapter 2

Literature Review

2.1 Introduction

Spinal cord injury (SCI) is a medical condition in which the brain cannot control the voluntary movements of the body parts below the level of injury on the spinal cord. This will result in significant changes in the individual's lifestyle and results in primary and secondary conflictions. These conflictions and changes in the SCI subject's lifestyle will be financially taxing to the National Health Service (NHS), insurance companies and the individuals themselves. Therefore, providing these individuals with solutions to reduce these effects and side effects are of high priority to organisations and researchers involved. Due to the fact that the neurones themselves are still excitable, functional electrical stimulation (FES) can be employed to stimulate these neurones in order to provide muscles with contractions resulting in extension and flexion of them. By controlling the stimulus signals one can achieve a proper functional movement.

In this chapter, the anatomy of nervous system and how it functions is briefly discussed firs. This is followed by definition of SCI, its causes, how it is classified, the problems associated with it and a brief discussion on its economic implications. FES is introduced as a tool enabling subjects to perform functional tasks and the way it will recruit motor units will be discussed. Advantageous and disadvantageous of FES, the literature available on FES exercise facilities and where they lack in further investigation are also addressed in this chapter.

2.2 Nervous system

The nervous system is the most complex and highly organized of the various systems which make up the human body. It is the mechanism concerned with the correlation and integration of various bodily processes and the reactions and adjustments of the organism to its environment. In addition the cerebral cortex is concerned with conscious life. It may be divided into two parts, central and peripheral. The central nervous system

(CNS) consists of the encephalon or brain, contained within the cranium, and the medulla spinalis or spinal cord, lodged in the vertebral canal; the two portions are continuous with one another at the level of the upper border of the atlas vertebra (Gray, 1918). The brain contains over 12 billion neurones and 50 billion supporting glial cells. With the spinal cord the brain regulates bodily processes and coordinates voluntary movements (Baggaley, 2001). The peripheral nervous system (PNS) consists of a series of nerves by which the central nervous system is connected with the various tissues of the body (Gray, 1918). The PNS is divided into three divisions: autonomic which controls the involuntary actions, sensory which transmit information from the body to CNS and motor nerves that carry signals from the brain to voluntary skeletal muscles (Baggaley, 2001).

2.2.1 Spinal cord

The spinal cord is a cable of approximately 43 cm in length running from the brain stem to the lower back. It contains two types of tissue. The inner core is grey matter made up of nerve cell bodies, unmyelinated axons, glial cells and blood vessels; the outer white matter is mainly composed of tracs of myelinated axons that relay impulses to and from the spinal cord and the brain (Baggaley, 2001). Figure 2.1 shows a transverse section of spinal cord.

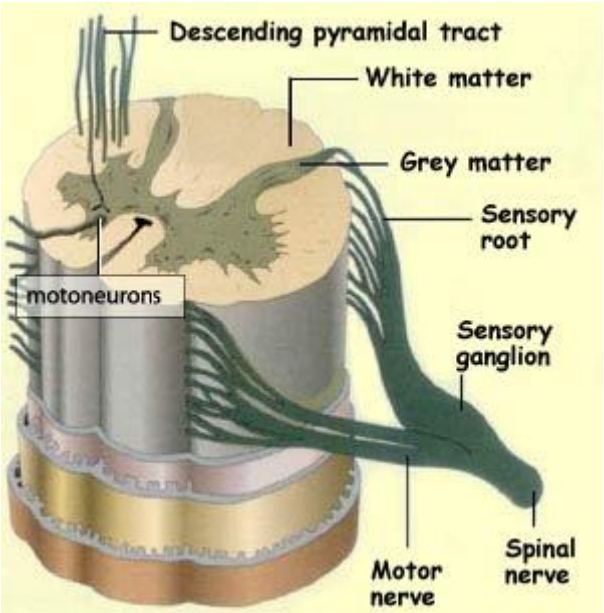


Figure 2.1: Spinal cord (Baggley, 2001)

The vertebral column and its supporting ligaments contain and protect the spinal cord. The vertebral column is a flexuous and flexible column, formed of a series of bones called vertebrae. The vertebrae are thirty-three in number, and are grouped under the names cervical, thoracic, lumbar, sacral, and coccygeal, according to the regions they occupy; there are seven in the cervical region, twelve in the thoracic, five in the lumbar, five in the sacral, and four in the coccygeal (Gray, 1918). In addition to the vertebral column further protection is provided by cerebrospinal fluid which acts as shock absorber and epidural space, a cushioning area of fat and connective tissue that lies in between the periosteum and the dura matter (Baggaley, 2001).

The spinal nerves spring from the spinal cord, and are transmitted through the intervertebral foramina. They number into thirty-one pairs, which are grouped as follows: Cervical, 8; Thoracic, 12; Lumbar, 5; Sacral, 5; Coccygeal, 1.The first cervical nerve emerges from the vertebral canal between the occipital bone and the atlas, and is therefore called the suboccipital nerve; the eighth issues between the seventh cervical and first thoracic vertebrae (Gray, 2005). By means of spinal nerves the brain can communicates with the rest of the body. The spinal nerves emerge from the gap between adjacent vertebrae. Each nerve divides and subdivides into a number of branches; two main divisions serve the front and the back of the body in the region they supply. Figure 2.2 illustrates how the spinal cord is protected and also how the spinal nerves are branched from the spinal cord while Figure 2.3 portrays the overall nervous system of the human body.

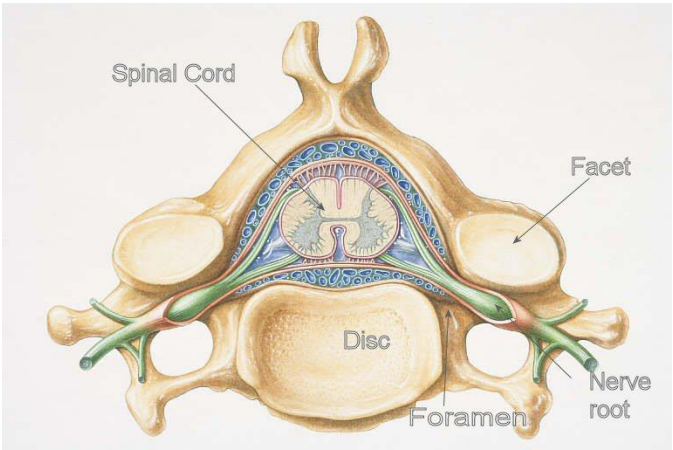


Figure2.2: Protection of spinal cord (Baggley, 2001)

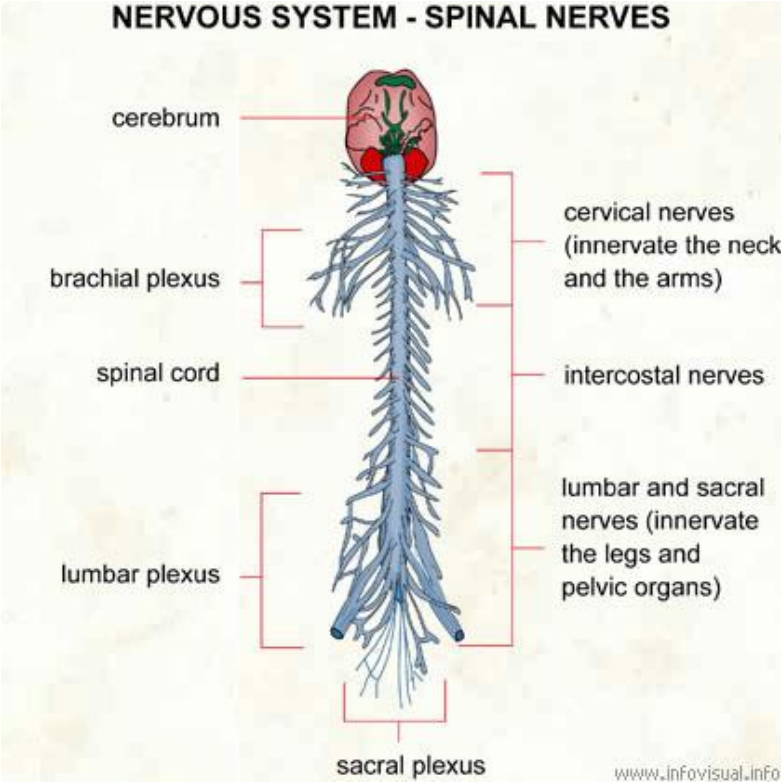


Figure 2.3: Overall nervous system (Baggley, 2001)

2.2.2 The neurone

The basic unit of the nervous system is a special cell called a neurone or a nerve cell (Baggaley, 2001), which consists of the cell body, axons and dendrites (Sherwood, 2011).The cell body does have a central nucleus which does not divide and multiply as most of the other cells do and any damage to the cell body may result in degeneration of the entire neurone (Baggaley, 2001). Axons, also known as nerve body, are the longest process extending from the cell body and carrying nerve impulses away from the cell whereas dendrites receive impulses from the other neurones (Gray, 2005).

According to their function and location in the body the shape and size of neuron cell bodies vary as do the type, number, and length of their processes. Figure 2-4 shows the three main types of neuron: unipolar, bipolar, and multipolar neurones. Multipolar neurones are the most common neurones found in the brain and spinal cord (Baggaley, 2001).

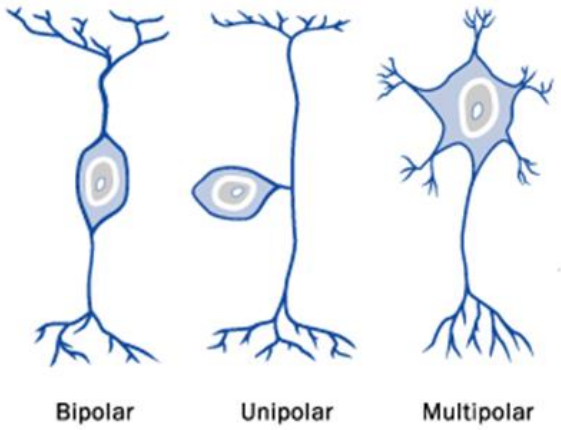


Figure 2.4: Different Neuron types (Baggley, 2001)

Neurones are considered to be excitable. They must be triggered by a stimulus to produce nerve impulses. When the neuron receives a stimulus, the electrical charge on the inside of the cell membrane changes from negative to positive. A nerve impulse travels down the fibre to a synaptic knob at its end, triggering the release of chemicals (neurotransmitters) that cross the gap between the neuron and the target cell, stimulating a response in the target. The communication point between neurones is called synapse which comprises the synaptic knob, the synaptic cleft and the target site (Baggaley, 2001). Action potential is the transmission of nerve pulses throughout a neuron which ends in the axon and finally to the synaptic knob. Four general stages are recognised for the action potential starting from the resting situation when the threshold is passed. Threshold is the level at which a stimulus is strong enough to transmit a nerve impulse. The process of sending a nerve impulse is illustrated in Figure 2-5.

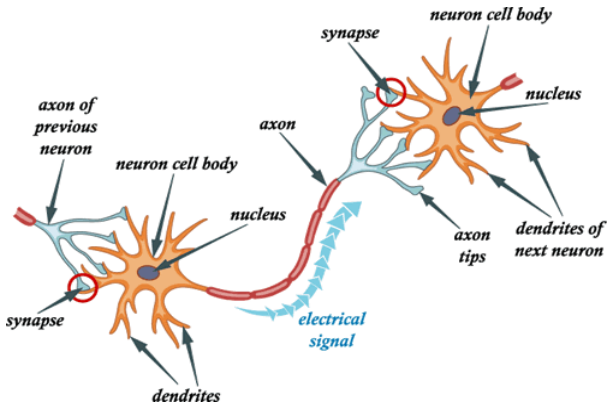


Figure 2.5: Nerve impulse (Baggley, 2001)

2.2.3 Skeletal muscles

In general there are three types of muscle, skeletal, smooth and cardiac muscles. Skeletal muscles consist of densely packed groups of elongated cells, known as muscle fibres, held together by fibrous connective tissue. Numerous capillaries penetrate this tissue to keep muscles supplied with abundant quantities of oxygen and glucose needed to fuel muscle contraction. (Baggaley, 2001). Figure 2-6 provide a schematic of these different muscle types.

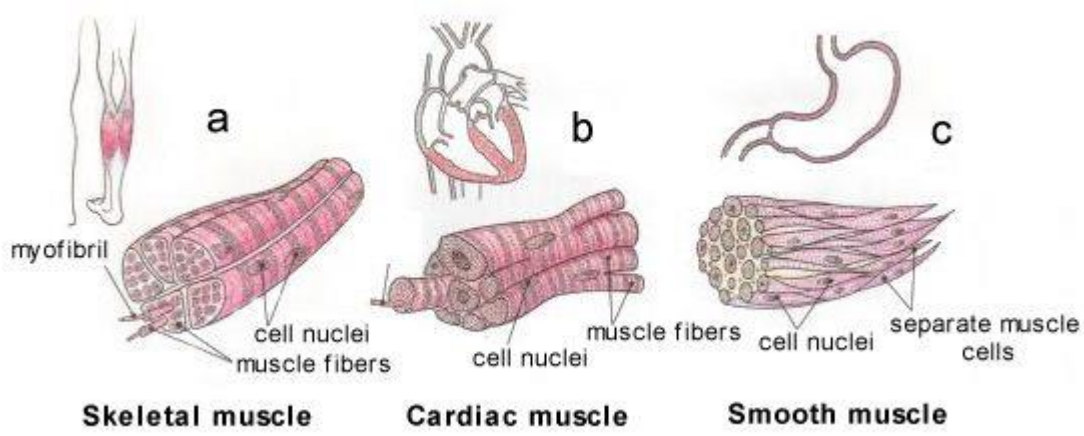


Figure 2.6: Different types of muscle (Gray, 2005)

(a) The structure of skeletal muscles, (b) The structure of cardiac muscles, (c) The structure of smooth muscles.

Muscle contracts as a result of a complex process involving a number of cellular proteins and energy production systems. The action potential causes the nerve cell to fire, inducing a reaction that releases a neurotransmitter (acetylcholine) at the level of the synapse. Thus, the process of muscular contraction begins when a nerve impulse arrives at the neuromuscular junction. The micro skeletal structure of the muscles and detailed explanation of how the skeletal muscles contract is provided in chapter 3 section 2.

2.3 Spinal cord injury

Spinal cord injury (SCI) occurs as a result of an injury to the spinal cord, a part of the central nervous system, or as a result of trauma, infection, or disease. It is a condition that extremely changes the patient's lifestyle, physically and mentally. Individuals with SCI have limited mobility because of the loss of voluntary muscle control and limitations in sensory, autonomic, reflexes, and visceral organ functions. Secondary complications arising from these limitations include muscle atrophy, osteoporosis, abnormal thrombus formation, pressure ulcer formation, urinary tract infection, heart disease, and cardiopulmonary de-conditioning (Sisto et al., 2008).

The experience of SCI is one of the most devastating injuries which might affect an individual. It usually necessitates considerable changes in the life of an individual, and their family members (North, 1999). It has been argued that traumatic spinal cord lesion causes not only physical injury but also an emotional state of crisis. The adjustment process is characterized by various phases, such as denial, depression and aggression (Orbaan, 1986).

Prior to the early 1940s, 80% to 90% of people with SCI died within weeks (Carroll, 1970). Some with chronic ill-health did manage to live for 2–3 years before they eventually succumbed to sepsis, mainly from the urinary tract and pressure sores (Guttmann, 1976). Improved medical and surgical treatment and better rehabilitation care have changed the quality of life of SCI patients and prolonged survival (Schurch et al., 1996). Due to these advancements in medicine and technology the number of individuals with SCI passing away is now more to do with unrelated causes such as cancer or cardiovascular disease, similar to that of the general population (NSCIA, 2011). Currently the main cause of death in individuals suffering from SCI is cardiovascular disease (Wilder 2002).

2.3.1 Causes

Spinal cord injuries can be traumatic or due to nontraumatic disorders. Falls, traffic accidents, war injuries, sport accidents and assaults are among the main causes of traumatic spinal cord injuries while lateral or multiple sclerosis and tumours are among the main causes for nontraumatic spinal cord injuries (Mohr, 1997).

In 2002 it was reported that the incidents of SCI in the United Kingdom were 10 to 15 per million people per annum, i.e. 600 to 900 new cases per year (Swain, 2002). The incidence of traumatic SCI in the USA was estimated to be over 10000 cases per year, with an estimated prevalence exceeding 200000. The number is more than double when non-traumatic causes are included Stover as well (Stover, 1995). The latest figures by (Aspire, 2011), show that in the UK, there are around 1200 people paralysed from a SCI every year. This statistic only shows the people who have been through an SCI centre, and does not include those who have suffered paralysis and been treated in a general hospital. In the UK, one person is paralysed every 8 hours. There have not been any studies of the overall incidences of SCI in the United States during the past two decades. However, state based SCI registries in the 1980's and 1990's provided both an estimate of SCI incidence in the United States as well as the identification of regional differences in SCI incidence (NSCISC, 2011).

In the United Kingdom falls are the main cause resulting in SCI forming 41.7% of the incidents, closely followed by road traffic accidents at 36.8%. 11.6% of injuries are due to sport accidents, 4.2% resulted from knocked over, collision and lifting, 3.3% from traumas and 2.7% due to sharp trauma or assaults. 12.6% of the falls are caused by falling from a height, 11. 7% from falling down the stairs, 7.5% from fall downs and 3.3% from jumping. 3.5% of sport accidents are due to diving, 2.6% from horse riding and 2.4% due to rugby (Apparalyzed, 2011).

In the United States motor vehicle crashes rank first at 41.3% followed by falls at 27.3%, acts of violence at 15.0%, sports at 7.9% and all others at 8.5%. These figures are for all injuries reported to the National Database since 2005. The percentage of cases due to acts of violence increased through the early 1990's but has since declined. The percentage of cases due to falls has increased steadily since 1973. Cases due to sports injuries have been decreasing (NSCISC, 2011).

In Australia it is estimated that over 10,000 people are living today with spinal cord injuries. Each year in Australia there are about 300-400 new cases of SCI. Transport related accidents represent 52% of all spinal cord injuries while falls represent 28% of the SCI cases. Sport related incidents form 16% of the figure, out of which 9% is due to water related (mainly diving and surfing) sports (Cripps, 2006).

It is interesting in these figures that the difference between the percentage of the causes in different regions which could be linked to cultural and local habits. This can

be further investigated for the benefit of the health care societies in order to provide better health care solutions to the spinal cord injured individuals. Figure 2.7 graphically illustrates the SCI causes in the three discussed regions.

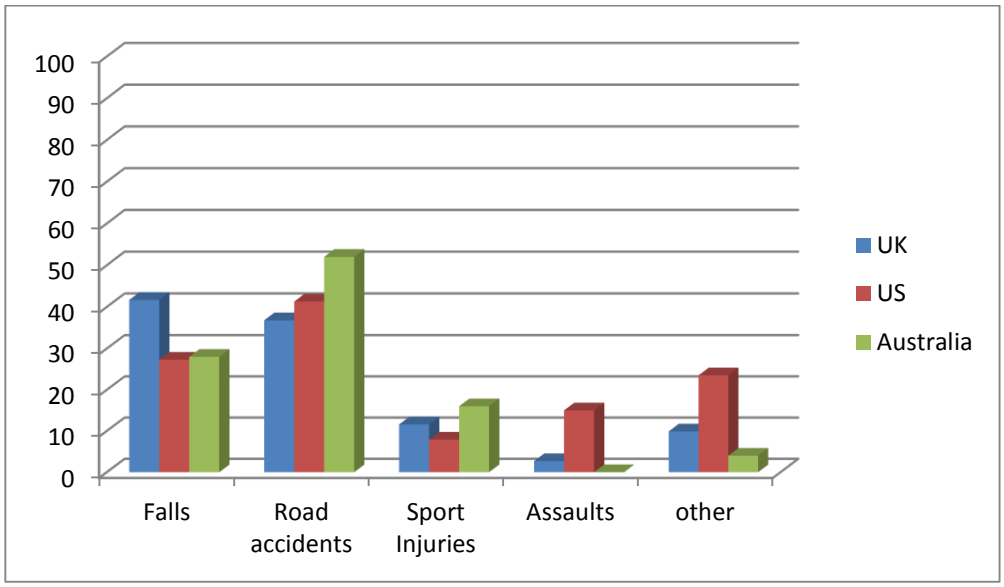


Figure 2.7: Causes of SCI in the UK, US and Australia. (Apparalyzed, 2011; Cripps, 2006; NSCISC, 2011)

2.3.2 Classification

When the spinal cord is hit the communication between the brain and the parts of the body below the level of injury will be disconnected. The extent of the damage depends on the location (level) of the injury. The higher the location of injury on the spinal cord the greater the damage.

When the injury occurs in the cervical (neck) region between cervical levels 1 and 7, it is referred to as Tetraplegia, previously Quadriplegia. This means that all four limbs are affected. These injuries, when complete, result in full paralysis of the legs and partial paralysis of the trunk and arms. There is also a loss of sensation, which can lead to pressure ulcers if there is improper seating or weight relief. If the injury is in the thoracic (trunk) region or the lumbar (low back) region, resulting paraplegia ensues. In this case,

the arms are fully functional, and if complete, there is partial paralysis of the trunk and the legs are completely paralyzed. In many cases, SCI results in an incomplete injury that may include a person having only some feeling (sensation) below the injury level or some ability to contract the muscles and move the limbs below the injury level. When there is an incomplete injury, the term paresis can be used, such as Paraparesis (Sisto, 2008). Figure 2-8 shows the relation between the level of the injury and the extend of the paralysis.

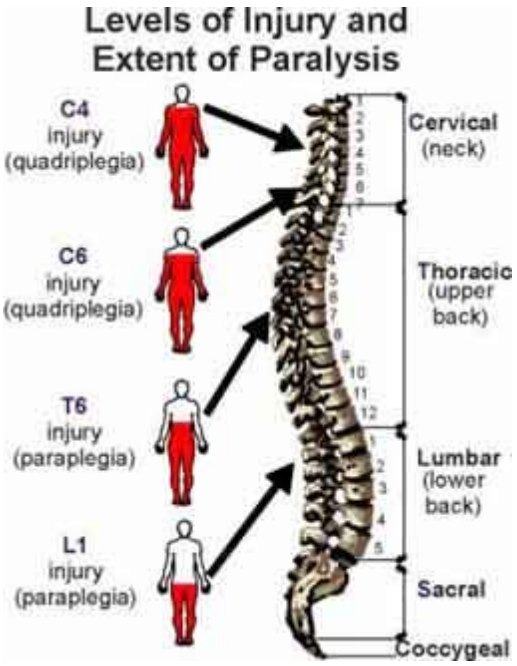


Figure 2.8: The relation between level of the lesion and the extend of the paralysis (NSCISC, 2011)

Using the American Spinal Injury Association (ASIA) impairment scale the severity of injury can be classified. The extend of the injury is defined using five main gardes(A-E) where grade A defines the injury as being complete while B, C and D represent different levels of incomplete injury and E indicates that normal motor and sensory function is available. Table 2.1 describes these grades with distinctive detail.

Table 2.1: The relation between level of the lesion and the extend of the paralysis (McDonald, 2002)

ASIA IMPAIRMENT SCALE

- ☐ **A = Complete:** No motor or sensory function is preserved in the sacral segments S4-S5.
- ☐ **B = Incomplete:** Sensory but not motor function is preserved below the neurological level and includes the sacral segments S4-S5.
- ☐ **C = Incomplete:** Motor function is preserved below the neurological level, and more than half of key muscles below the neurological level have a muscle grade less than 3.
- ☐ **D = Incomplete:** Motor function is preserved below the neurological level, and at least half of key muscles below the neurological level have a muscle grade of 3 or more.
- ☐ **E = Normal:** motor and sensory function are normal

CLINICAL SYNDROMES

- ☐ Central Cord
- ☐ Brown-Sequard
- ☐ Anterior Cord
- ☐ Conus Medullaris
- ☐ Cauda Equina

Since only the channel between the brain and the peripheral nervous system is damaged, the nerves and muscles below the lesion are still functional. It is therefore possible to restore motor functions of paralyzed people by artificially activating their nerves below the lesion.

2.3.3 Associated problems

A spinal cord injury usually begins with a sudden, traumatic blow to the spine that fractures or dislocates vertebrae. It all happens within seconds and the consequences will remain with the individual for the rest of his/her life. It does not only result in movement restrictions and lack of sensation but also in several long term associated problems.

According to National Institute of Health (2011) people who survive an SCI will most likely have medical complications such as chronic pain and bladder and bowel dysfunction, along with an increased susceptibility to respiratory and heart problems. Successful recovery depends upon how well these chronic conditions are handled day to day. These problems can be listed as breathing if the injury happens above C3, C4 and C5 segments, respiratory complications, primarily as a result of pneumonia, which are the leading cause of death in people with SCI, irregular heart beat and low blood pressure as a result of an injury in the cervical level, blood clots in which people with SCI are at triple risk of, spasm which are basically exaggerated reflexes over time, Autonomic dysreflexia which is a life threatening reflex action happening without the brain permission and affects vascular and organ systems controlled by the sympathetic nervous system, pressure sores, pain, bladder and bowel problems, and last but not the least Reproductive and sexual function.

2.3.4 Economic burdens

In the UK, it is estimated that the current annual cost of caring for people paralysed due to SCI is more than £500 million. 21% of people discharged from Spinal Cord Injury Centres go into nursing homes, hospitals or other institutionalised settings rather than their own homes. Around 20% of patients leave Spinal Cord Injury Centres clinically depressed (Apparalyzed, 2011).

In the United States the estimated life time cost of an individual with SCI is between $528,726 and $3,273,270 depending of the level of injury and the age when the individual injured (NSCISC, 2010). Based on NSCISC figures, Proneuron Biotechnologies (2011), estimates that spinal cord injuries cost the U.S. over $14.5 billion per year. This figure does not include lost productivity, which accounts for an additional $5.5 billion.

In Australia annual financial costs of SCI is $1.2 billion while the burden of disease costs $803.2 million. The lifetime costs per incident case of traumatic brain injury (TBI) were estimated to be $2.5 million (Access Economics, 2009).

Considering the low average age of individuals struck by SCI and the associated costs to this, investment in new technologies to improve the living condition of these individuals are not only morally right but also will save the governments substantial sums of money in the long term (NINDS, 2011).

2.4 Functional electrical stimulation

Functional electrical stimulation (FES) is the application of electrical current to excitable tissue in order to supplement or replace function that is lost in neurologically impaired individuals (Peckham, 2005). It is a means of producing contraction in muscles, paralysed due to central nervous system lesion by utilising electrical stimulation. Due to the nature of FES denervated muscles cannot be stimulated with commercially available stimulators (Jaeger, 1996). Physiologically explaining, individuals with an upper motoneurone lesion can only use FES successfully in day to day action (Glaser, 1991).

The functional reanimation of paralyzed limbs has been a longstanding goal of neural prosthetic research, but clinically successful applications have been elusive. The electrical stimulation of neuromuscular systems has been employed clinically since the discovery of electric fish (Loeb, 2005).

In healthy individuals, the central nervous system (CNS) generates the command signal that contracts the muscle. As described in section 2.2.2 this signal finds its way to the muscle transferring from spinal cord to the spinal nerves, subsequently to the neurones and reaching the target muscle by passing through the synapsis transferred to the muscle, where it induces the contraction. When a lesion happens, the natural muscle activation process is interrupted and the CNS command cannot reach the muscles resulting in paralysis. In the absence of the CNS activation command a control system with a stimulator sends a stimulation signal to the muscles via electrodes. When the activation signal reaches the muscle it will act the same way as the neurologically intact body by generating activation potential in the peripheral nerve. Figure x10 shows the schematic of an FES system.

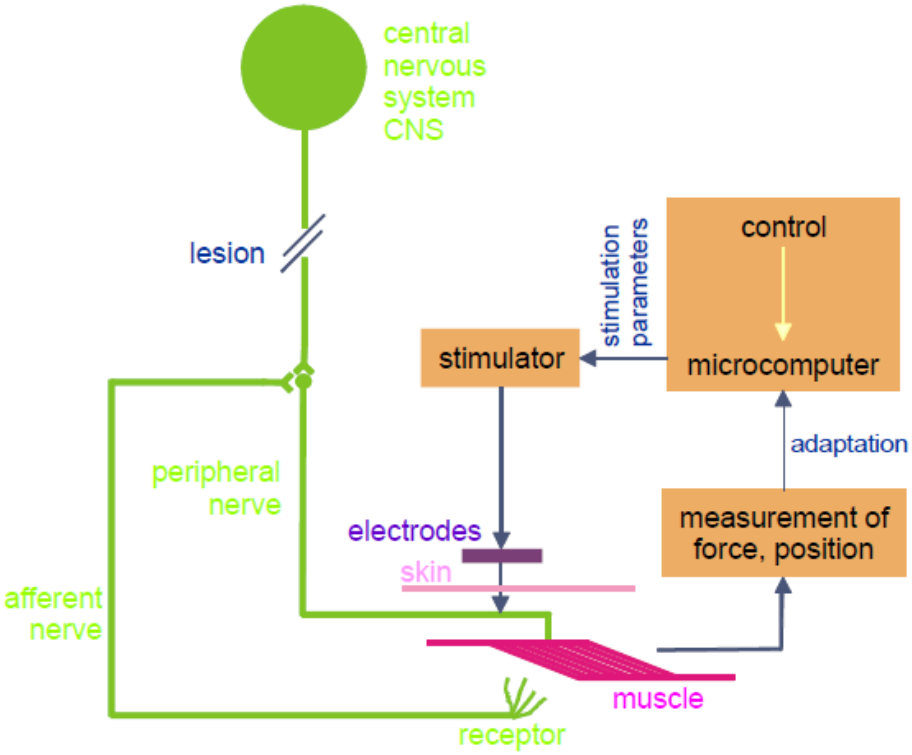


Figure 2.9: Control of movement by means of functional electrical stimulation (FES) (Gföhler et al., 2002)

Two electrodes are essential to close the current circuit. If one of the electrodes is close to the stimulated tissue and the other is distant, this is referred to as mono polar. If both are close it is referred to as bipolar. With more electrode leads, mono polar channels can share the same remote electrode. For FES, and particularly when using surface electrodes; the fewer leads the better (Rushton 1997).

As described earlier the stimulation signal is delivered to the muscles using electrodes. Three types of electrodes are typically used for FES, and each of them presents advantages and disadvantages.

a) **Surface electrodes:** Surface electrodes are non-invasive and easy to apply, but because of high electrical resistance of the human tissue, the stimulus should be relatively high and also the electrodes should be relatively large, to generate sufficient stimulation current (Rushton 1997). The other disadvantage is poor selectivity in discrete stimulation of muscles. They can also sometimes provoke allergic reactions on the skin. It is difficult to position the electrodes in exactly

the same way day to day. Surface electrodes stimulate sensory nerves too, thus it is possible that if the subject has intact or partial sensation the stimulation becomes uncomfortable. Moreover, stimulation of the sensory nerve can cause spasticity or reflex responses (Ferrario, 2006).

b) **Percutaneous electrodes:** This method uses a tiny wire electrode placed close to a motor nerve percutaneously, using a hollow needle as a guide and adjusting electrode location by trial stimulations before withdrawing the needle (Handa et al., 1989). The selectivity is better with respect to surface electrodes and they can easily be removed and replaced in case of breakage, the current needed to stimulate the muscles is lower with respect to surface electrodes and they can reach deeper muscles. The main disadvantages of these electrodes are the need for their replacement periodically due to their failure rate of 4% per month after 6 months and also the insertion site has to be maintained clean as they have problems of infection (Ferrario, 2006).

c) **Implanted electrodes:** Implanted electrodes are surgically installed providing perfect access to the specific stimulation site. These electrodes are suited for long term application. The installation allows direct identification of the anatomy and wider range of possible electrode designs. However it is invasive, scarring makes repeat surgery more difficult; and the number of surgical incision increases in line with the number of separate stimulation channels. In the recent years a new kind of implanted electrode has been developed that does not require major surgery to be positioned (BION microstimulators, Alfred Mann Foundation, Valencia, CA, USA). (Ferrario, 2006; Rushton, 1997).

These three types of electrodes are shown extensively in Figure 2.10.

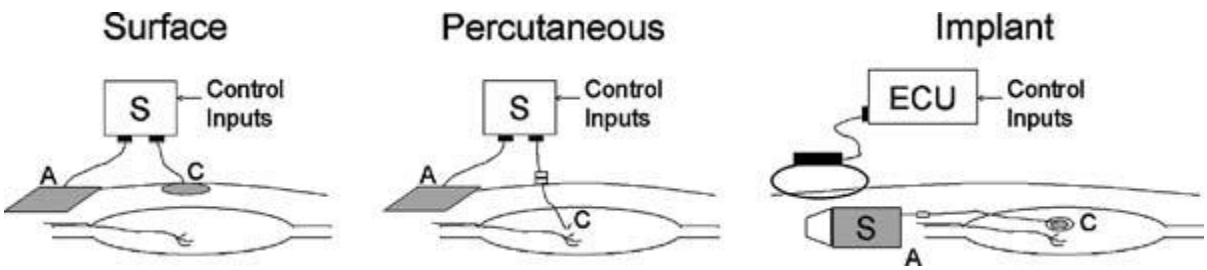


Figure 2.10: Neuroprosthetic system configurations. S = stimulator, A = anode (reference electrode), C = cathode (active electrode), ECU = external control unit. Single-channel monopolar stimulation of one muscle near its motor point is shown fora surface, percutaneous, and implanted system.

Currently the most common way to deliver the signal is by using surface electrodes. Electric current flows from one electrode (the anode) to the second electrode (the cathode) mounted on the skin and stimulate the nerve bundle lying underneath. The capacity of the nerve to be stimulated depends on the properties of the nerve itself (membrane permeability, diameter of the axon, relative position of electrode and nerve) and on the parameters of the electrical impulse (Jaeger, 1996). The electrical current will lead to physiological changes in the underlying tissue. These changes can be thermal, chemical, or physical. Thermal effects are based on Joule's law and could measure the heat generation in an electrical resistor. Physical and chemical changes are due to the direct excitation of neurons and cause transportation of Na+ and K+ ions across semipermeable cell membranes (Sisto, 2008).

Electrical stimulation may be delivered through either open or closed-loop control systems. In open-loop control system, the patient sends out a command that delivers a certain amount of stimulation, regardless of the actual response of the muscle using visual or auditory feedback. In closed-loop control system, electrical stimulation is being initiated with the user's command and then modified based on some feedback measurement such as force or position. With closed-loop control, the delivery of electrical stimulation is continuously modulated to control the quality being measured by the sensors (Sisto, 2008).

There are several parameters in each FES system which need to be adjusted in order to provide an effective functional move. These parameters include waveform, amplitude, current duration, frequency, ramping, and the duty cycle. The waveform is a stimulus pulse consisting of a phase (positive or negative), a shape (sine, rectangular, etc.), and amplitude. The effectiveness of the FES depends on the type of waveform, which affects both the excitability and the fatigability of the muscle. The amplitude or current intensity is the amount of electrical current that is delivered. Increasing the intensity recruits more motor units. The current duration, also referred to as pulse width or pulse duration, is the amount of time that the electrical current is delivered. The current duration affects the intensity that is required to generate a motor response, thus affecting the comfort of the stimulation. The frequency, or pulse rate, is the rate of individual electrical pulses delivered in trains of pulses at a specified frequency. Depending on the muscle mass, a fused contraction (tetanization) can occur with 15–50 pulses per second (pps). An average pulse rate is about 25–30 pps. Ramping is the rate of rise and fall of the current. Rise time allows for sensory accommodation to the

stimulation and thus affects the comfort of stimulation. Fall time is less significant since the muscle will stop contracting as the stimulus decreases during fall time. Finally, the duty cycle is the cycle of stimulation, characterized by the time the unit is on versus the time the unit is off. This is also referred to as on or off time. This ratio is important because a muscle needs adequate rest to avoid fatigue. Adjustment of these various stimulation parameters will affect the quality of the muscle contraction 2-5-1 Motor unit recruitment (Sisto, 2008).

In the case of physiological activation at first the thin, slow motor units of the muscle are activated, when higher forces are needed bigger, fast motor units are subsequently added. Similarly the stimulation frequency is low at the beginning and rose for higher forces. (slow motor units reach their maximum force output at about 30 Hz, fast motor units at about 100 Hz). The activated motor units are distributed in the muscle. In the case of artificial stimulation the recruitment order is reversed to so called inverse recruitment, what means that big, fast motor units are activated first. Additionally motor units in the region of the muscle where the electrical field is stronger are activated first; therefore some parts of the muscle might be active while other parts are totally inactive (Gföhler, 2002).

2.4.1 Motor unit recruitment

It is believed that during voluntary muscle contractions in healthy individuals the nerve fibres are recruited from the smallest to the largest or in a more scientific way they are recruited is size order (Enoka, 2002; Garland, 2004; Knaflitz, 1990; Zajac, and Faden, 1985).

Great difference exists when it comes to the matter of motor unit recruitment in FES. When applying electrical current to induce function in a muscle, the activation of motor units is done in those closer to the stimulation sources that are larger in diameter (type II) and are easier to get into. Generally the larger motor units have large, fast conducting axons (Sisto et al., 2008). It has also been discussed that due to the reverse recruitment order of muscle fibres, muscle fatigue and reduced force gradation has been spotted in FES exercise (Fang and Mortimer, 1991). To overcome these said issues in electrical stimulation, a higher level of electrical current for stimulation may be required to reach the deeper fibbers (Robinson, 1995). The effectiveness of the electrical stimulation to induce muscle contraction is also dependant on the type of stimulator,

electrode size, and FES parameters (e.g., waveforms, frequency, and duty cycle) (Feiereisen et al., 1997; Knaflitz et al., 1990; Sisto et al., 2008).

Having noted argues supporting recruitment size order it is mentionable that in some other studies different conclusion have been drawn such as weaker motor units are activated first or that the fibres are recruited all at the same time resulting in no control on overcoming the fatigue (Thomas et al., 2002; Gregory and Bickel, 2005). To the authors best of the knowledge and experience with FES exercise using surface electrodes while considering Knaflitz et al. (1990) study concluding the reverse recruitment order exists in FES using surface electrodes while vice versa exists when using implanted electrodes, it seems that the argument for recruitment size order is having more ground.

2.4.2 Physiological benefit

The goal of functional electrical stimulation is to enable the disabled body to perform some functional tasks ranging from exercise training to daily activity, providing the individual with physical and psychological support in order to live a healthier life. The collection of research strongly suggests that lower limb training can offer multiple therapeutic benefits for individuals with SCI (Glaser and Shuster, 1998). FES-induced exercise provides the individual with some major physiological improvements ranging from cardiovascular system enhancement (Hooker et al., 1995; Davis, 1993; Pollack et al., 1989; Mutton, and Scremin, 1997, Janssen et al., 2002; Haisma, 2006) muscle atrophy reduction (Davis, 1990; Glaser et al., 1991; Sloan, et al., 1994; Sheffler and Cahe, 2007) muscle endurance and muscle growth (Stein et al,1992; Edwards and Marsolais,1990; Mohr et al., 1997, Nash et al., 1991) blood circulation enhancement (Glaser et al., 1991; Nash and Montalvo, 1996), reduction in the incidence of pressure sores (Petrofsky, 1992) increase of the bone density (Frotzler, 2008). Moreover, one should not forget the enormous psychological (Sipski, 1989; Grundy and Swain, 2002) advantages the individual can receive by being able to perform daily tasks. The improvement in the individual's self perception is also noted.

2.4.3 Disadvantageous

There are also disadvantages to using FES. These include difficulties with electrode placement, inadequate selectivity, the need for costly technology, and insufficient present knowledge of neurology, neuroanatomy. Inefficient user-machine interface, high

cost, and issues related to muscle fatigue and noncompliance are other problems that need attention (Sisto, 2008).

2.5 Literature review on the thesis aim

In order to address some of the difficulties arising from the FES researchers have extensively explored different techniques to limit these complications and produce a smooth and sustainable activity regime. Though, of all the issues attended by the research community the issue of muscle fatigue and how it limits the functionality of an FES exercise facility, remains a topic to be discussed. Moreover, the effect of electrical stimulation on long term injured individuals is different to those who have been recently injured. This difference is more noticeable when the long term injured individual has lacked the suitable physical activity for a long time. Neuromuscular fatigue is the inability of an individual muscle or muscle group to sustain the required or expected force regardless of the task to be performed (Mizrahi (1997). In comparison with nondisabled muscles, fatigue happens faster in paralysed muscles (Vignes, 2004). Muscle fatigue can be divided into natural and premature fatigue. Premature fatigue happens when muscle force declines before even the muscle reaches 50% of its working capacity. Researchers have identified different elements of the stimulation signal to be affecting the muscle fatigue which includes stimulation pattern and stimulation parameters (Sisto et al., 2008, Faghri et al., 2009). Variable firing rate, intermittent stimulation, doublets and N-lets are some of the commonly used patterns of stimulation while frequency or pulse width is mainly the modulated parameter which are controlled in order to perform a functional task (Mizrahi, 1997).

In animal testing, variable firing rate is shown that it reduces the fatigue (Binder-McLeod and Baker, 1991). While using this method has been proved to reduce fatigue, however it resulted in the decline of maximum achievable force (Mizrahi, 1997). The intermittent stimulation can be divided into regular and irregular intermittent stimulation patterns. Although it is known that irregular intermittent stimulation pattern results in less fatigue but the effect of fatigue and recovery are highly dependent on the prearranged stimulation parameters and allocated resting time (Giat et al., 1996). Doublets and N-lets are another type of pattern for stimulation (Routh and Durfee, 2003). For FES and muscle monitoring applications, there are advantages to doublet stimulation over a range of activation levels that modulate motor unit recruitment; the

very famous one is the reduction in muscle fatigue (Freeman and Durfee, 2006). Karu et al. (1995) reported that optimal N-let stimulation resulted in 36% increase in isometric torque tracking when compared to traditional singlet stimulation. They conclude that there is considerable subject to subject variation in the results.

The findings on using different stimulation parameters have often proved contradicted to each other. Binder-Macleod and Snyder-Mackler (1993) that stimulation frequency and pulse intensity have a supreme impact on muscle fatigue. In 1999 Binder-Macleod and Russ concluded that muscle fatigue was greater at lower frequencies in intermittent stimulation while opposite results have been obtained during continuous stimulation (Kesar and Binder-Macleod, 2006). Furthermore, several studies have suggested different ways to overcome muscle fatigue including using random modulation (Granham et al., 2006, Trasher et al., 2005), N-let pulse trains (Karu et al., 1995) and variable frequency pulse trains (Mourselas and Granat, 1998). Trasher et al. (2005) and Granham et al. (2006) concluded that random modulation of the stimulation frequency, amplitude and pulse-width did not have any effect on the muscle fatigue rate. Routh and Durfee (2003) and Bigland-Ritchie et al. (2000) proposed doublet stimulation signal to reduce muscle fatigue during FES. However, Routh and Durfee (2003) wrap up that doublet stimulation signal did worsen fatigue reduction and singlet stimulation signal led to 33 more cycles than doublet, with the same stimulation parameters. Another difficult technique which is difficult to implement practically in closed-loop FES control applications is what Bigland-Ritchie et al. (2000) proposed that uses doublet stimulation signal for the first 2 minutes and then keeping on with singlet stimulation. They conclude that this method will cut muscle fatigue drastically as compared with constant rate trains.

In another study on muscle fatigue using healthy subjects only Kesar et al. (2007) found that stimulation with frequency modulation gives less muscle fatigue rate compared with stimulation with pulse-width modulation. However, stimulation with frequency modulation is impossible with currently available off the shelf programmed stimulators since most of the stimulators allow pulse-width modulation with fixed frequency. Moreover, in their paper, they found that constant stimulation signal gives the least muscle fatigue compared to the others. Previously, Kesar and Binder-Macleod (2006) also concluded that using the lowest stimulation frequency and longest pulse duration could maximize muscle performance if the stimulation frequency and intensity are kept constant. Earlier, Mourselas and Granat (1998) performed the same experiment

with five healthy subjects and one SCI subject with which the results of Kesar et al. (2006, 2008) agree and conclude that stimulation with frequency modulation increases muscle performance, but this effect is very small and almost non-existent for many subjects. Clearly the literature lacks a concluding remark on the effect of stimulation regimes and parameters in spinal cord injured individuals. Studies lack a targeted investigation into the effect of electrical stimulation on premature muscle fatigue in spinal cord injured individuals especially those with unattended care since the time of injury.

The stimulation pulses can be generated using open loop controllers (Buckett et al., 1988, Hoshimiya et al., 1989, Adamczyk and P. E. Crago, 2000) and closed-loop controllers (Crago et al., 1980, Chizeck et al., 1991, Watanabe, et al., 2002). Control strategies such as fuzzy logic (Feng and Andrews, 1994, Kuen and Chizeck, 1994, Chen et al., 2004, Davoodi and Andrews, 2004), PID (Massoud 2007, Hussain, 2009), neural networks (Chen et al., 2004; Graupe and Kordylewski, 1994), adaptive fuzzy logic (Feng and Andrews, 1994, Ezenwa et al., 1991) and genetic algorithm (Davoodi and Andrews, 1999) are used to regulate the stimulation pulses. In all these control methods the target was for the controller to provide a smooth movement with the least disturbances in the system. Comparing the results from these different control strategies shown that they were all been able to track down the required movement however due to the fact that they employ different stimulation parameters therefore inflicting different amount of energy into the muscles an investigation is essential to see what is their effect on the muscle fatigue. Moreover, in order to provide a smooth and accurate movement the use of mechanical support elements in the design has proven to be of great help. Davoodi and Andrews (1999), Gharooni et al. (2001), Wheeler et al. (2002), Durfee et al. (2005). Massoud (2007) and Hussain (2009) have used springs, breaks and two bar mechanism to facilitate the exercise using FES. The FES exercise facilities can still greatly benefit from the extensive usage of mechanical support facilities such as rotary dampers implemented in the two bar mechanism which will make sure the stimulated muscles are exercising in safe environment and that the stimulation will not resolve in the premature muscle fatigue.

2.6 Summary

In this chapter the anatomy of nervous system, its components and how it functions has been described. SCI has been defined and the complication arises from it is briefly discussed. The SCI causes and its demographic in different countries been summarised before arguing how this devastating injury is classified. The associated problems arising from the SCI are also addressed in this chapter. Due to the primary and secondary complications with SCI an economic burden is posed on the society which has been also discussed in this chapter. FES as a means of producing contraction in the muscles paralyzed due to SCI, the way it functions and recruits motor units, its advantageous and disadvantageous have been extensively discussed. Finally, a brief overview of the literature around fatigue induced in the paralysed muscles when they are under stimulation is presented and the lacks in the literature is identified.

Chapter 3

Dynamic Humanoid Modelling

3.1 Introduction

In any scientific field developing an accurate model to represent the dynamics of the system adequately is the main issue for simulation and control design. The more information available with regard to the details of the actual system, the more accurate the modelling process and therefore the more precise simulation and control design.

In this chapter the process of development of the humanoid model for the purpose of this thesis using two different methodologies is discussed and the results are assessed on a comparative bases. The model is first developed using the MSC.visualNastran 4D (VN4D), a CAD based modelling software. A simpler method using a combination of basic mathematical equations and Simulink modelling is also employed and this is used to evaluate the performance of the VN4D model in characterising the dynamic behaviour of the system.

In order to create a precise model, passive and active parameters of the knee joint need to be identified and integrated into the physical model. In this chapter passive parameters such as stiffness and damping which will be used in the modelling phase are measured using pendulum test on 15 SCI subjects. The active parameters which are more to do with the relationship between stimulation signal and active torque are discussed in Chapter 4. In order to incorporate the weight and the length of the body segments, specifically lower extremities, anthropometric data is used.

3.2 Anthropometry

Anthropometry is the application of scientific physical measurement methods to human subjects for the development of engineering design standards and specific requirements and for evaluation of engineering drawings, mock-ups, and manufactured products for the purpose of assuring suitability of these products for the intended user population (NASA, 2011). It is the single most universally applicable, inexpensive, and non-invasive method to assess the size, proportions, and composition of the human body (Corish and Kennedy, 2003). The French criminologist and anthropologist, Alphonse Bertillon (1853–1914) created the first system of physical measurements, photography, and record-keeping that has been used by police in order to identify recidivist criminals (U.S National Library of Health, 2011). Anthropology and obtaining anthropometry data has been further developed since then. Currently the main interest in body segment parameters is due to motion analysis and prosthesis design (Bjornstrup, 1995).

Anthropometric data by Winter (2009) is one of the most valid and common estimation techniques. Winter (2009) showed that the body segments' length, weight, centre of mass, density and volume can be expressed as a fraction of the total body height (H) and weight (M). In order to define position of structures relative to each other and movement of various parts of the human body anatomical planes of the body are defined: Sagittal plane divides the body into left and right, coronal or frontal plane divides it to front and back and the transversal or horizontal plane divides it into upper and lower parts. Figure 3.1 illustrates this while it will help to define some other relative terminologies.

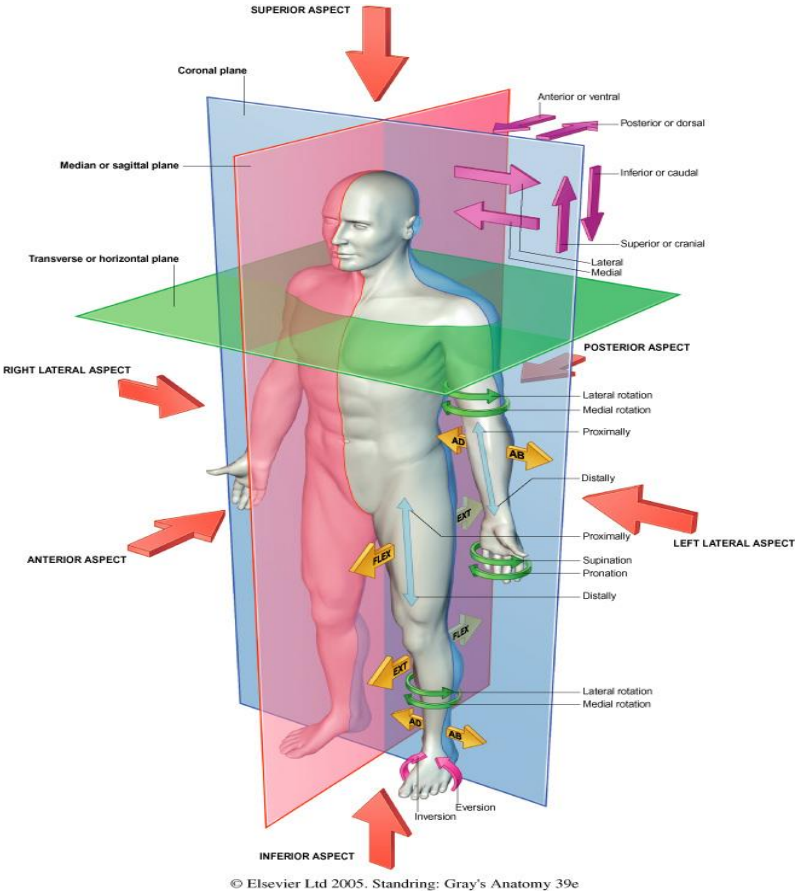


Figure 3.1: Anatomical planes and illustration of different descriptive terminologies (Gray's 2005)

The simulation results in this thesis are based on a population of 15 randomly chosen long term injured paraplegic subjects. Their heights and weights have been measured and averaged in order to build a precise model representing this population. Table 3.1 represents the demographic data of this group.

Table 3.1: Demographic data of the subjects. (Mean ±SD)

Variables		Paraplegic SCI patients (n=15)
Age (years)		34.60 ± 9.18
Weight (kg)		70.86 ± 14.45
Height (cm)		169.80 ± 11.02
Sex	Male	(n=8) 53.3%
	Female	(n=7) 46.7%
Cause of SCI	Motor vehicle accident	(n=9) 60%
	Fall	(n=2) 13.33%
	Gunshot	(n=1) 6.66%
	Lifting heavy objects	(n=1) 6.66%
	Myelin Virus	(n=2) 13.33%

Since the results were collected throughout a joint program in Iran, they are validated by comparing to a national study by Haghdoost et al., (2008) based on a large scale national population- based survey that recruited 89,532 healthy subjects aged from 15 to 64. It is also compared to a regional study by AI-Haboubi (1992) on cross sectional Middle Eastern and West Asian population living in Saudi Arabia. It should be noted that the estimated weight and height in Haghdoost et al., (2008) and AI-Haboubi (1992) for the general population have not been used in the process of modelling but given as a guideline to compare the paraplegic sample to the normal population and illustrate the slight difference between these two groups. It should be noted that while Haghdoost's study did not include the weight however the averaged weight for the studied population is calculated using relevant height to weight ratios for an average individual. Table 3.2 compares the results of Haghdoost and Al-Haboubi with the paraplegic sample used in this study.

Table 3.2: Comparison between Al-Haboubi, Haghdoost and the paraplegic sample

	Height (m)	Weight (kg)
Al-Haboubi	1.70±0.67	69.6±11.7
Haghdoost	1.67±0.74	68.74 ± 12.07
Paraplegic sample	169.80 ± 11.02	70.86 ± 14.45

Based on Winter (2009) anthropometric calculation of and using the paraplegic height and weight model presented earlier the lower extremities body segments' length, mass, centre of mass, density and volume were calculated. Figure 3.2 presents the graphical overview of segments' length calculation while Table 3.3 outlines the estimated segmental values used in this project.

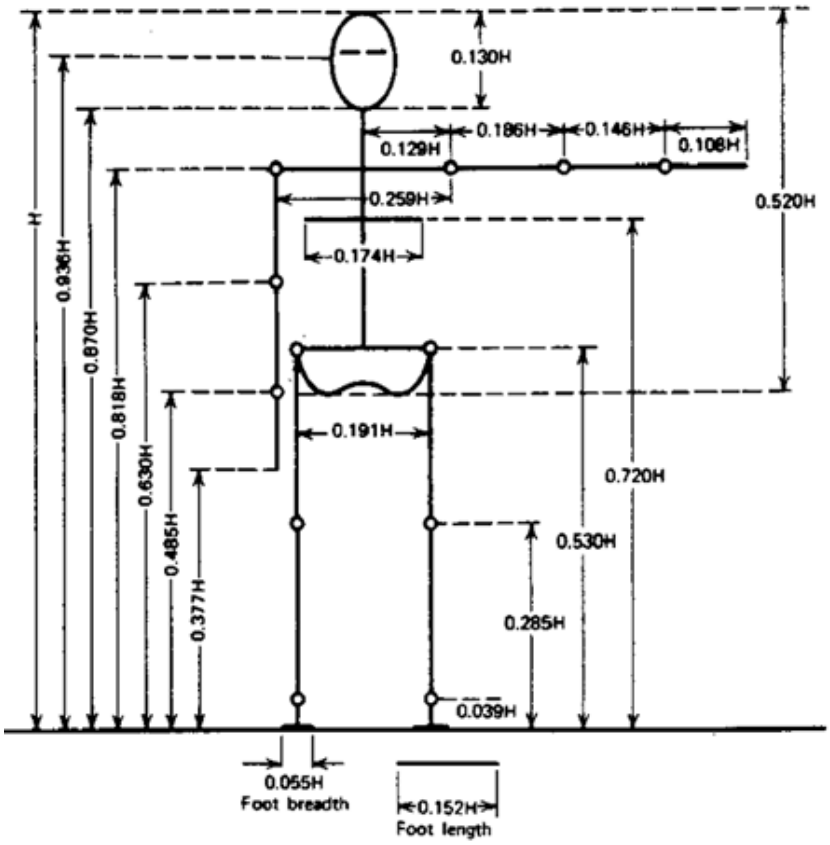


Figure 3.2: Standard human dimensions (Winter, 2009)

Table 3.3: Anthropometric data based on paraplegic sample

Segment	Length [m]	Weight [Kg]	Centre of mass Proximal [m]	Density Kg/l
Hand	0.1944	0.42	0.098366	1.16
Lower arm	0.2628	1.12	0.113004	1.13
Arm	0.3348	1.96	0.1459	1.07
Total arm	0.792	3.5	NA	NA
Head	Radius 0.1	4.65	NA	0.111
Neck	0.0936	1.16	NA	NA
Trunk	0.5184	34.79	0.2592	1.03
Foot	0.2736	1.015	0.1368	1.10
Shank	0.4428	3.255	0.192	1.09
Thigh	0.441	7	0.190	1.05

3.3 Humanoid modelling using VN4D

MSC.visualNastran 4D (VN4D) is a CAD based modelling software used to simulate a moving mechanism. Previously humanoid models have been created using VN4D software for use in the simulation of FES based activity with acceptable results (Gharooni, 2002;Massoud, 2006; Huq, 2009; Hussain, 2009; Jailani, 2011). For the purpose of this project lower extremities' VN4D humanoid model is developed using appropriate anthropometric data and accurate passive parameters based on actual tests which will be more elaborated in section 3.5.

3.3.1 Visual Nastran software

VN4D is a windows® -based CAD software that is able to model and simulate any moving system where by coupling with Matlab®/Simulink® package merges two

powerful toolboxes into one system that allow the user to control multibody dynamics perfectly (Altidis, 2002). The VN4D is able to combine CAD elements with motion, finite element analysis (FEA), and control technologies into a single functional modelling system. It can be linked to Matlab®/Simulink® through a simple block called vNPlantBlock and therefore the control commands can be sent, feedback received and as a result a set of parameters such as velocity, position and torque can be controlled and recorded. Unfortunately the contract between Mathwork® and MSC Software® did not extended after Matlab® version 7.0 so some of the newer tool boxes and futures of the newest version of Matlab®/Simulink® cannot be used (Aertia, 2011; Mathworks®, 2008).

VN4D is named for four parts of the design cycle, namely draw it, move it, break it, and control it (Wang, 2001). The draw it tools consist of CAD integration with CAD programs as well as native three dimensional (3D) solid modelling tools. The move it tools consist of sophisticated motion analysis combined with key-framed animation. VN4D measures force, torque, position, velocity, acceleration, kinetic energy, etc. Bodies are joined with constraints in VN4D through coordinate frames (Coords). Break it combines the motion simulation with FEA. This analysis shows the stresses developed by the parts due to the different forces applied. Control it consists of integration with MATLAB/Simulink and links through Visual Basic to Excel and Excel macros. Visualizing a MATLAB/Simulink control system with real CAD geometry is an important advantage of VN4D (Massoud, 2006). Almost any type of parameter imaginable is measurable including, but not limited to, displacement, velocity, acceleration, power, frictional force, impact-force, and mode shapes (Anderson, 2002).

3.3.2 Humanoid modelling

The humanoid model is built up using anthropometric data. In this work, the anthropometric estimation proposed in Table 3.3 is utilised. In reality in order to reduce the lower extremities degree of freedom (DoF) the thigh is fixed using a cushioned velcro while the ankle is fixed to a very light weighted carbohydrate ankle foot orthosis providing the leg with movement in the sagittal plane only. The leg model consists of three body segments (thigh, shank, and foot) and two joints (knee and ankle). The thigh is fixed horizontally to represent the seat position while the shank is left to swing freely

around the knee joint. The knee joint is represented as a motor constraint in VN design, which is driven by a controlled torque. On the contrary the ankle joint is rigid in order to replicate the situation properly. Using the location of centre of mass (COM) together with segment's density and the general assumption of segment's shape one is allowed to determine the shape of the segment. The density of each segment was used to determine its volume which then determines the segment's width. The overall properties of human joints are based on the paraplegic sample and have been presented in Table 3.3. For the purpose of modelling the joints in the VN4D software the way the joints are moving in an actual human is considered. Table 3.4 summarises the properties of these joints as used in the modelling and simulation.

Table 3.4: Properties of the human joints

Joints	Number	Type	Axis of rotation	Control parameter	DOF
Head	1	Rigid	N/A	N/A	0
Neck	1	Rigid	N/A	N/A	0
Shoulder	2	Revolute	X	N/A	1
Elbow	2	Revolute	X	N/A	1
Wrist	2	Revolute	Y	N/A	1
Hip	2	Revolute motor	X	Torque	1
Knee	2	Revolute motor	X	Torque	1
Ankle	2	Rigid	N/A	N/A	0

Visualisation of the humanoid model illustrating the location of centre of the mass, location of the joints and the representation of the lower extremities built based on the anthropometric data and the joints properties in Table 3.4 is given in Figure 3.3.

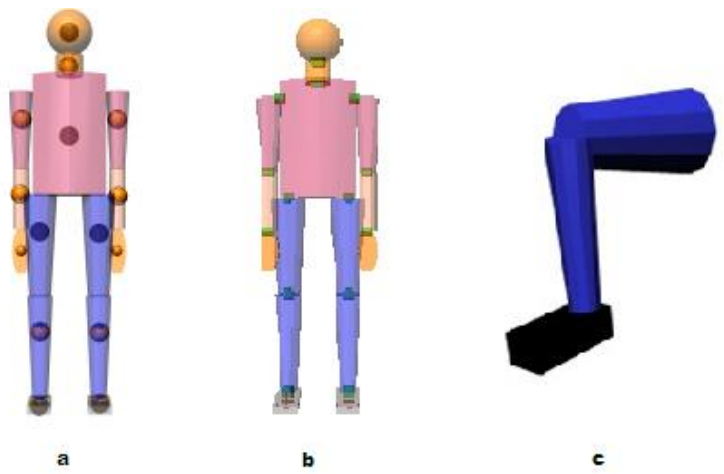


Figure 3.3: The overall look of the humanoid built. a) The humanoid with centre of mass at each segment, b) The humanoid with the joints, c) Representation of the lower extremities

3.3.3 Accessories modelling

For the purpose of this project a leg extension exercise facility is also built. The basic design of the machine is adopted from Keiser's® Air300 leg extension exercise machine (Keiser, 2011) presented in Figure 3.4. However, for the purpose of VN4D simulation the weight stack is replaced by variety of detachable weights which can be placed in the frontal part of the ankle joint and also a gas damper is attached to the SPO which is used to keep the whole shank in sagittal plane and also control the movement of the leg back in the flexion phase. In chapter 7 the actual designed equipment for evaluating the simulation result is presented which will discuss the details of the actual design in further depth. For the purpose of simulation and the actual design, the ankle joint is fixed to the end of the extension bar to make sure that the shank and the weight move at the same time. In the simulation the weight can be added up to 100 kg and the C constant of the damper can be adjusted between 0 and 3. Figure 3.4 shows the Keiser's design while Figure 3.5 illustrates the VN4D designed leg extension equipment .It is assumed that the humanoid does not slip while seated.

Figure 3.4: Keiser's® Air300 leg extension exercise machine (Keiser, 2011)

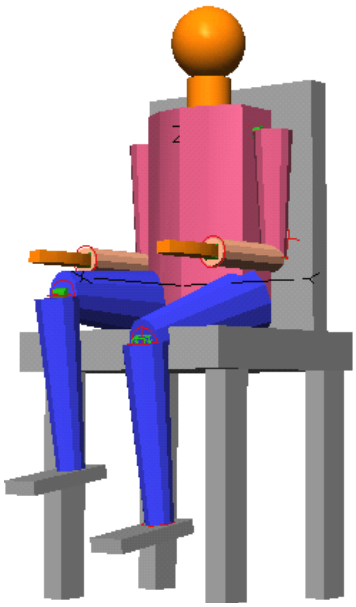


Figure 3.5: VN4D leg extension model

3.4 Pendulum modelling

The human's lower extremities can be divided into two main parts, the thigh and the shank-foot complex. As considered earlier it is desired to reduce the degree of the freedom as much as possible without affecting the essential details in the process of modelling due to the high nonlinearity of the system. In this case the ankle joint is considered rigid and thus the shank and the foot are both be a considered as one part. This assumption is fair as in practical forms of FES activity such as FES rowing, FES cycling or FES leg extension, which we will be discussed extensively in this thesis, the movement of the ankle joint is restricted via different apparatuses. Also, as discussed, in section 3.5 when carrying-out the Watreberg's pendulum test the movement of the ankle joint is restricted by the use of a very light ankle foot orthosis (AFO) so this restriction would not affect the results of the simulation at all.

Combining the anthropometric data gathered earlier together with the dynamics of the system it is obvious that by using the mentioned earlier restrictions the cyclical motion of the lower extremities can be modelled as a simple double pendulum. Knowing the weight and the length of the body segments from the anthropometric data it is possible to calculate the location of the centre of the mass for both parts and therefore by utilising the mechanical toolbox of the Simulink the double pendulum model of the lower extremities can be achieved as presented in Figure 3.6.

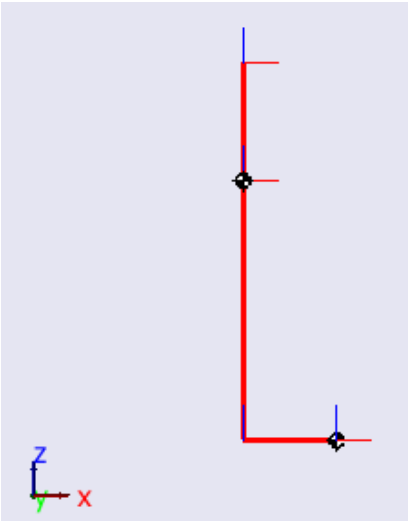


Figure 3.6: Representation of the lower extremities in the form of a double pendulum model with centre of the masses illustrated

The double pendulum has two centres of mass, represented by two black dots in Figure 3.6. The first is the CoM of shank and the second is CoM of foot. The degree of freedom of the second joint is set to zero as discussed earlier. Furthermore, a joint sensor and a joint actuator are attached to the knee joint by connecting these two blocks to a revolute block. The joint sensor measures the angular position, angular velocity, angular acceleration and the torque applied to the joint. The joint actuator supplies the active torque to the joint and the motion input comes from the general SIMULINK signal.

3.5 Implementing the passive knee parameters in the model

In order to develop an accurate lower extremities model representing the paraplegic population many parameters should be considered. Due to the nature of this injury some of these parameters are totally different from healthy population. Inactivity of a joint often leads to chronic changes in the tissues surrounding it, ultimately altering the viscous-elastic properties and passive joint moments. Because of physical inactivity and lack of joint motion following SCI, persons with SCI are susceptible to changes in viscous-elastic properties of the tissues, tightness and shortening of ligaments, stiffening of joint capsules, shortening of muscles (Franken, 1993). Joint contractures and range of motion (ROM) limitation (Riener, and Edrich, 1999). Therefore, measuring and using these parameters for the purpose of modelling is very crucial. Moreover, a quantitative assessment of stiffness, viscosity and joint flexibility is important for the evaluation of the effects of treatment interventions. The main passive parameters which are very important for the purpose of this modelling are stiffness and damping. FES activity can be significantly hampered by changes in viscous-elastic parameters, passive joint moments (Huxley, 1970) and muscle weakness and disuse atrophy. Thus, changes in viscous-elastic parameters should be considered in using FES and in studies of lower-limb neuroprostheses in subjects with SCI (Williams, 2011).

3.5.1 Passive pendulum test

The passive pendulum test is a biomechanical method commonly used to measure viscous-elastic parameters (stiffness, viscosity) during passive swing of the lower limb to represent passive resistances to joint motion associated with structural properties of

the joint tissues and of musculotendinous complex. It is simple and easy to use, specific to the quadriceps (an important muscle for functional activities) and has shown to give reliable and consistent results evaluated the reliability of this method in 96 healthy adults and concluded that the method was reliable and quick for measurement of stiffness in the knee joint. Previous studies used this technique to calculate stiffness and viscosity in patients with arthropathy compared with healthy subjects (Oatis et al., 1995; Valle et al, 2006, Hamstra-Wright 2005). Pendulum test of Wartenberg has also been used to measure passive knee motion with the aim to assess spasticity and rigidity in patients with neurological disorders. Katz et al. (2001) suggested the pendulum test as a robust and practical measure of spasticity with minimal variability in repeated measures. The knee joint is the most measured and standardized joint of SCI patients using this technique by various researchers. The trajectory of the oscillating leg from passive pendulum test presents a set of kinematic parameters such as peak angular values, useful to monitor the changes in the range of knee motion. Lin and Rymer (1998) used this test to understand the underlying neurophysiological disturbances in spasticity. The kinematic outcome depends on a combination of forces acting at the joint. Stiffness and viscosity represent the passive resistances provided by tissues to the angular motion. While stiffness is considered as resistance of an elastic body to resist deformation, viscosity is related to the friction between adjacent layers of tissues. Both parameters may influence the range of motion of knee joint affecting angular displacement. Some investigators measured several kinematic variables using the pendulum and evaluated changes in knee angular displacement, passive stiffness and viscosity to build a leg model. In this work passive pendulum test is used to measure passive knee stiffness, viscosity and damping ratio in a relatively large sample of paraplegic patients and healthy subjects.

3.5.2 Methodology

Experiments were performed on 15 individuals with paraplegic SCI (age: 34.60 ± 9.18 years; height: 169.80 ± 11.02 cm; weight: 70.86 ± 14.45 kg) and 21 able-bodied individuals (age: 30.66 ± 11.13 years; height: 168.67 ± 11.02 cm; weight: 67.10 ± 16 kg) as control group. The subject population in this study was a sample of convenience made up of subjects who were between the ages of 21 and 55 years. They were

consecutive patients who agreed to participate as they fulfilled the inclusion criteria. The SCI subjects (n = 15) had all clinically stable lesions between the neurological levels of T4 to L1, signified by a lack of sensation and voluntary movement below the level of injury. Patients with higher level injuries were excluded to provide a group of patients with similar physiological characteristics. All the subjects signed an informed consent form approved by the human subjects committee before participating in the study. Table 3.1 summarises the physical characteristics of the SCI subjects but for the purpose of clarity and in order to compare the sample sizes of SCI and healthy subjects the breakdown of these two groups is presented in Table 3.5.

Table 3.5: Demographic data of the SCI and healthy subjects. (Mean ±SD)

Variables		Paraplegic SCI patients (n=15)	Healthy subjects (n=21)
Age (years)		34.60 ± 9.18	30.66 ± 11.13
Weight (kg)		70.86 ± 14.45 kg	67.10 ± 16
Height (cm)		169.80 ± 11.02	168.67 ± 11.02
Sex	Male	(n=8) 53.3%	(n=14) 66.66%
	Female	(n=7) 46.7%	(n=7) 33.34%
Cause of SCI	Motor vehicle accident	(n=9) 60%	NA
	Fall	(n=2) 13.33%	NA
	Gunshot	(n=1) 6.66%	NA
	Lifting heavy objects	(n=1) 6.66%	NA
	Myelin Virus	(n=2) 13.33%	NA

For the purpose of the pendulum test the subjects were placed in a semi-upright sitting position, between 45º and 60º in a position that they felt best comfortable, with the shank-foot complex hanging from the edge of a bed as illustrated in Figure 3.7. To

ensure that the ankle and shank complex moved together with the ankle joint fixed at 90º and to reduce the interruption of the passive characteristic of the knee due to ankle movements a very low weight carbon fibre ankle foot orthosis (AFO) was used. The AFO used, AFO Light, is a product of Ossur prosthesis technologies weighting 80 grams for the small version and 90 grams for the medium size version. The subject's shank-foot complex was extended up to 60 degree from the original position and held by the examiner until the knee muscle was completely relaxed. The subject's leg was then released and was allowed to swing and oscillate freely and the shank-foot complex angular movement was recorded until the complex reached its final resting position and stopped movement. The viscoelastic properties of the joint and surrounding tissues, together with the mass of the moving shank-foot complex cause the leg to finally come to rest close to the vertical position. Using a flexible twin axis electronic goniometer (Biometrics Ltd., Model: SG110/A), movement of the leg was recorded during passive pendulum test. To eliminate the interaction of brain while doing the test, during the whole process the subject's eyes were kept close using sleep masks. The test was performed three times and the mean value of three measurements was taken for analysis.

Figure 3.7: Goniometer setup for the pendulum test

Several variables including onset angle (OA), resting angle (RA), first three peak flexion angles (F1, F2, F3), first three peak extension angles (E1, E2, E3), amplitude of

initial flexion (F1Amp= F1-OA), amplitude of initial extension (E1Amp= F1-E1), plateau amplitude (PA= RA-OA), relaxation index (RI=F1Amp/PA), extension relaxation index (ERI=E1Amp/PA) and period of the first cycle (T) were measured from the Waterberg (1951) technique and could be used in deriving several useful variables from kinematics of pendulum test.

Some kinematic data of knee motion and anthropometric measures were used to compute the viscosity and stiffness. Knee stiffness (K) and viscosity (B) were estimated by computing the damping ratio (ζ) and the natural frequency (ω_n) obtained from the test data using the following equations reported.

$$\zeta = \frac{B}{2\sqrt{JK'}} = \sqrt{\frac{(\ln D)^2}{4\pi^2 + (\ln D)^2}} \tag{3.1}$$

$$D = \frac{\theta_1}{\theta_2} \tag{3.2}$$

$$\omega_n = \sqrt{\frac{K'}{J}} = \frac{2\pi}{T} \tag{3.3}$$

The estimation for J and mass characteristics (m and l) were obtained for the subjects according to Winter [13]. The values of viscosity (B) and stiffness (K) were obtained as follows:

$$B = 2 \times \zeta \times \omega_n \times J \tag{3.4}$$

$$K = K' - \frac{mgl}{2} \tag{3.5}$$

where J is the sagittal moment of inertia applied to the shank-foot complex rotation around the knee axis; m is the leg-foot complex mass; g is the gravity acceleration and l

is the leg-foot length from the knee axis; ϑ_1 is the peak angle of one cycle; ϑ_2 is the peak angle of the following cycle and τ is the period of one cycle. To assure reliability of measurements, the same examiner tested all participants. The data were normalized by dividing the results of moment of inertia, stiffness and viscosity by the fifth power of body stature according to the procedure previously described by Jailani (2010).

A statistical analysis was performed using SPSS version 17.0. The data was tested by using 2×2 ANOVA, accounting for gender, health status (paraplegic SCI vs. healthy), and interaction of health status and gender effects. Pooled data was used where there was no significant health status by gender interaction effect. P-value which in general is the quantile of the value of the test statistic, with respect to the sampling distribution under the null hypothesis was calculated. Statistical significant was attributed to P value less than 0.05.

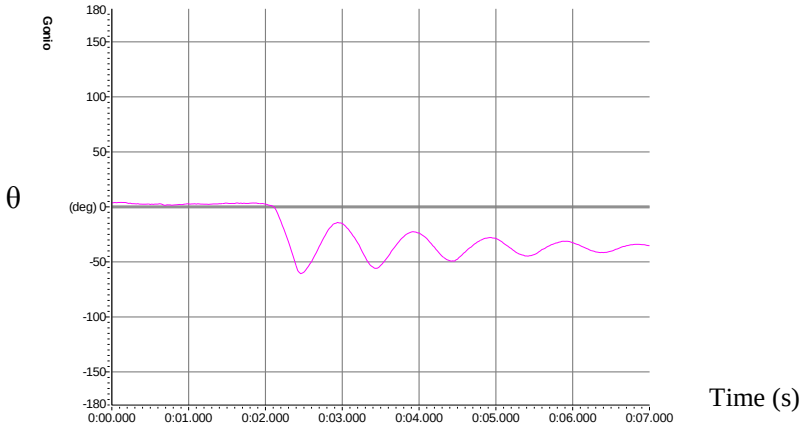
3.5.3 Results

The above statistical analysis showed that there was no statistically significant difference in subjects' age (P=0.27), height (P=0.87) and weight (P=0.51) among the two groups. Pendulum test angular values for one control subject and two paraplegic SCI patients are presented in Figure 3.8.

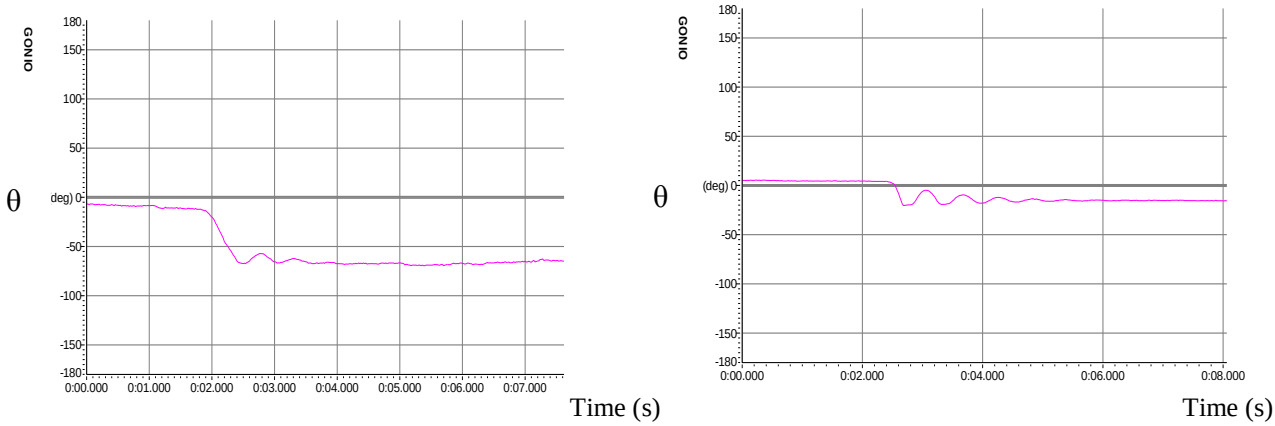
Figure 3.9 depicts the mean values of the first three peak flexion and extension angles (F1, F2, F3, E1, E2, E3) in subjects with and without SCI. Descriptive statistics (Mean±SD) for the kinematic data assessed during the pendulum test in two groups are presented in Table 3.5.

The result of 2×2 ANOVA revealed that gender by health status interaction effect was not significant for stiffness (F=1.16, P=0.28), viscosity (F=0.77, P=0.38) and damping ratio (F=1.08, P=0.30) at $\alpha = 0.05$. Therefore, pooled data was used to compare means of these variables across patients with SCI and healthy subjects. No significant difference was found in stiffness (P=0.15), viscosity (P=0.23) and damping ratio (P=0.60) between healthy subjects and those with paraplegic SCI (Table 3.6, Figure 3.10), although there was a trend towards greater stiffness and viscosity in patients with SCI compared to the healthy subjects. Except for relaxation index (RI), all other

displacement parameters exhibited no statistically significant difference between normal subjects and SCI patients.



a): Typical electrogoniometric output during pendulum test



b) Pendulum test angular response for a paraplegic SCI patient

Figure 3.8: Pendulum test angular values for one healthy subject and two paraplegic SCI patients

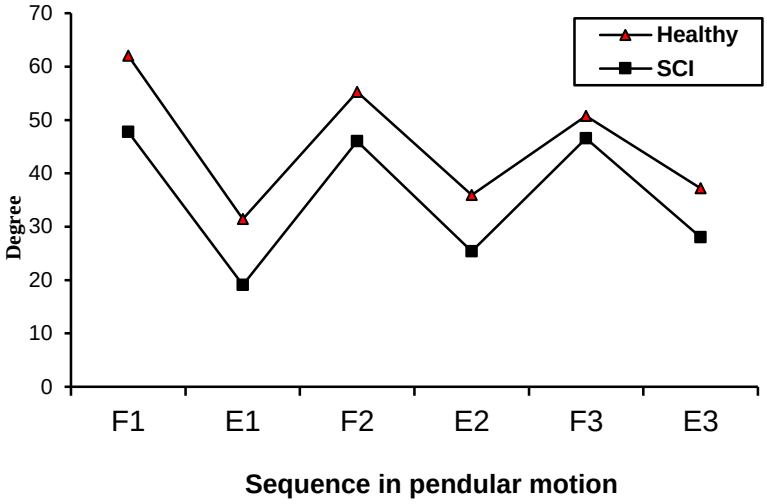
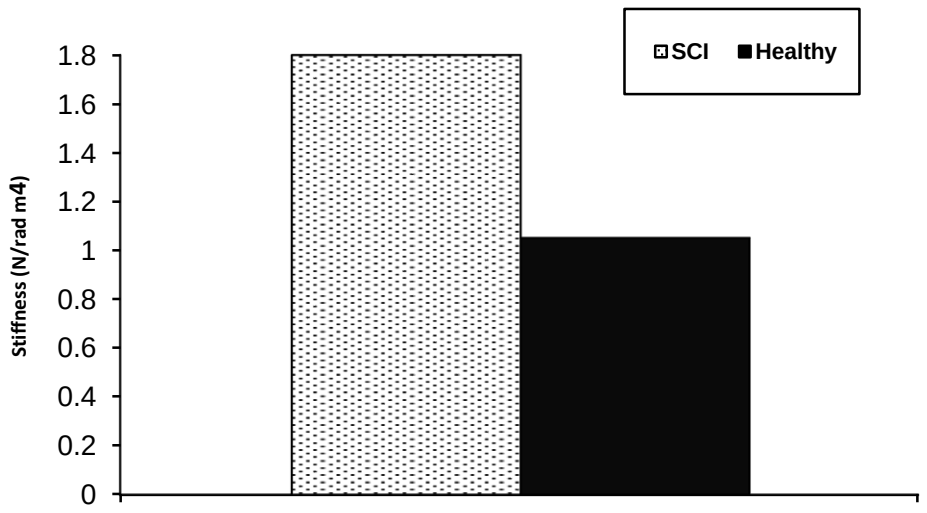


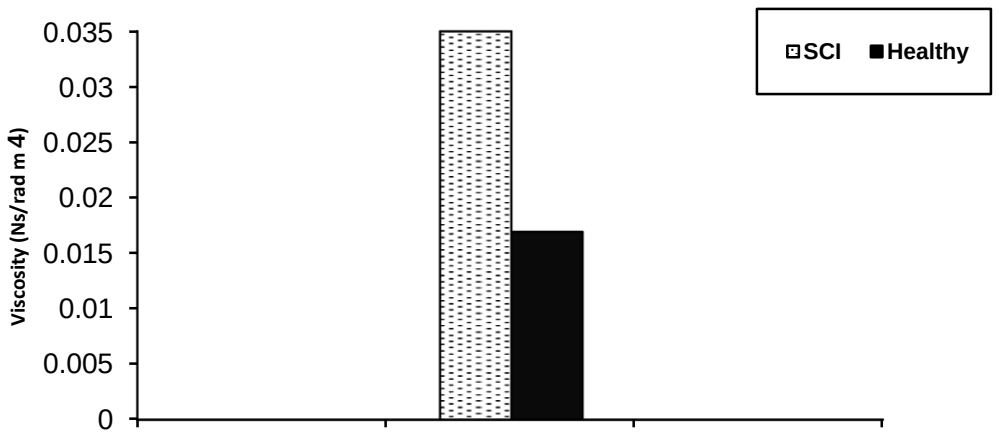
Figure 3.9: The mean values of the first three peak flexion and extension angles (F1, F2, F3, E1, E2, E3) in patients with SCI and healthy subjects

Table 3.6: Descriptive statistics (Mean ± SD) for the kinematic data assessed during the pendulum test in each group

Variables	Paraplegic SCI patients (n=15)	Healthy subjects (n=21)	P-value
Onset Angle	12.28 ± 9.06	11.87 ± 12.51	0.91
$F1_{Amp}$	46.49 ± 9.36	69.35 ± 10.52	0.15
$E1_{Amp}$	26.68 ± 16.96	32.27 ± 12.98	0.28
Plateau amplitude (PA)	37.82 ± 14.48	35.34 ± 13.44	0.60
Relaxation index (RI)	1.26 ± 0.48	1.71 ± 0.60	0.02
Extension relaxation index (ERI)	0.73 ± 0.44	1.05 ± 0.65	0.07
Damping ratio	0.013 ± 0.019	0.010 ± 0.007	0.60
Stiffness (N/rad m^4)	1.80 ± 1.66	1.05 ± 1.43	0.15
Viscosity (Ns/rad m^4)	0.035 ± 0.064	0.017 ± 0.017	0.23



a) Stiffness



b) Viscosity

Figure 3.10: The stiffness and viscosity values in two groups

3.6 Final model

Based on the performed passive experiments, the mean stiffness, viscosity and sagittal moment of inertia of the paraplegic sample were evaluated. By incorporating these parameters into the already built physical model using anthropometric data a finalised model to be used in the simulations was achieved. Due to the cross sectional subject selection and by having subjects representing different levels of injuries, age range, height, weight, gender and the cause of injury the finalized model represents the actual paraplegic society at its best. Table 3.7 summarizes the essential parameters needed to build a passive model of the lower extremities. These parameters would be then incorporated into the two modelling approaches, the VN4D and the pendulum modelling and the results are compared in section 3.6.1.

Table 3.7: Summary of essential passive parameters

Passive parameters	**Values**
Stiffness (N/rad m^4)	1.80
Viscosity (Ns/rad m^4)	0.035
Moment of inertia (kgm^2)	0.36

The parameters shown in Table 3.7 were incorporated in both the VN4D and the pendulum models to provide a proper representation of the lower extremities for the sample SCI subjects. The two constructed models of the lower extremities were extended for 60 degrees by applying the necessary torque to the motors located in the knee joint and then released resulting in a pendular movement which damped after a few seconds. The mean values of the first three peak flexion and extension angles of the SCI subjects were compared to the first three peak flexion of the VN4D and pendular models and the results are presented in Figure 3.11.

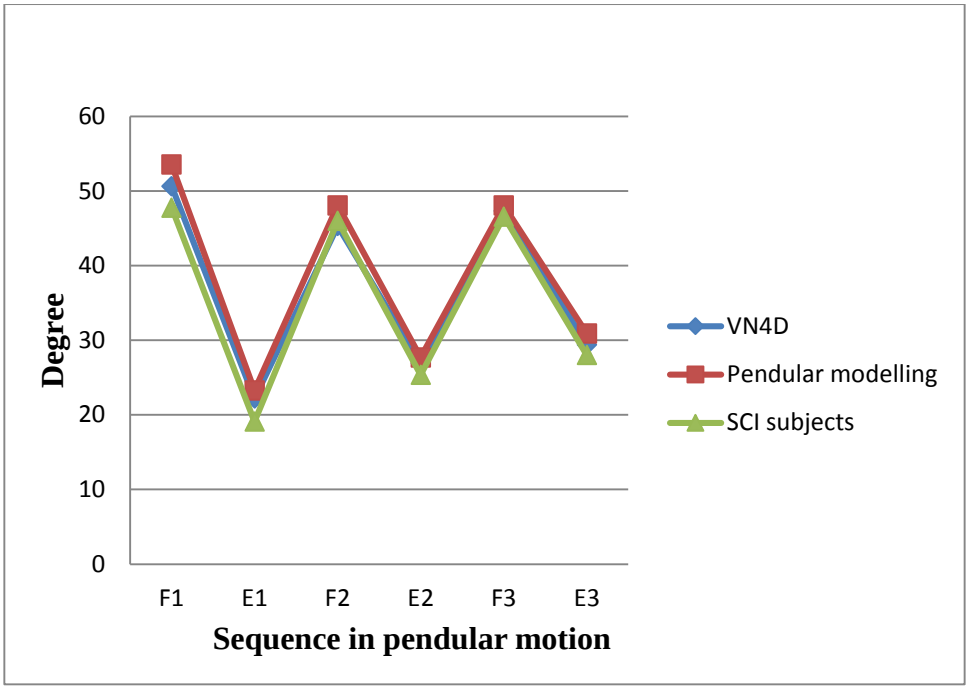


Figure 3.11: Comparison of the models with the actual subject's data

Based on Figure 3.11 it appears that the VN4D model represented the SCI sample relatively better. This is due to the fact that in the VN4D model the weight is distributed in the body parts not evenly but in accordance with the their and the shape, especially the lower extremities, while in the pendulum modelling the mass is located at the centre of the mass. It is also important to note that the types of joints employed in the two methods are also different adding to the difference noticed in the models. Having said this, the averaged error of the 3 first extension and flexion angles for the pendular modelling compared to actual subjects was 8.98% while the error between the VN4D and the actual subjects was 5% which is good supporting argument for the use of VN4D as the modelling technique. Moreover, VN4D is an easy to use platform which will make available the use of mechanical features during the modelling. To compare the effects of this difference in the quality of the control of the system, a brief comparison between the VN4D and pendulum modelling is provided in chapter 6.

3.7 Summary

To elicit an accurate simulation, designing a general model which represents the actual system at its best is necessary. To ensure that the developed model is representing the

SCI subject accurately 15 subjects have been recruited and essential background history and measurements have been collected from them. These measurements have been transferred to anthropometric data and then used in forming the physical structure of the model. It has been assumed that like most of the FES based activity the ankle joint is fixed and therefore the system's degree of the freedom is reduced to one. Moreover, to get a more realistic lower extremities model the passive parameters needed to be incorporated in the knee joint, and these have been calculated using the Waterberg's pendulum test and the average results have been integrated into the finalised model.

In this chapter two types of models have been created in order to represent the lower extremities. The first one is the VN4D model which was created using Visual Nastran software. The model and the accessories needed for the purpose of simulation have been built in this environment and proved able to represent the actual SCI sample. The second type of the modelling techniques used has been pendulum modelling of the lower extremities in which the shank-foot complex has been treated as an L-shaped pendulum moveable around the knee joint. The segments masses have been place at the centres of the mass calculated using the available data. Both models have been created using anthropometric data and passive parameters available from the measurements and calculations.

The finalised models have been tested against each other by extending them to 60º from their resting position by applying an appropriate amount of torque to the motor placed at the knee joint and then releasing them to continue a pendular movement till the movement was damped. The recorded movements have been compared to the 15 sets of pendulum tests of the subjects and it has been decided to use VN4D model throughout this project based on the relatively better performance of the model and also better platform for design and adjustment of the current model. However, the performances of both models are compared in chapter 6 to evaluate them when they are controlled as well via the use of different control techniques.

Chapter 4

Muscle Model

4.1 Introduction

Muscle is an organ which transforms chemical energy into movement and builds up between 40% and 60% of the weight of the human body. In general there are three types of muscle; skeletal, smooth and cardiac muscles. Skeletal muscle is made up of long fibres making up the majority of the human body's muscle weight. The skeletal muscle which is also known as voluntary muscle is responsible for locomotion and can be delicately controlled. Smooth muscle is found in the walls of internal organs and is built of packed spindle-shaped fibres. Cardiac muscle is found only in the heart and is composed of short interconnecting fibres. Smooth and cardiac muscle's contraction happens unconsciously via autonomic nerves system.

The skeletal muscle is innervated by the somatic nervous system and is subject to conscious control. It can be stretched beyond its resting length. However it is elastic since the muscle fibre can return to its resting length after being stretched. Most muscles cover the joints and are attached to bones in at least two places called insertion and origin. In the muscles of limbs, the origin typically lies proximal to the insertion. Skeletal muscle is excitable so it can respond to the stimulus by contracting. It is the main tissue which is stimulated by FES to produce functional movement in disabled people. In this thesis wherever the term muscle is used it refers to skeletal muscle as this is the only muscle type controllable consciously and provides locomotion and posture maintenance.

In order to capture the complicated actions of human body's skeletal muscles a number of simplified versions of muscles mainly named muscle models have been developed. Archibald Hill (1938) has created his very popular muscle model known as Hill model based on experiments performed on the limbs of frog. The Hill model is purely a mechanical muscle model, hence the lack of physiological knowledge when the model is proposed. In another attempt and based on sliding filament theory Huxley and Simmons (1971) have created their own muscle model based on the actual molecular structure of the muscle. Since then many efforts have been dedicated to developing

accurate representation of muscles some of which include the attempts made by Ferrarin et al. (1996, 2000) and Reiner (1996, 1997).

In this chapter the microscopic anatomy of skeletal muscles is reviewed and the natural mechanism in which the muscles contract in order to produce movement is also presented. The adopted muscle models and the way they are built up are discussed and a structure is derived on how to develop a muscle model based on a few key goals. Based on this structure the advantageous and disadvantageous of some of the current used models are discussed and the need for a simpler and effective muscle model to be used in FES simulation is argued. Modelling the knee joint using MPL type artificial neural network (ANN) and nonlinear autoregressive moving average with exogenous input (NARMAX) is discussed and the advantageous and disadvantageous of each method are pointed out. The NARMAX model proves to be highly stable, however due to the lack of its fatigue predictability another type of muscle model using artificial neuro-fuzzy inference system (ANFIS) theory is used. In later chapters the effect of using a non-fatigue predictable model with a fatigue predictable model on the control aspect of the electrically stimulated muscles is discussed in further detail.

4.2 Microscopic anatomy of skeletal muscles

Skeletal muscles consist of densely packed groups of elongated cells, known as muscle fibres, held together by fibrous connective tissue. Numerous capillaries penetrate this tissue to keep muscles supplied with abundant quantities of oxygen and glucose needed to fuel muscle contraction. The bundles of fibre that make up a muscle like wire within a cable are called fascicles. These are made from a thread like cells which range from 10 to 100 µm in diameter, called muscle fibre which itself consists of many thinner fibres called myofibril. The tiniest element skeletal muscle structure is thin myofilament, made up of actin, and thick myofilament, made up of myosin. Actin and myosin are two major proteins. The myofilaments in each myofibril are divided transversely by Z bands into units called sarcomere. It is through these units that neural impulses stimulate contraction. Thin filament slides along the thick filaments resulting in muscle shortening. Skeletal muscle fibres have two intracellular tubules called sarcoplasmic reticulum (SR) and the T-tubules. The tubules release and regulate the signals for

contraction. (Baggaley, 2001; Marieb, 2001). Figure 4.1 illustrates this structure graphically.

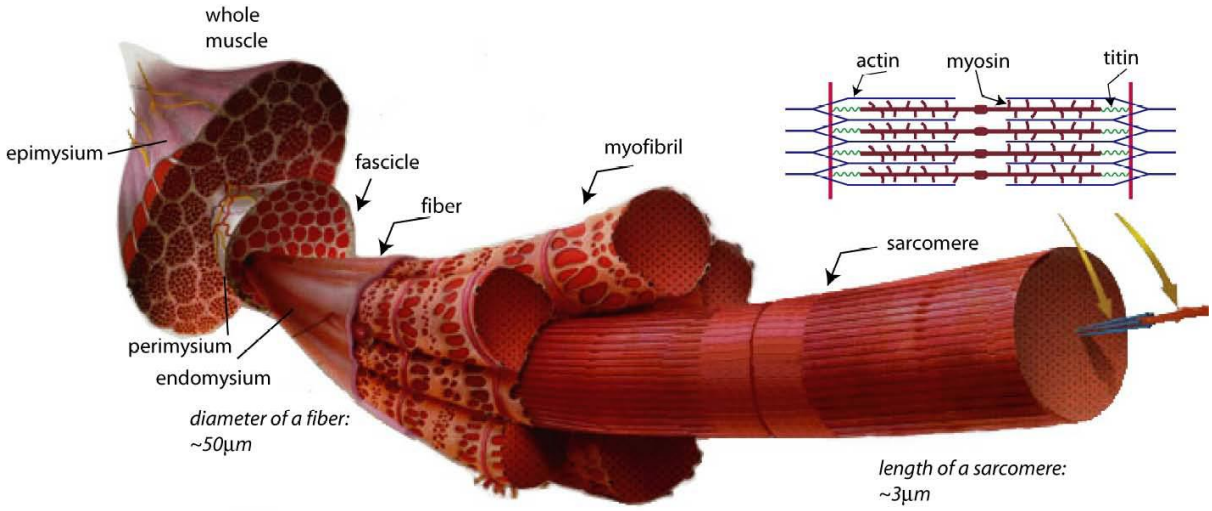


Figure 4.1: Structural and organizational levels of skeletal muscle (Applied Biomechanics,2011)

4.3 Skeletal muscle contraction

Ferrario (2006) explains the muscle contraction as a complex process involving a number of cellular proteins and energy production systems. The action potential causes the nerve cell to fire, inducing a reaction that releases a neurotransmitter (acetylcholine) at the level of the synapse. Thus, the process of muscular contraction begins when a nerve impulse arrives at the neuromuscular junction. The neuromuscular junction is the connection between the neural axon and the muscle fibre. It is at this level that the signal is transferred from one cell to another through the action of acetylcholine, and this causes the depolarization of the muscle cell. The motoneurone passes the signal to the muscle cell and it can supply different muscle fibres. Generally, the slower the electric conduction in a motoneurone is, the fewer fibres it innervates and vice versa. Usually, the fibres innervated by a single motoneurone are not adjacent, but are scattered in the muscle. The motoneurone and the fibres it innervates form a motor unit. When a

motoneurone fires, all the fibres belonging to the corresponding motor unit are activated synchronously. Each motor unit is composed by a varying number of muscle fibres. This ranges from 10 to several thousand fibres, and in a single muscle different sizes of motor units can be found. In general, in muscles used for precisely controlled movements, the size of the motor units is small and so is the number of fibres innervated by a single motoneurone. The diameter of the axon determines its conduction velocity. The larger the diameter the higher the conduction velocity.

4.3.1 The motor unit

Neurones are considered excitable. They must be triggered by a stimulus to produce nerve impulses. When the neuron receives a stimulus, the electrical charge on the inside of the cell membrane changes from negative to positive. A nerve impulse travels down the fibre to a synaptic knob at its end, triggering the release of chemicals (neurotransmitters) that cross the gap between the neuron and the target cell, stimulating a response in the target. The communication point between neurones is called synapse which comprises the synaptic knob, the synaptic cleft and the target site (Baggaley, 2001). Action potential is the transmission of nerve pulses throughout a neuron which ends in the axon and finally to the synaptic knob. Four general stages are recognised for the action potential starting from the resting situation when the threshold is passed. Threshold is the level at which a stimulus is strong enough to transmit a nerve impulse. The process of sending a nerve impulse is illustrated in Figure 4.2.

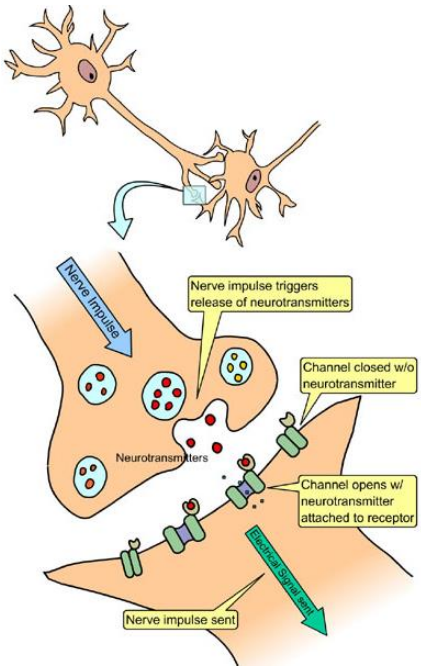


Figure 4.2: The process of generating nerve impulse (Baggaley, 2001)

4.3.2 Generation of action potential

Action potential is a conducted change in the trans-membrane potential of excitable cells, initiated by a change in the membrane permeability to sodium ions. Muscular contraction is explained by the sliding filament model. The sliding filament theory explains what happens to the sarcomere during a contraction, but not the mechanism involved. The contraction of the muscle decreases the distance between couples of Z lines. This happens through the sliding of actin and myosin filaments across each other. For this purpose, the myosin molecule contains heads which can bind to actin. The action potential that arrives from the motoneurone propagates along the T-tubules. This stimulates the sarcoplasmic reticulum to release Ca2+. The Ca2+ binds troponin, which in turn causes a positional change in tropomyosin uncovering the active sites of actin. The actin is thus free to bind with myosin. This process results in the release of energy stored within the myosin molecule, thus causing the shortening of the muscle. The addition of fresh adenosine-triphosphate (ATP) breaks the strong binding state of the cross-bridge, resulting in a weak binding state. The ATPase enzyme hydrolyses the ATP providing energy for attachment to another active site on the actin molecule. The cycle is possible as long as free Ca2+ and ATP are available. The required ATP is in part

stored in the muscle cells and in part produced in real time during exercise through metabolic pathways (Ferrario, 2006; Martini 2005; Lieber and Friden, 2000). Figure 4.3 shows this in more detail.

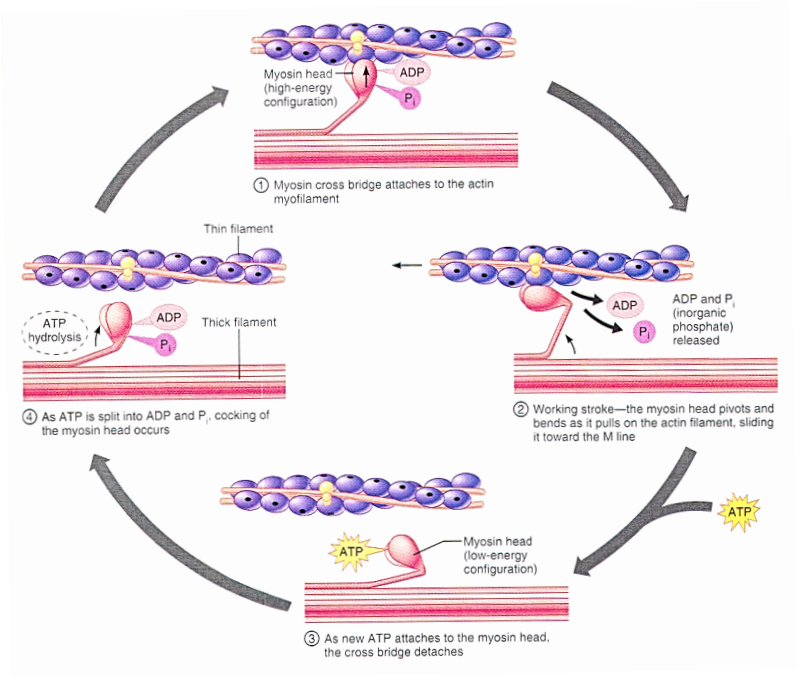


Figure 4.3: Sequence of events involved in the sliding of the actin filaments during contraction (Marieb, 2001)

4.3.3 Different types of muscle contractions

Muscle contractions can be divided into isotonic, isometric and isokinetic. An isotonic contraction results in the muscle length change and thus movement. It is divided into concentric contraction, in which the muscle length decreases against an opposing load, and eccentric contraction, in which the muscle increases in length as it resists a load. An isometric contraction happens when the muscle contracts without any movement or changes in the length. In this case normally the related joints do not move. Isokinetic contractions are alike to isotonic contraction with regard to the changes in length, but producing movements at a constant speed. They are rare to find in daily activity due to their nature (Murray et al., 1980)

4.3.4 Length-tension and force-velocity relationship

In its most basic form, the length-tension relationship states that isometric tension generation in skeletal muscle is a function of the magnitude of overlap between actin and myosin filaments. The more muscle fibres are excited the more muscle force will be, and the greater the cross sectional area of the muscle the greater the muscle strength (NSMRC, 2011). Length-tension curve, relates the strength of an isometric contraction to the length of the muscle at which the contraction occurs. Muscles operate with greatest active force when close to an ideal length (often their resting length). When stretched or shortened beyond this the maximum active force generated decreases (Gordon, 1966).

The force-velocity (F-V) relationship characterizes the dynamic capability of the neuromuscular system to function under various loading conditions and, therefore, has considerable significance in the performance of movement (Fitts and Widrick, 1996). Essentially, the F-V relationship is a hyperbolic curve constructed from the results of numerous experiments describing the dependence of force on the velocity of movement (Hill, 1953). Muscle contraction velocity and its contraction duration depend on the muscle fibre type, the load applied to it and the number of the recruited muscle fibres. The greater the load, the slower the contraction and the shorter the duration of contraction, Also the more activated motor units the faster and the more prolonged contraction (Cronin et al., 2003; NSMRC 2011). Typical length-tension and force-velocity relationships are presented in Figure 4.4.

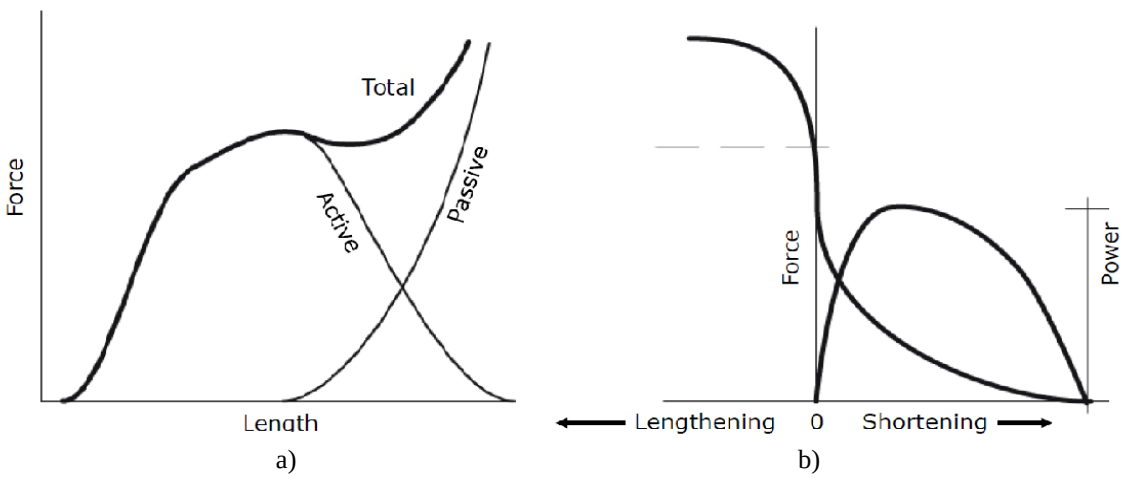


Figure 4.4: a) length-tension relationship b) force-velocity relationship (Gordon, 1996)

4.3.5 Types of skeletal muscle fibres

The human body has three major types of skeletal muscle fibres: fast fibres, slow fibres, and intermediate fibres (Martini 2007).

Most of the skeletal muscle fibres in the body are called fast fibres, because they can contract in 0.01 sec or less after stimulation. Fast fibres are large in diameter; they contain densely packed myofibrils, large glycogen reserves, and relatively few mitochondria. The tension produced by a muscle fibre is directly proportional to the number of sarcomeres, so muscles dominated by fast fibres produce powerful contractions. However, fast fibres fatigue rapidly because their contractions use ATP in massive amounts, so prolonged activity is supported primarily by anaerobic metabolism. Several other names are used to refer to these muscle fibres, including white muscle fibres, fast-twitch glycolytic fibres, and Type II-A fibres.

Slow fibres are only about half the diameter of fast fibres and take three times as long to contract after stimulation. Slow fibres are specialized to enable them to continue contracting for extended periods long after a fast muscle would have become fatigued. The most important specializations improve mitochondrial performance. Slow muscle tissue contains a more extensive network of capillaries than is typical of fast muscle tissue and so has a dramatically higher oxygen supply. In addition, slow fibres contain the red pigment myoglobin. This globular protein is structurally related to hemoglobin, the oxygen-carrying pigment in blood. Both myoglobin and hemoglobin are red pigments that reversibly bind oxygen molecules. Although other muscle fibre types contain small amounts of myoglobin, it is most abundant in slow fibres. As a result, resting slow fibres contain substantial oxygen reserves that can be mobilized during a contraction. Because slow fibres have both an extensive capillary supply and a high concentration of myoglobin, skeletal muscles dominated by slow fibres are dark red. They are also known as red muscle fibres, slow-twitch oxidative fibres, and Type I fibres.

With oxygen reserves and a more efficient blood supply, the mitochondria of slow fibres can contribute more ATP during contraction. Thus, slow fibres are less dependent on anaerobic metabolism than are fast fibres.

Some of the mitochondrial energy production involves the breakdown of stored lipids rather than glycogen, so glycogen reserves of slow fibres are smaller than those of fast fibres. Slow fibres also contain more mitochondria than do fast fibres. Figure 4.5 compares the appearance of fast and slow fibres.

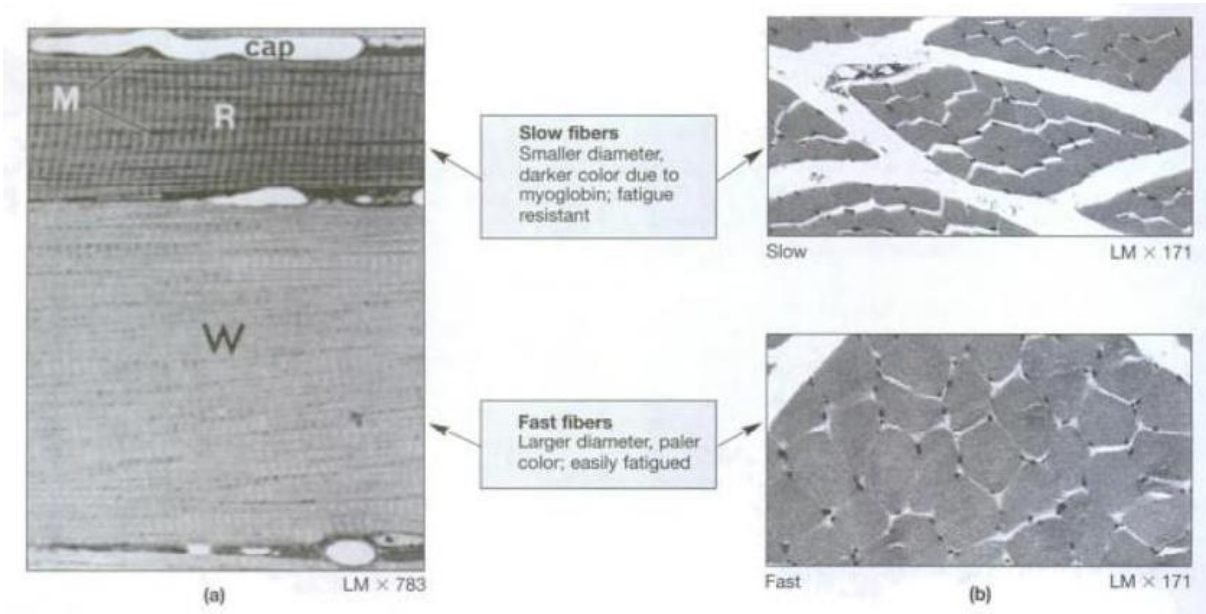


Figure 4.5: The appearance of fast and slow fibres, R for red, W for white, Cap for capillary supply, M for mitochondria. a) The general setting of slow and fast fibres, slow fibres have more extensive capillary supply and more mitochondria. b) Zoomed in cross section of slow and fast fibres (Martini, 2007)

The properties of intermediate fibres are intermediate between those of fast fibres and slow fibres. In appearance, intermediate fibres most closely resemble fast fibres, for they contain little myoglobin and are relatively pale. They have a more extensive capillary network around them, however, and are more resistant to fatigue than are fast fibres. Intermediate fibres are also known as fast-twitch oxidative fibres and Type II-B fibres.

The three types of muscle fibres are compared in Table 4.1. In muscles that contain a mixture of fast and intermediate fibres, the proportion can change with physical conditioning. For example, if a muscle is used repeatedly for endurance events,

some of the fast fibres will develop the appearance and functional capabilities of intermediate fibres. The muscle as a whole will thus become more resistant to fatigue.

Table 4.1: Properties of skeletal muscle fibre type (Martini, 2007)

Property	Slow	Intermediate	Fast
Cross-sectional diameter	Small	Intermediate	Large
Tension	Low	Intermediate	High
Contraction speed	Slow	Fast	Fast
Fatigue resistance	High	Intermediate	Low
Color	Red	Pink	White
Myoglobin content	High	Low	Low
Capillary supply	Dense	Intermediate	Scarce
Mitochondria	Many	Intermediate	Few
Glycolytic enzyme concentration in sarcoplasm	Low	High	High
Substrates used for ATP generation during contraction	Lipids, carbohydrates, amino acids (aerobic)	Primarily carbohydrates (anaerobic)	Carbohydrates (anaerobic)

The percentages of fast, intermediate, and slow fibres in a skeletal muscle can be quite variable. Muscles dominated by fast fibres appear pale and are often called white muscles. Chicken breasts contain "white meat" because chickens use their wings only for brief intervals, as when fleeing from a predator and the power for flight comes from fast fibres in their breast muscles. As noted earlier, the extensive blood vessels and myoglobin in slow fibres give these fibres a reddish colour; muscles dominated by slow fibers are therefore known as red muscles. Chickens walk around all day, and the movements are performed by the slow fibres in the "dark meat" of their legs.

Most human muscles contain a mixture of fiber types and so appear pink. However, there are no slow fibres in muscles of the eye or hand, where swift but brief contractions are required. Many back and calf muscles are dominated by slow fibres; these muscles contract almost continuously to maintain an upright posture. The percentage of fast versus slow fibres in each muscle is genetically determined. As noted earlier, the proportion of intermediate fibres to fast fibres can increase as a result of athletic training.

4.4 Musculoskeletal modelling

In the field of FES when simulating a proposed musculoskeletal movement using an accurate easy to use muscle model is essential. If the muscle model predicts the muscle behaviour under stimulation accurately it will provide the researchers and clinicians with a proper understanding of the effect of the control strategy implemented and the stimulus properties used. This will result in less bone fracture, less muscle fatigue induced, less time consumed on getting the stimulus properties right and last but not the least it can be used as a feedforward control structure in an online setting.

In 1922, A.V.Hill, first noted that activated muscles produce more force when held isometrically than when they shorten. When muscles shorten they appear to waste some of their active force overcoming an inherent resistance. This resistance could not result from the series elastic element because it resists lengthening not shortening. Hill thought of this resistance as another kind of passive active force and found that the faster a muscle shortens, the less total force it produces. Assuming a constant active force, Hill concluded that the faster shortening leads to larger resistive force (Shadmehr and Wise, 2008). In 1938 Hill performed a series of experiments on frog muscles hanging a weight from their muscle and stimulating them via electrical current. By recording the weight, the muscle's contractile velocity, length and heat production during the contraction he was able to determine the force –length-velocity properties of the muscle. These experiments have been performed on single fibres within a sarcomere and then been scaled up to the sarcomere itself and consequently to the whole muscle. In order to capture these force-length-velocity properties of whole larger muscle, Hill created a mechanical muscle model. Hill muscle model is composed of three elements only. These elements are series element (SE), contractile element (CE) and parallel element (PE). The first two elements are arranged in series and are in parallel to the last element. The serial element accounts for the muscle elasticity during isometric force conditions and connects serially to the contractile element which is freely extendable when at rest but capable of shortening when activated by electrical stimulation. These two elements are joined in parallel to the other elastic element which accounts for the elasticity of the muscle at rest. The parallel element accounts for the inter-muscular connective tissues surrounding the muscle fibres while the series elastic element accounts for the elasticity of cross-bridges within the muscle (Vignes, 2004). The Hill model with its series and

parallel elements is shown in Figure 4.6. Hill muscle model is a purely mechanical muscle model built from the systems engineering perspective and still in demand due to its simplicity and effectiveness.

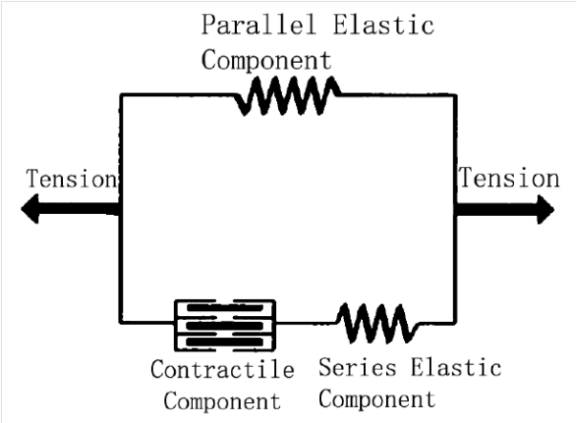


Figure 4.6: Hill's mechanical muscle model comprising of 3 elements

Another type of muscle model was formulated once the sliding filament theory was proposed by Huxley (Huxley and Simmons, 1971). It was built using a reductionist approach that takes into account the actual molecular structure of the muscle and attempts to predict the developed tension by simulating the forces produced by the cross-bridge attachments between the actin and myosin molecules. The cross-bridge-generated sliding-filament model was not universally accepted until the invention of electron microscope enabled direct observation. Huxley initiated publishing his findings on muscle modelling in 1957 and his muscle model is still used by experimenters. Huxley (1957) pictured the myosin heads, attached to the parent myosin filament by elastic tails, by utilizing a cartoon replicated in Figure 4.7. When the muscle is stimulated, those heads which are in the vicinity of an attachment site on the actin filament can be expected to attach to that site. Force will then would be applied to the actin filament by the stretched elastic tail of the cross-bridge; a contractile force being created if the elastic tail is in a state of extension. Since a contraction velocity would tend to shorten the elastic tail, in order that force be created one must suppose that the cross-bridge tail is already extended when the cross-bridge attaches.

Huxley proposed that this extension could be provided by thermal agitation of the cross-bridge head. Since this agitation should be as likely to contract as to extend the tails of individual cross-bridges, Huxley further suggested that attachment would, by an unspecified chemo-mechanical mechanism, be facilitated for the cross-bridges which are displaced positively and made difficult for those whose tails are in a rest position or contracted (Williams, 2011).

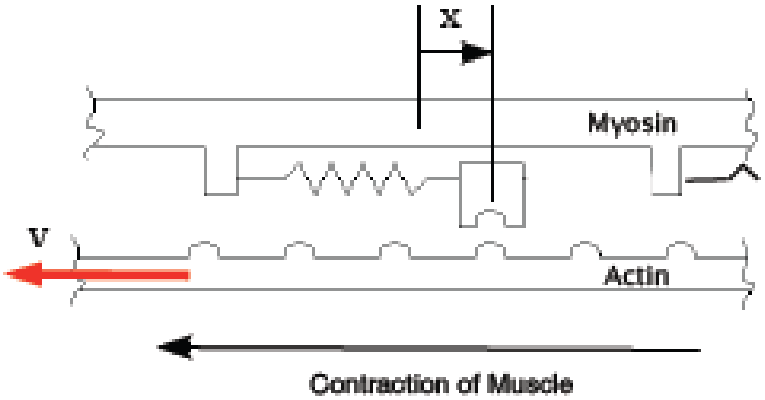


Figure 4.7: Huxley's basic cartoon, X: displacement (Williams, 2011)

Since the introduction of the Hill model, various modifications have beenmade to more accurately incorporate further complexities and increase the model's accuracy. One of the most notable was made by Zajac et al. (1986). They extended the Hill model to include the tendon connection and pennation angles for muscle fibre. As shown by the muscle schematics in Fig. 4.8, the pennation angle is an angle made between the muscle and tendon at the point where they connect (Vignes, 2004). Based on these modifications, more important physiological properties of muscle-tendon complexes are created. In Fig. 4.8, α is pennation angle, l_{se} is length of serial element, l_{ce} is length of contractile element (CE), l_T is length of tendon, l_M is length of muscle, and l_{MT} is length of musculotendon system. K_{se} is series elements stiffness, K_{pe} is parallel elements stiffness, K_M is muscle stiffness and K_T is series tendon stiffness.

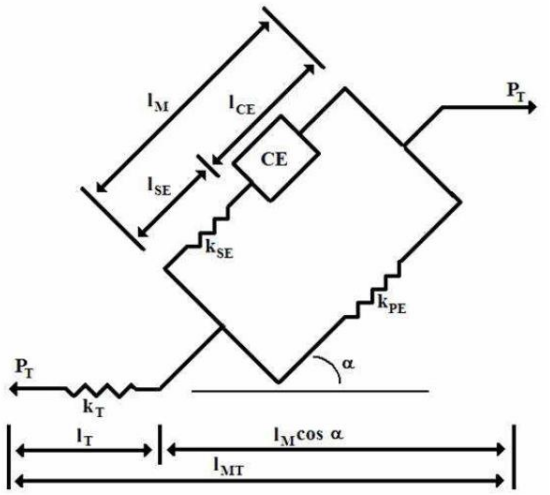


Figure 4.8: Zajac mechanical muscle model, including tendon stiffness and pennation angle

During the past half a century there have been numerous attempts to model muscle force mathematically, ranging from the simplest to the most comprehensive ones that consider many physiological and mechanical factors of the muscle such as muscle length, shortening velocity, neural activation, and muscle architecture apart from the very famous one mentioned earlier. These attempts include Coggshall and Bekey, (1970), Pell and Stanfield (1972), Christakos and Lal, (1980), Woittiez et al. (1984), Hannaford (1990), Schultz et al., (1991), Fuglevand et al. (1993), Wexler et al. (1997), Bobet and Stein (1998), Herbert and Gandevia (1999) and Studer et al. (1999).

When a muscle contraction is sustained, muscle becomes fatigued, and force production is affected by underlying fatigue and recovery effects in the neuromuscular system (Merton, 1954; Bigland-Ritchie, 1981; Enoka and Stuart, 1992; McComas et al., 1995). However, previous force models did not generally consider fatigue and recovery effects; therefore, they cannot be used to describe the time course of force production for an extended period of time, during which fatigue and perhaps recovery effects become more apparent (Liu et al., 2002). Hawkins and Hull (1992, 1993) recognized the importance of fatigue effect during tasks lasting long periods of time. They considered

the fatigue effect in their prediction of muscle force production by incorporating several empirical fatigue indices such as fibre endurance times and fatigue rates into a muscle fibre-based model that calculated muscle force as the sum of individual fibre forces. Rather than deriving the force–time function based on a consistent simple biophysical principle, they established the force–time dependence based on empirical data. Because these empirical quantities need to be determined from other experiments and the accuracy is difficult to achieve, this model could not give satisfactory prediction of force. A group of investigators developed a model to predict force output as a function of time in paralyzed quadriceps muscle under interrupted FES based on electromyogram data and muscle metabolic history (Giat et al., 1993, 1996; Levin and Mizrahi, 1999). This model relies heavily on accurately measuring temporal changes in muscle metabolites, i.e., the inorganic phosphorus (Pi or H2PO4-) measured by in vivo 31P magnetic resonance spectroscopy, intracellular pH, and other data obtained from various sources and literature. Riener and colleagues developed their model based on a motor unit recruitment function and considered muscle fatigue and recovery effects by introducing a muscle fitness function (Riener et al., 1996; Riener and Quintern, 1997). Both of these empirical functions need to be predetermined.

A common feature of these models is that many physiological and biomechanical parameters need to be determined. For example, in Riener's model, there are more than 28 parameters, and in Giat's model, more than 30. The complicated formulae in these models have obscured the biophysical principles of muscle force generation and hindered their more general applications.

4.5 Development of a joint model

As explained earlier due to the reverse recruitment order, under electrical stimulation the paralyzed muscles tend to fatigue faster and the force decline is steeper compared to a normal sequence which leads to the movement of healthy muscles. Modelling of muscle behaviour under electrical stimulation considering fatigue and recovery has gained a lot of momentum. This is due to the fact that to incorporate an accurate closed loop control system, a precise dynamic model of the muscle or the joint is needed to recalculate the best trajectory to complete a desired task. This will in return help the controller in adjusting the controlled parameters in order to finish the task by reduced fatigue. Therefore development of a joint model to predict the behaviour of the system and use it

as a feedback is an important step prior to the implementation of the movement synthesis and associated control strategy.

The human knee joint has been modelled extensively (Chizeck, 1999); (Franken, 1993); (Eom, 2006); (Ibrahim, 2009). Most of the reported models have limited prediction capabilities, since they describe the knee joint under very restricted conditions and contain too many parameters, making them unidentifiable when only the state variables are known. Eom et al. (2006) developed their joint model considering joint recruitment feature, activation dynamics and contraction dynamic, where all the muscle fibre parameters have to be estimated. These values are difficult to measure and require special equipment and experimental procedures. In their paper, they estimate these values to calculate their joint model. Franken et al. (Franken, 1993) introduced a simpler method using least square algorithm in combination with Levenburg-Marquardt algorithm to develop a joint model. However, the joint model developed failed to model the leg joint for movement above 10Hz. Ibrahim et al. (2009) has a similar approach for their joint model. The joint model is developed using fuzzy logic (Ibrahim, 2009) by optimizing the model parameters using genetic algorithm to fit the model output to the experimental leg motion. The joint model developed is not generalized and is subject based.

Most of the natural phenomena are highly nonlinear processes. There are two approaches to model dynamics of these types of systems. The first is using physical laws in order to design a model for the system where the second solution utilizes system identification techniques. However, the first method is not as easy to implement due to three main reasons (Zito, 2005). Setting the right values for physical parameters in order to have specific model, difficulties in identification of physical parameters from data samples and complexity of this model are the main reasons of avoiding physical laws and to use strategies such as system identification in order to model the system.

For the second approach, a number of methods have been developed, some of which assume that the whole system is like a black box. In these methods, there is no need to have a lot of information about the physics of a system. In fact, these methods utilize nonlinear mapping between the inputs and output (Sjöberg, 1995). For the purpose of this chapter, artificial neural network (ANN), and NARMAX_OLS are used as two different strategies in order to model the knee joint.

4.5.1 Experimental setup

A number of experiments were performed on 3 paraplegic individuals with an average height of 1.70 metre and weight of 76 Kg after signing a consent form. The subjects had injuries at T2/T3, T4/T5 and L4 level and suffered from the lesion for an average of 17 years.

Throughout the experiments, paraplegic subjects were placed in a semi-upright sitting position (45º) with the thigh hanging using thigh support on a frame to avoid any constraint on the leg movement. Velcro straps were used to stabilise the subject's upper trunk, waist and thigh. The knee angle of the leg movement was recorded via a goniometer (Biometrics Ltd, UK), Figure 4.9 shows the position of goniometer. The position of the leg is recorded instantaneously using Matlab software through analogue to digital converter (ADC) card and serial connection. Angle of 90 degree between the shank and foot was considered as zero degree. Displacements in the dorsiflexion direction were considered as positive and those in the plantar flexion direction as negative.

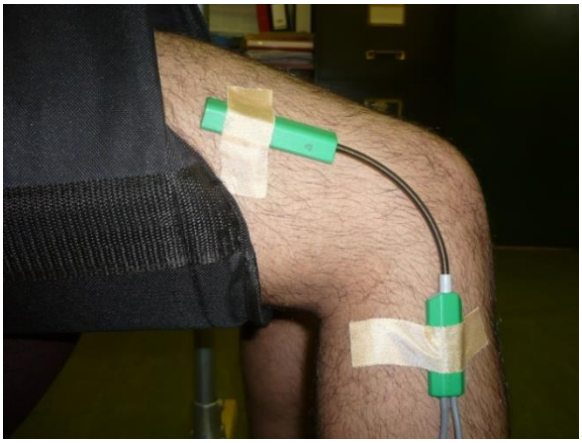


Figure 4.9: Experiment set up

Electrical stimulation was delivered via two MultiStick™ gel surface electrodes (Pals platinum, Axelgaard Mfg. Comp, USA, 50mm x 90mm) to the dominant leg of the subjects. The cathode was positioned over the upper thigh, covering the motor point of rectus femoris and vastuslateralis. The anode was placed over the lower aspect of thigh,

just above patella. Prior to each test, the electrodes were 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 through the entire trials. A RehaStim Pro 8 channels (Hasomed GmbH, Germany) stimulator received stimulation pulses generated in Matlab software through USB connection for application to the muscle. One of the drawbacks of recording swinging leg by tracking goniometer angle is drifting from set value after some time due to the leg movement. More than 3600 simulation pulses with simulation frequency at 10 Hz and pulse widths varying from 200µsec to 350sec were used to develop the joint model. The experimental data was sampled every 0.05s using Hasomede device. To avoid muscle spasm, the electrical input current was limited to 40 mA for maximum period of 1s. The experimental data used in these studies were not filtered or prepossessed.

The input to the muscle stimulator is presented in Figure 4.10, which shows the electrical current, applied to the subjects' quadriceps muscles. The higher amplitude means the higher current stimulation level is resulting in broader shank extension.

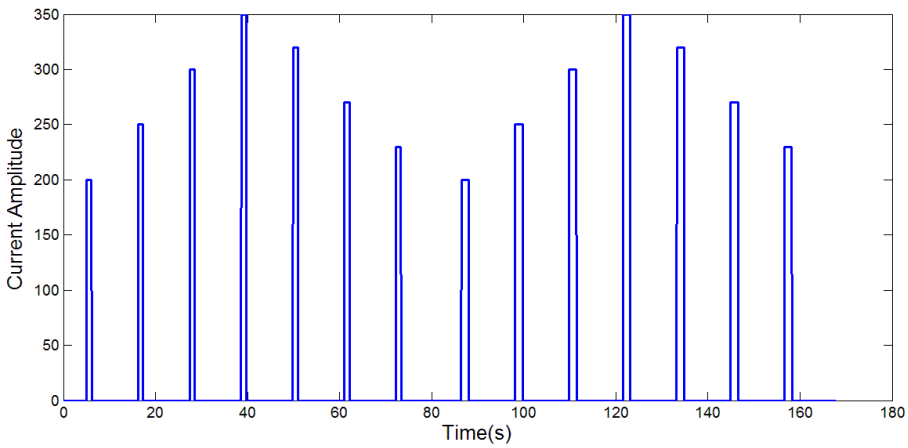


Figure 4.10: Current amplitude during the time of stimulation

4.5.2 Joint model development using neural network

The artificial neural network (ANN) is a mathematical model or computational model that simulates the structure and functional aspects of biological neural networks. It consists of an interconnected group of artificial neurons that are connected to each other in layers. ANN is able to learn, adapt to the environment and acquire knowledge through some learning patterns. ANN has been applied to real-world problems in considerably complex systems and many disciplines (Luis, 1994).

Few methods of training the ANN are available including the most widely used: back propagation. Back propagation neural network, also in general referred to as multi-layer perceptron neural network, consists of an input layer (input nodes), hidden layer(s) (hidden nodes) and output layer (output nodes). In order to train a neural network to perform some task, the weight of each connection between neurons is adjusted in such a way that the error between the desired output and the actual output is reduced. This process requires the neural network to compute the derivative of error with respect to the weights. In other words, it must calculate how the error changes as each weight is increased or decreased slightly. Several different types of neural networks were trained in order to find the optimum model. Finally, a two layer MLP network including four neurons in each layer with five inputs was selected from the trained neural networks. To model the knee dynamics *u(t), u(t-1), u(t-2), y(t-1)* and *y(t-2)* were used as input to the neural network. The network was trained using back propagation method. .

4.5.3 Joint model development using polynomial NARMAX model estimation

The nonlinear autoregressive moving average with exogenous input (NARMAX) is a commonly used structure for modelling nonlinear dynamics and is a good example of the black box modelling approach. The orthogonal least squares (OLS) algorithm (Leontaritis, 1985) is utilised to estimate parameters of the model. In this approach the output can be considered as a nonlinear function f(.) of different terms, depending on inputs, output and error. The greatest challenge in writing this function is to find the most important terms that should be included. OLS algorithm searches for the most important terms; which ones in the f(.) are the most significant terms on an orthogonal basis (Boynton, 2011).

The NARMAX-OLS algorithm includes three different procedures: parameter estimation, structure selection and model validation. In fact, finding the most significant terms, which affect the output, is a part of structure selection procedure, which is the most important and complicated part of the NARMAX-OLS algorithm. It depends on different parameters, such as sampling frequency, prior knowledge of the system, etc. For the purpose of this thesis the error reduction ratio (ERR) method, which is commonly employed and very useful in terms of system identification, is used (Billings, 1989).

To estimate the parameters of the model the nonlinear relationship between output and input should be expressed as in:

$$y(t) = f\begin{pmatrix} y(t-1), \dots, y(t-n_y), u_1(t-1), \dots, \\ u_1(t-n_{u1}), \dots, u_m(t-1), \dots, \\ u_m(t-n_m), e(t-1), \dots e(t-n_e) \end{pmatrix} + e(t) \tag{4.1}$$

where y, u and e are the output, input and error respectively. This is also representing the maximum lag of the output, input and error.

Noise terms included in the model can avoid bias and take into consideration uncertainties and un-modelled dynamics. Billings and Tsang (Billings, 1989) showed that the NARMAX algorithm is able to present a satisfactory model for every system around an equilibrium point, if the following conditions are met: The system should have a finite and realizable response and a linear model should exist around the system's equilibrium point.

Long calculations to reach the final estimated model results in

$$\hat{y}(t) = \emptyset_{yu}^{T}(t-1)\hat{\Theta}_{yu} + \phi_{yu\xi}^{T}\hat{\Theta}_{yu\xi} + \phi_{\xi}^{T}(t)\hat{\Theta}_{\xi} + \xi(t) \tag{4.2}$$

where, ϕ_{yu}^{T} is a matrix that includes all linear and nonlinear combinations of input and output terms up to and including time (*t-1*) with the maximum degree of *l*. $\hat{\Theta}_{yu}$

represents the parameters corresponding to these terms. In addition, $\xi(t)$ stands for residuals which can be defined as

$$\xi(t) = y(t) - \hat{y}(t) \tag{4.3}$$

Now to clarify equation (4.2), it is written in the format.

$$\hat{y}(t) = \phi^T(t-1)\hat{\theta} + \xi(t) \tag{4.4}$$

To solve this equation, $\hat{\theta}$ value has to be estimated, meaning that a cost function must be defined. It is clear that cost function should depend on $\xi(t)$. A good choice for cost function is the following.

$$c(\hat{\theta}) = \|y(t) - \emptyset^T(t-1)\hat{\theta}\| \tag{4.5}$$

In equation 4.5 $\| \ \|$ denotes the Euclidean norm. Chen et al. (1989) suggested three possible approaches to solving this problem. The most common method is the orthogonal estimation algorithm. Complexity of a large scale m-dimensional system can be reduced by transforming the m-dimensional space into m different one-dimensional spaces. This idea can be applied into estimation problems. It makes processing easier and results in an independent computation of each coefficient (Wei, 2005). *y(t)* can be expressed as

$$y(t) = \sum_{i=0}^{n_\theta} \theta_i p_i(t) + \xi(t) \tag{4. 6}$$

Where n_θ, P_i,, θ_i represent the number of coefficients, regressors and coefficients respectively. Equation (4.6) can be rewritten in other forms, including orthogonal terms. Therefore, a transformation capable of mapping equation (4.6) into the orthogonal form is required. The general form of transformed equation should appear as in

$$y(t) = \sum_{i=0}^{n_\theta} g_i \omega_i(t) + \xi(t) \tag{4.7}$$

where $\omega_i(t)$ is orthogonal regressor and g_i is coefficient. All the $\omega_i(t)$ values should satisfy orthogonality over the allotted time. Therefore the following condition should be met

$$\frac{1}{N}\sum_{t-1}^{N}\omega_i(t)\omega_{j+1}(t)=0, \qquad i=1,2,....j \tag{4. 8}$$

where N represents the number of data points. To find ω_m, the following relationships are used.

$$\omega_m(t)=p_m(t)-\sum_{i=0}^{m-1}a_{im}\omega_i(t)$$

$$\omega_0(t)=p_0(t)=1 \qquad m=1,2,....,n_\theta \tag{4. 9}$$

where a_{im} is defined as

$$a_{im}=\frac{\sum_{t=1}^{N}p_m(t)\omega_i(t)}{\sum_{t=1}^{N}\omega_i(t)^2}, \qquad 0\le i\le m-1 \tag{4.10}$$

In addition, to find the coefficients in the equation 4.7, the following relationships are defined by Korenberg (1988) and Billings (1989).

$$\hat{g}_i=\frac{1}{N}\sum_{t=1}^{N}\frac{y(t)\omega_i(t)}{\omega_i^2(t)}, \qquad i=1,2,\dots,n_\theta \tag{4.11}$$

$$\omega_i^2(t)\neq 0$$

The following can transform back the coefficients to the original form:

$$\hat{\theta}_i=\sum_{i=m}^{n_\theta}\hat{g}_i v_i\,, \quad m=1,2,\dots,n_\theta$$

$$v_m=1 \tag{4.12}$$

$$v_i=-\sum_{r=m}^{i-1}a_{ri}v_r\,, \quad i=m+1,\dots,n_\theta$$

$$a_{ij}=\frac{1}{N}\sum_{t=1}^{N}\frac{\omega_i(t)p_j(t)}{\omega_i^2(t)} \quad i=1,2,\dots,j-1 \text{ and } j=2,\dots,n_\theta$$

Chen et al. (1989) studied four different methods for orthogonal decompositions and suggested the modified Gram-Schmidt model. This model was preferred over the others available because of its greater precision.

One of the drawbacks of the polynomial NARMAX algorithm is the large number of regressors, which are produced after the first step. Even when all the regressors for the small values of n_y, n_u, n_e and l are calculated, there are plenty of terms that can be used. For example, if there are just 14 terms (regressors), then 16384 (2^{14}) different models can be produced. It is not possible to check all these models to find the most appropriate one. One does not know how many terms are necessary to model the principle dynamics of the system. This is similar to the problem that occurs when neural networks are used to model a system. Usually, there is a tradeoff between generalization and over-fitting. The network should be able to predict the behavior of the system for an unseen input, and it should also model the small dynamics of the system. In order to model the small dynamics, as many terms as possible should be included. However, this leads to a problem with over-fitting. In addition, this may produce some dynamic regimes, which do not exist in the original system. This network would also not have a good performance when using unseen data. The same problem occurs when using the polynomial NARMAX algorithm (Aguirre, 1995). It is rare that all the terms are necessary to form a nonlinear model. Usually, no more than 10 terms are required for creating a good model. Even after finding out how many terms are necessary, the question remains as to which terms should be selected. The terms with the most significant effects should be selected and the others eliminated. Several methods have been suggested for solving this problem (Aguirre, 1994). They can be grouped into two major categories: Constructive techniques and Elimination techniques. Both groups have the same goal but use different approaches. Constructive techniques start by gradually making and improving the model while elimination techniques try to eliminate less important terms from the initial all-encompassing model. Most of these techniques use statistical analysis to decide which terms should be added or eliminated.

Draper and Smith (1988) and Billings and Voon (1985) suggested a stepwise method, which became a basis for future research. Korenberg et al. (1988) and Billings et al. (1989) introduced a method that became very popular in subsequent years, called the error reduction ratio (ERR) method. This method has considerable advantages and is

able to facilitate analysis without the need for all the terms to be included, and is compatible with the orthogonal estimation algorithm. The ERR test can be explained briefly as below:

Multiplying equation 4.7 by itself and calculating the average of the whole equation gives:

$$\frac{1}{N}\sum_{t=1}^{N} y^2(t) = \frac{1}{N}\sum_{t=1}^{N}\{\sum_{i=0}^{n_\theta} g_i^2\omega_i^2(t)\} + \sum_{t=1}^{N}\xi^2(t) \tag{4.13}$$

Assuming that $\xi(t)$ is the zero mean average, the orthogonality of $\omega_i(t)$ leads to

$$\frac{\frac{1}{N}\sum_{t=1}^{N}\{\sum_{i=0}^{n_\theta} g_i^2\omega_i^2(t)\}}{\frac{1}{N}\sum_{t=1}^{N} y^2(t)} \le 1 \tag{4.14}$$

which can be rewritten as

$$\sum_{i=1}^{n_\theta}\frac{\frac{1}{N}\sum_{t=1}^{N} g_i^2\omega_i^2(t)}{\frac{1}{N}\sum_{t=1}^{N} y^2(t)} \le 1 \tag{4.15}$$

If $ERR_i = \dfrac{\frac{1}{N}\sum_{t=1}^{N} g_i^2\omega_i^2(t)}{\frac{1}{N}\sum_{t=1}^{N} y^2(t)}$ then the equation (4.15) can be rewritten as

$$\sum_{i=1}^{n_\theta} ERR_i \le 1 \tag{4.16}$$

In fact, when no terms are put into the model, this results in the maximum squared error and its value is $\frac{1}{N}\sum_{t=1}^{N} y^2(t)$.

When one term is added to the model, it reduces the total error by $\frac{1}{N}\sum_{t=1}^{N} g_i^2 \omega_i^2(t)$. Therefore, to find the significance of each term in comparison to the other terms, it is necessary to find its error reduction value as a fraction of the total squared error.

To overcome this limitation and find a simpler model that can predict the output properly, NARMAXOLS-ERR algorithm was used. The following model is suggested by NARMAX-OLS method:

$$y(t) = K_1 y(t-1) + K_2 y(t-2) + K_3 u(t-2)^2$$

$$+K_4 y(t-1)^2 + K_5 y(t-1) y(t-2) \qquad (4.17)$$

The terms and coefficients are summarized in Table 4.2.

Table 4.2: Summarisation of coefficients used

$y(t-1)$	k1= 1.78
$y(t-2)$	k2= 8.57e-01
$u(t-2)^2$	k3 = 2.947665e-06
$y(t-1)^2$	k4= 3.222674e-02
$y(t-1)y(t-2)$	k5= 2.974369e-02

The main advantage of using NARMAXOLS-ERR method is that it leads to a low prediction error. Figure 4.11 shows the prediction output of the nonlinear model. Similar to neural network model, *y(t-1)* and *y(t-2)* are generated online by delaying the predicted output from the nonlinear model. It is obvious that the system has very good prediction performance, although there was accumulation error. A delay is seen between the real-time data and the NAMAX prediction after 50 seconds. As it has been show in Figure 4.11, the NARMAX model has a very good robustness to the system noise. The experimental data that was used for NARMAX-OLS algorithm was not filtered or prepossessed. From the equation 4.17, it is obvious that the NARMAX-OLS model has just few terms, which makes the implementation of the model remarkably easy in experimental studies.

The NARMAX-OLS-ERR method shown a solid performance in following of the experimental data however as the experimental data was not taken long enough for

the effect of fatigue to kick in therefore the model was unable to predict the fatigue induced by electrical stimulation. Figure 4.11 shows the knee angle recorded by the Goniometer for an actual subject. This figure also compares the result with NN representation of the system previously built by Bijanzadeh et al. (2010) within group

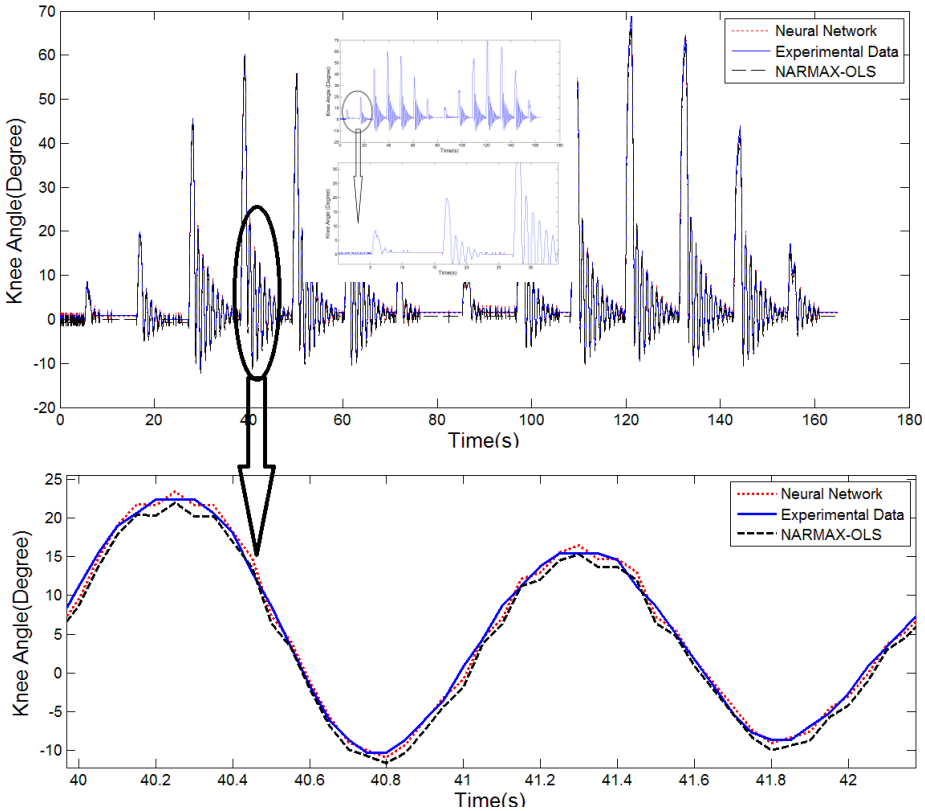


Figure 4.11: The recording of shank's angular movement in time using a Goniometer

4.6 Adaptive neuro fuzzy inference system modelling of muscle fatigue

Adaptive neuro fuzzy inference system (ANFIS), previously known as adaptive network fuzzy inference system (Jang, 1993), is a fuzzy inference system implemented in the framework of adaptive networks. Jang (1997) argues that ANFIS is designed to allow If-Then rules and membership functions (fuzzy logic) to be constructed based on the historical data and also includes the adaptive nature for automatic tuning of the membership functions. This is the integration of the best of fuzzy logic and neural networks and predicts the input-output relationship of a given set of data (Bonissone, 2002).

ANFIS method constructs fuzzy inference system (FIS) in which the parameters of membership function are tuned using a back propagation gradient descent and a least-squares type of method which allows fuzzy systems to learn from the data they are modelling. Generally ANFIS applies two techniques in updating parameters: gradient descent and least square method. Since this approach employs gradient descent method as well as the least-squares method it is named hybrid learning method. The gradient descent term is used to fine tune the premise parameters that define membership functions while the least square method is employed to identify consequent parameters that define the coefficients of each output equations (Jang, 1993); Desale and Jahagirdar, 2001).

The ANFIS modelling process is performed by feeding in a set of collected data (input–output data pairs) and dividing it into training and checking data sets. The training data set is used to find the initial premise parameters for the membership functions by equally spacing each of the membership functions while the checking data set is used after the training to compare the model with actual system. A threshold value for the error between the actual and desired output is determined. If the model does not represent the system a lower threshold value is then used. In general in the training phase the consequent parameters are found using the least-squares method. The error for each data pair is then found and by checking if this error is larger than the threshold value, the premise parameters are updated using the gradient decent method. This process is terminated when the error lies below the threshold value and the checking data set course starts (Jang, 1993, Desale and Jahagirdar, 2001).

According to Bonissone (2002) ANFIS is one of the best tradeoffs between neural and fuzzy systems, providing smoothness due to the FL interpolation and adaptability due to the NN backpropagation and is proven to have faster convergence comparing to typical feedforward NN. However strong computational complexities, restricts the use of ANFIS in some applications. The computational complexity is not proven to be an issue when it comes to modelling the muscle fatigue and the fast convergence together with the smoothness of the predictive model has proved the ANFIS modelling of the muscle to be highly respected.

Jailani (2010) developed an ANFIS muscle model which replicates the muscles behaviour under electrical stimulation reasonably well. However, her model lacks generality as well it is developed using isometric force measurement not by isokinetic torque measurement. This is especially of importance as most of the FES activities are

done with muscles extending and flexing throughout the activity and the model representing is required to present the muscle under this circumstances.

4.6.1 ANFIS structure

In the area of biomedical engineering, medicine and rehabilitation engineering employing fuzzy approach is becoming a routine mainly because of the fact that the real world is too complicated and the quantitative approaches are dissatisfactory. Therefore by making use of fuzziness concept the knowledge can be formulated in a systematic manner to give it an engineering flavour (Mahfouf, 2010). However, it has already been seen that fuzzy systems present particular problems to a developer especially with the rules and membership functions. The issue with the rules is that if-then rules have to be determined somehow. This is usually done by 'knowledge acquisition' from an expert and is normally a time consuming process that is fraught with problems. A fuzzy set is fully determined by its membership function which has to be determined as well as its parameters (John, 2011).

Several fuzzy inference systems have been described by different researchers (Zadeh, 1965; Mamdani, 1974; Takagi and Sugeno, 1985 Sugeno, and Kang, 1988; Sugeno and Tanaka, 1991). The most commonly-used systems are the Mamdani-type and Takagi–Sugeno type, also known as Takagi–Sugeno–Kang type. In the case of a Mamdani-type fuzzy inference system, both premise (if) and consequent (then) parts of a fuzzy if-then rule are fuzzy propositions. In the case of a Takagi–Sugeno-type fuzzy inference system where the premise part of a fuzzy rule is a fuzzy proposition, the consequent part is a mathematical function, usually a zero- or first-degree polynomial function (Mamdani, 1974; Takagi, and Sugeno, 1985).

The ANFIS approach learns the rules and membership functions from data. As it is an adaptive network it is constructed of nodes and directional links. Associated with the network is a learning rule. It's called adaptive because some, or all, of the nodes have parameters which affect the output of the node. These networks are learning a relationship between inputs and outputs. The ANFIS architecture for a two rule Sugeno system is shown in Figure 4.12. In this structure the circular nodes represent nodes that are fixed while the square nodes are nodes that have parameters to be learnt.

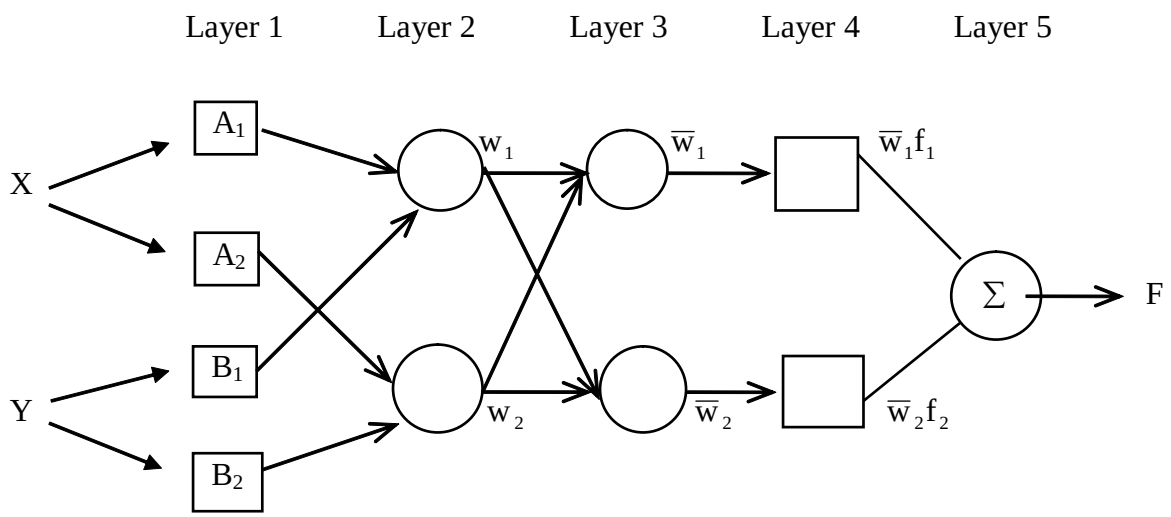


Figure 4.12: An ANFIS architecture for a two rule Sugeno system

A two rule Sugeno ANFIS has rules of the following form:

$$\begin{aligned} &If\ x\ is\ A_1\ and\ y\ is\ B_1 \quad THEN\ f_1 = p_1x + q_1y + r_1 \\ &If\ x\ is\ A_2\ and\ y\ is\ B_2 \quad THEN\ f_2 = p_2x + q_2y + r_2 \end{aligned} \tag{4.18}$$

For the training of the network, there is a forward pass and a backward pass. The forward pass propagates the input vector through the network layer by layer. In the backward pass, the error is sent back through the network in a similar manner to backpropagation. Layer one is also called the membership function layer. The output of each node is represented as below:

$$O_{1,i} = \mu_{A_i}(x) \qquad for\ i = 1,2 \tag{4.19}$$

$$O_{1,i} = \mu_{B_{i-2}}(y) \qquad for\ i = 3,4 \tag{4.20}$$

In the above representation x (or y) is the input to the node and A_i (or B_{i-2}) is the fuzzy set associated with this node. The $O_{1,i}(x)$ is essentially the membership grade for x and y. The membership functions could be anything but for illustration purposes the bell shaped function is used as follow. In this representation a_i, b_i, c_i are the parameters to be learnt. These are the premise parameters. The effect of changing parameters a, b and c is examined in Figure 4.13.

$$\mu_A(x) = \frac{1}{1 + \left| \frac{x - c_i}{a_i} \right|^{2b_i}} \tag{4.21}$$

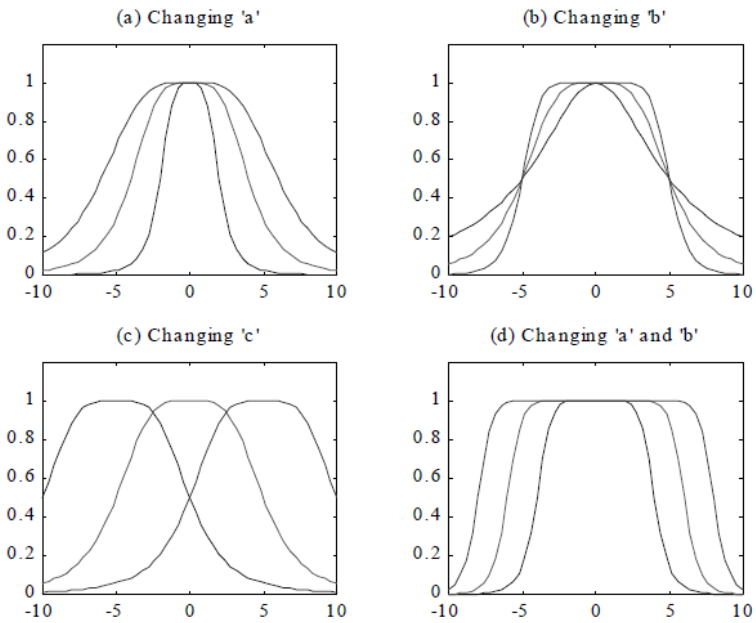


Figure 4.13: The effect of changing parameters a, b and c in layer one (Bonissone, 2002)

Layer two is also known as multiplication layer. Every node here multiplies the inputs of membership degrees and produces the firing strength of the rule or the degree with which the corresponding rule is fired.

$$O_{2,i} = w_i = \mu_{A_i}(x)\mu_{B_i}(y), \quad i = 1,2 \tag{4.22}$$

Layer three also known as normalization layer contains fixed nodes which calculate the ratio of the firing strengths of the rules.

$$O_{3,i} = \overline{w}_i = \frac{w_i}{w_1 + w_2} \quad i = 1,2 \tag{4.23}$$

The nodes in layer four are adaptive and perform the consequent of the rules. This layer applies Sugeno's processing rule.

$$O_{4,i} = \overline{w_i} f_i = \overline{w_i}(p_i x + q_i y + r_i) \tag{4.24}$$

The parameters in this layer (p_i, q_i, r_i) are to be determined and are referred to as the consequent parameters.

Layer five consists of a single node that computes the overall output as the sum of all incoming signals.

$$O_{5,i} = \sum_i \overline{w_i} f_i = \frac{\sum_i w_i f_i}{\sum_i w_i} \tag{4.25}$$

4.6.2 Hybrid learning algorithm

In employing ANFIS another issue which needs to be addressed is how the ANFIS learns the premise and consequent parameters for the membership functions and the rules. There are a number of possible approaches, yet in this thesis the hybrid learning algorithm proposed by Jang, Sun and Mizutani (1997) which uses a combination of Steepest Descent and Least Squares Estimation (LSE) is adopted. A very high level description of how the algorithm operates is briefly discussed here. It can be shown that for the network described if the premise parameters are fixed the output is linear in the consequent parameters. The total parameter set is split into set of total parameters (S), set of premise (nonlinear) parameters (S_1) and set of consequent (linear) parameters (S_2). ANFIS uses a two pass learning algorithm, forward pass and backward pass. In forward pass S_1 is unmodified and S_2 is computed using an LSE algorithm while in backward pass S_2 is unmodified and S_1 is computed using a gradient descent algorithm such as back propagation. This means the hybrid learning algorithm uses a combination of steepest descent and least squares to adapt the parameters in the adaptive network.

The forward pass is performed by presenting the input vector, calculating the node outputs layer by layer, repeating for all data then forming matrices A and Y, identifying parameters in S_2 using least squares and finally computing the error measure for each training pair. In backward pass the steepest descent algorithm is used to update parameters in S_1 (backpropagation) and then for given fixed values of S_1 the parameters in S_2 are found and are guaranteed to be the global optimum point.

4.6.3 Fatigue predictable muscle model

In order to develop a fatigue predictable muscle model which correlates the stimulus properties with the output torque a series of experiments were performed on the same 15 subjects mentioned earlier in Chapter 3. The collected data from 10 of these subjects are used in development of the muscle model while the other 5 subjects are used as a control group. The proposed model does not include the metabolic measurements and deals directly with the input and output measurements.

For the purpose of the experiment the isokinetic output torque of the knee joint in extension and flexion is recorded using a KIN-COM isokinetic dynamometer. The limb to be tested is attached to the dynamometer via a padded cuff, which is attached to housing containing strain gauges. The dynamometer head and the chair can be moved along a metal lever arm to accommodate different limb lengths. The machine is designed to ramp up the resistance if the subject attempts to move faster than the preset speed, producing a force equal in magnitude but opposite in direction, thereby resulting in a constant angular velocity of the limb. This is termed accommodating resistance. The Kin Com tests muscle performance. Basic muscle performance is related to the length tension curve. The point at which there is maximum attachment of cross bridges is the point of maximum torque production and this is represented as a graph. Torque can be measured in concentric mode (when the cross bridges attach) and eccentric mode (when the cross bridges are detached). By isokinetic dynamometry, strength evaluation is not limited to the weakest point in the range, since the resistance is accommodating (Takey et al., 2009). An illustration of the dynamometer used for the purpose of experiment is given in Figure 4.14.

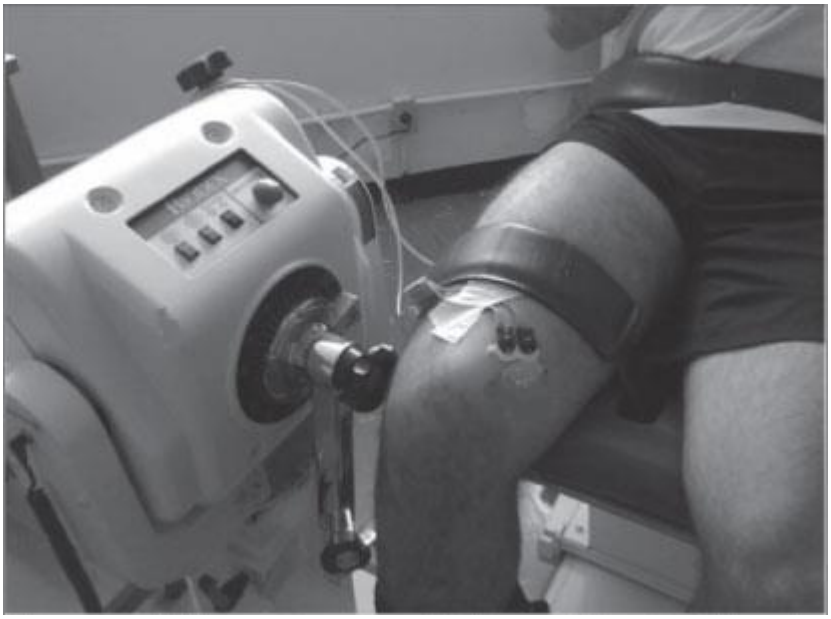


Figure 4.14: The experimental setting of the dynamometer

The steps of the test were explained to each subject to allow them to be oriented and familiar with the testing protocol. Calibration of the unit was performed prior to use according to the manufacturer guidelines. Subjects were tested by the Isokinetic dynamometer at an angular velocity of 60 degree per second for the knee extensor group during isometric contraction. The pads were carefully placed on the frontal part of the shank in a stable and comfortable position. Isokinetic knee strength at 60 degrees per second was measured with the subject's hips at slightly reclined posterior for about 10 to 15 degrees depending on the posture of the subject and knees flexed 90 degrees.

Electrical stimulation was delivered via two MultiStick™ gel surface electrodes (Pals platinum, Axelgaard Mfg. Comp, USA, 50mm x 90mm). The cathode was positioned over the upper thigh, covering the motor point of rectus femoris and vastus lateralis. The anode was placed over the lower aspect of thigh, just above patella. Prior to each test, the electrodes were 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 signals through the entire trials. A RehaStim Pro 8 channels (Hasomed GmbH, Germany) stimulator received stimulation pulses generated in Matlab software through USB connection for application to the muscle.

For the purpose of protecting the muscles from over stimulation and causing possible damage to the muscle and bone structure the muscle was considered fully fatigued if the peak torque was dropped by 50% from the initial peak torque. A combination of between 400 and 600 stimulation pulses for each subject with

stimulation frequencies and pulse widths varying from 10Hz to 50Hz and 200µsec to 600µsec respectively are used to develop the muscle model.

The pattern of stimulation was chosen with frequencies rising by a 5 Hz step starting from 10Hz with the rest of parameters fixed. The experiment was ended after the output torque is reached 50% of its starting value. The same experiment was performed with pulse widths rising by a 50µsec step from 200µsec up to 600µsec. The pulses were delivered with 2 seconds of burst on and 5 second of rest time. The peak torque is recorded for each cycle of each of the experiments resulting in an extensive network of over 2400 data sets. 1700 of these is used as training data while 750 was used as checking data. Another 1000 sets of data obtained from the control group were used in order evaluate the performance of the ANFIS fatigue predictable model.

The network converged after 720 iterations as can be seen from Figure 4.15. After the 420th iteration, there was no further improvement on the training error and it was assumed that the network had already reached the global minimum.

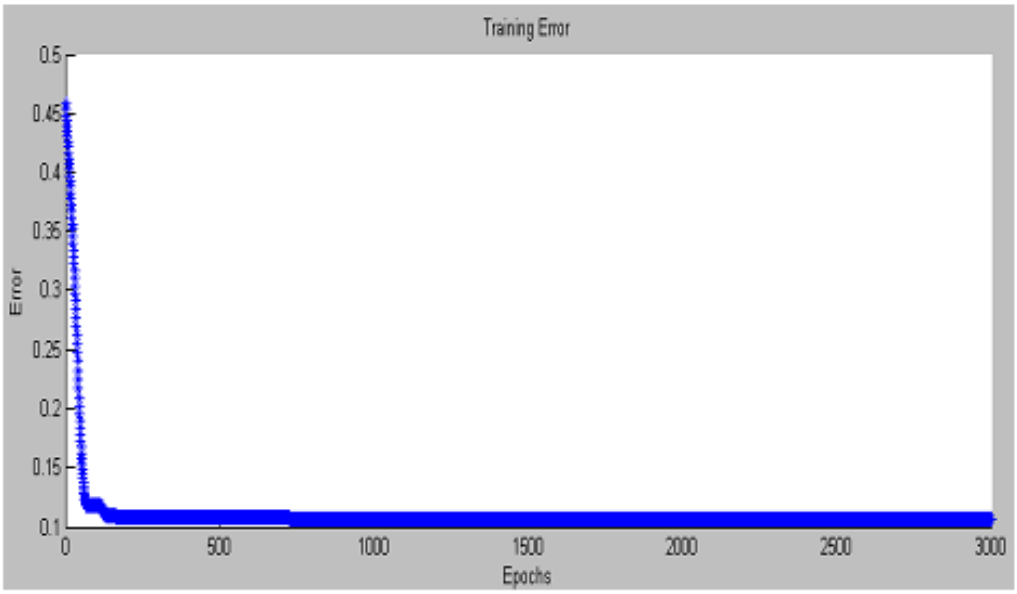


Figure 4.15: ANFIS's convergence curve

As mentioned earlier, 1700 of the data collected from the 10 paraplegic subjects was used as training data while 750 was used as checking data. These are compared in Figure 4.16.

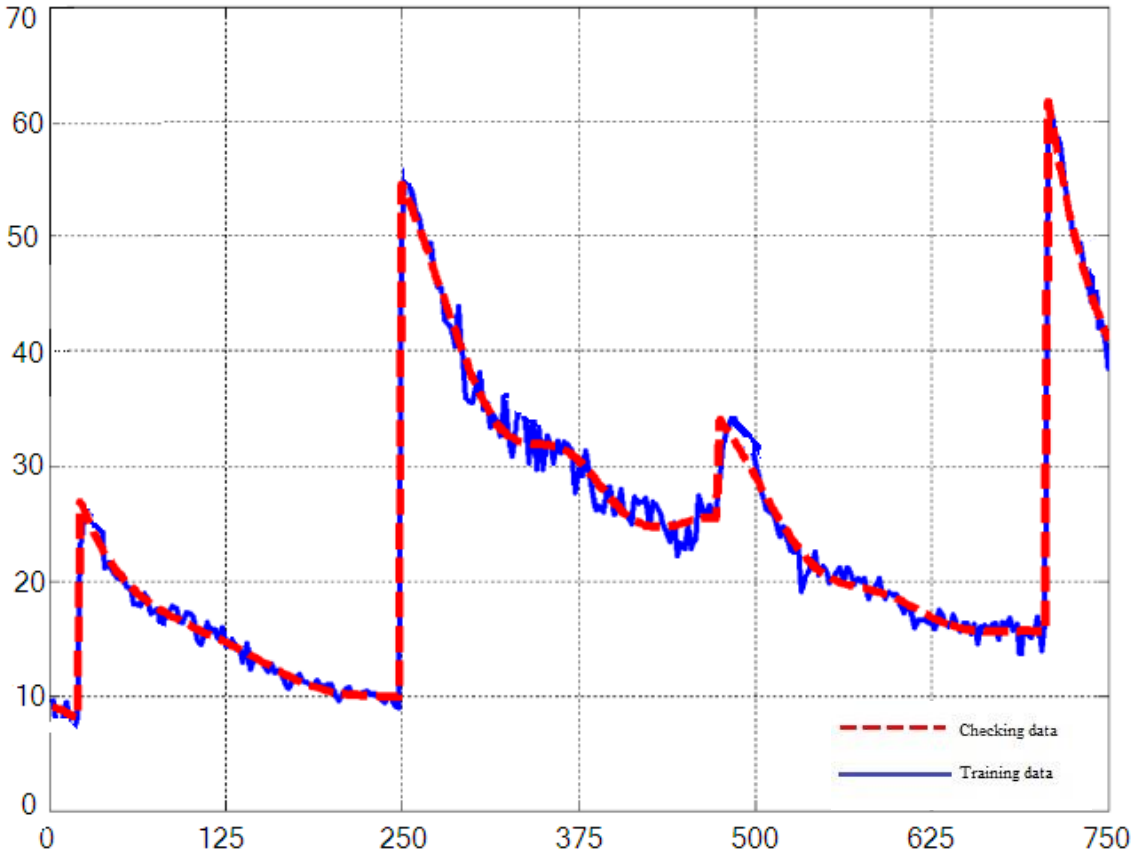


Figure 4.16: Comparison between the training and checking data

As it is shown in Figure 4.16 the ANFIS fatigue predictable muscle model was proved to be able to follow the mean of the output data variation of the control group for a biphasic square stimulus with Frequency of 22 Hz, pulse width of 425 µs and current value of 80 mA exceptionally well. It should be noted that due to the different muscle health statuses of the subjects some have been able to perform the exercise in 6 cycles while some have been able to perform the leg extension only up to 3 cycles under the lowest constant angular velocity. Figure 4.17 presents the second average cycle for all 5 subjects.

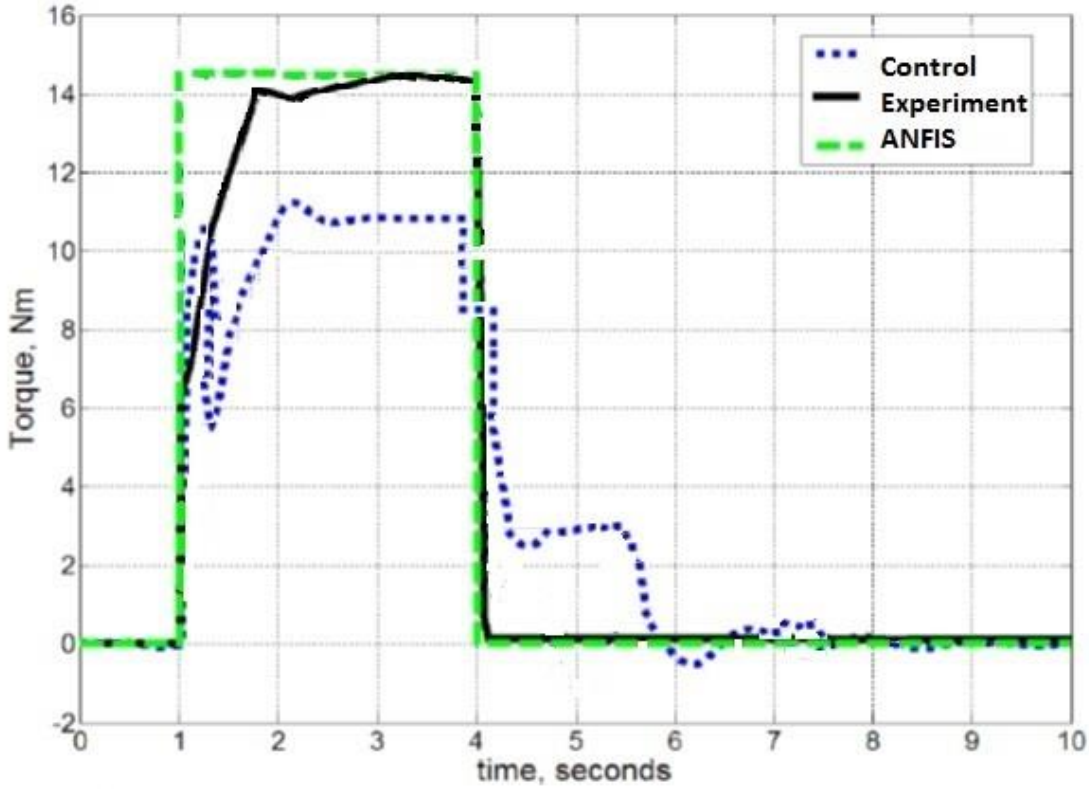


Figure 4.17: The performance of ANFIS fatigue predictable muscle model compared to the control group

It is notable that the absolute error between the ANFIS model prediction and that of the control group was under 3.9% as illustrated in Figure 4.19. This error arises from the high nonlinearity of the paralyzed muscles and the way they individually operate under electrical stimulation and also the tiredness of the subject at the time the data collection was performed. It was originally planned to perform only 2 set of experiments on each subject every day, a morning session and an evening session, but due to the shortage of time this was increased to 4 set of experiments per day which had some minor effect on the quality of the data collected. Moreover, defining muscle fatigue occurring by 50% decrease in torque from original value put a limit on how the muscle reacts to stimulation after this point. However this limit is necessary to prevent any tissue damage since the usual communication channels between the paraplegic individual's limb and brain are disconnected.

4.7 Summary

In order to simulate the effect of electrical stimulation to the muscles and study the output torque, force, angular displacement and angular velocity under a certain type of stimulus the development of a muscle model is essential. Similar to any other type of modelling the model should represent the actual system accurately that the simulation result from implementing it to an actual structure be valid thoroughly. Muscles have been extensively modeled throughout the last 50 years yet no one claims that their muscle model is the ultimate one due to the high nonlinearity of them when they are looked out from the perspective of system engineering. Similar to the muscle models the joints have been subject of the modelling too. When the joint is modelled it is normally looked to as a black box. In this type of modelling the only thing which matters is the input and output and the system specialist is not looking at the microstructure of the system therefore system identification technique become very useful.

In this chapter the microstructure of muscles in general have been looked into and from there the typical way of muscle contraction in healthy subjects and the types of muscle contraction have been studied. Muscle modelling and the way known muscle models such as Hill's, Huxley's, Zajac's, Ferrarin's and Reiner's have been developed have been reviewed generally and the issues surrounding each of them elaborated. The main limitation of the leading muscle models the fact that they either are not able to predict muscle fatigue or that the one able to predict it are way to complicated from the perspective of system engineering. This has resulted in development of an ANN and a NARMAX-OLS-ERR knee joint model in which the whole system is looked into as a blackbox with some specific input and output. Specially the NARMAX-OLS-ERR joint model has been shown to be able to predict the trajectries accuratley and the system proved to be very stable. However, due to the fact that the stimulation did not run long enough for the fatigue element to kick in the model at this stage is not able to predict the fatigue induced in the muscles but yet provides an excellent stable alternative to many other models available in research community. The ability of the muscle or joint model to predict the fatigablity of the muscles is of very high importance. Due to the reverse recruitment order which happens when the paralysed muscles are stimulated the fatigue in paralysed muscles happens faster and is of a different type compared to fatigue in healthy muscles. Therefore, development of a stable fatigue predictable muscle to be

used for the purpose of modeling in FES activities has been performed and resulted in development of a good general ANFIS muscle model.

Chapter 5
Factors Affecting Muscle Fatigue

5.1 Introduction

Muscle fatigue is the failure of the muscle to exert the necessary amount of force in order to perform a specific task. It is one of the major limitations that restrict the general application of FES-based facilities and ads up to the issue of high nonlinearity of the overall musculoskeletal system when modelling it especially for paralysed muscles. The paralysed muscles are different from healthy ones with respect to the fact that fatigue in them is of a peripheral nature and that they lack sensory feedback. In addition to these differences the activation in paralysed muscles is by simultaneous firing of all motor units and the known fact that the easily fatigable fibres are recruited at low threshold. Studies show an increase in the relative proportion of the fast type II fibres in longstanding paralyzed muscles, if compared to normal healthy control subjects (Scelsi, 2001). The longer it takes for the paralysed muscle attended to, the harder to reverse the muscle's fatigability matter.

The physiological fatigue can be categorised as the inability to produce force due to a decline in sensitivity of troponin to calcium or fatigue due to the build up of lactic acid which occurs because of the need of limbs for oxygen. Many of the well established muscle models presented in chapter 4 do not take the issue of fatigue into account or when incorporating the fatigue into their models they limit it to the decline of calcium within the muscle and do not consider the issue of the muscle needing oxygen. This is crucial as most of the FES activities including cycling, rowing, sit to stand and walking are aerobatic with high consumption of oxygen.

In this chapter the issue of fatigue in muscles activated by FES is investigated in detail by comparing the difference between voluntary and electrically stimulated muscle contraction, and studying the factors affecting fatigue in electrically induced contractions including stimulation parameters and stimulation pattern. Furthermore, a series of experiments aimed at understanding the parameters affecting the muscle fatigue in quadriceps muscles are performed and the results is then used in determining the controlled parameters in controllers aiming to control the paralysed limbs.

5.2 Muscle fatigue in voluntary and electrically induced contraction

Although the sensation and voluntary movement of the limbs below the level of injury in spinal cord injured individuals disappears after the injury, the nervous systems as well as the muscles innervated by the nerves under the injury level are still functional and can be excited by means of electrical stimulation. If the electrical stimulus is well coordinated, it will result in a functional movement, hence the name FES. One of the main issues with FES is the decay of muscle force in time as a result of fatigue. In comparison with nondisabled muscles, fatigue happens faster in paralysed muscles (Vignes, 2004). Mizrahi (1997) defines neuromuscular fatigue as inability of an individual muscle or muscle group to sustain the required or expected force regardless of the task to be performed. Muscle fatigue should not be looked into as a negative effect always. The fatigue in healthy muscles is normally a protective mechanism of the body to stop the muscles to be overworked and to enable the recovering process. In the case of FES, premature fatigue is often the problem. In this case muscle force declines before even the muscle reaches 50% of its working capacity.

Muscle fatigue in voluntary contractions is of two origins: central fatigue and peripheral fatigue (Asmussen, 1979). The central fatigue normally happens to react to the feedback it receives from the muscles and the joints. It is the main cause of force loss in large muscle groups such as quadriceps and usually happens to protect the muscles and joints. Peripheral fatigue is normally induced by impairment in cell body excitability, action potential propagation, muscle action potential propagation along the scarcolemma into the T-tubules, synaptic transmission in the neuromuscular junction, ATP-dependent crossbridge cycling, ATP rephosporylation through metabolic processes, and Ca^{2+} release from the scarcoplasmic reticulum (Mizrahi, 1997). It can happen solely on one of the impairments above or could happen as a combination of a few.

In a voluntary contraction the command for muscle contraction is originated from the brain branching out from spinal cord to the nerves which are distributed in the muscle. Therefore the work is shared between different motor units of the target muscle. On the other hand, when the muscle is electrically stimulated using a surface electrode this sequence is interrupted and the whole muscle and therefore most of the motor neurons in the area of excitation to be stimulated which results in fast fatigue and rapid decay of muscle force. Reversed recruitment order is another difference between the voluntary and electrically induced contraction (Rabischong and Guiraud, 1993). In healthy muscles the small motor

units are recruited when tasks requiring low force are needed to be performed and fast fatiguing motor units are recruited when a higher level of force is required. On the contrary when the disabled muscle is stimulated, large axons innervating the fibres that are recruited at low stimulus magnitudes and are easily fatigable are stimulated. This sequence will result in the smaller axons following with increased stimulus levels (Mizrahi, 1997). Electrical stimulation can be used for therapeutic reasons and also for functional purposes.

5.3 Factors affecting fatigue in electrically induced contractions

When it comes to stimulus generated for muscle contraction, fatigue in FES is affected by several factors which can be generally categorised as stimulation parameters and pattern of stimulation. Some other factors including method of stimulation and electrodes, muscle condition and the task to be performed are also crucial. Different research groups have investigated muscle fatigue resulted from stimulation of muscles and have recommended solutions to overcome this problem. The issue of fatigue induced in muscles due to the selection of stimulation parameters is very important as it is normally the case that most of the stimulation parameters are fixed and with one or two controlled parameter the FES activity is performed and the issue of fatigue is prevailed. The activation pattern used during electrical stimulation is also very important and affects force and fatigue. Stimulating the muscles with the right activation pattern that produces the greatest force and generates the least fatigue is therefore of great importance. There are disagreements in the literature when it comes to the stimulation factors and parameters affecting muscle fatigue which are discussed in this section.

5.3.1 Stimulation parameters

When it comes to stimulation parameters in FES there are 3 parameters known which can affect fatigue in muscles. These parameters include intensity, pulse width, and frequency. A good controller design approach is the one that optimises the quantity of each of these parameters in a way that the task is performed as the best to the normal rate and the fatigue induced by stimulation parameter is the lowest. Recruitment curves of electrically stimulated and activated muscle indicate the existence of a stimulation intensity normally referred to as maximal intensity, above which the muscle contracts fully. It is convenient to express the stimulation intensity as a percentage of maximal intensity. It should be noted however that

intensity of stimulation is accepted differently in healthy and in paralyzed subjects depending on the presence or absence of sensory feedback. In healthy population, achieving an electrically induced contraction of comparable magnitude to that of maximum voluntary contraction is seldom possible due to the pain provoked by the high intensity, stimulation required. Comparison of force from maximum voluntary contraction with that from maximal titanic electrical stimulation has shown that in cases where maximal stimulation was possible forces of similar magnitudes were generated under both conditions. However to avoid pain most studies on nondisabled subjects make use of substantially smaller stimulation intensities some time as low as 10% of the maximal voluntary contraction force. With lesser amounts of stimulation fatigue is more limited because only partial recruitment of the motor unit takes place. As the stimulation intensity increases the effect of muscle fatigue is more significant.

Pulse width modulation (PWM) is another way of activating paralysed muscles. In this method the average value of voltage fed to the load is controlled by turning the switch between supply and load on and off very fast. Whenever the switch is on longer in comparison to the off periods, the higher power is supplied to the load. PWM uses a rectangular pulse wave. This pulse width is then modulated resulting in variation in the average value of the waveform. Activation of the muscle using electrical stimulation can be achieved by either PWM with fixed intensity or by modulating the intensity with fixed pulse width.

Stimulation frequency is the other parameter which affects the occurrence of fatigue. Fatigue due to frequency can be divided into low and high frequency fatigue. The fatigue induced in the muscles is called low frequency when the stimulus frequency is between 10 to 20 Hz. It is suggested that the type of fatigue provoked in muscles using this range of frequency has a long lasting depression of the force due to the failure of excitation-contraction coupling mechanism (Mizrahi,1997; Faghri et al., 2009). According to Mizrahi (1997), a fewer crossbridges would be available for force generation due to the fact that in this stage each action potential fails to release regular amount of Ca^{2+} . Recovery from this type induced fatigue is mostly done within minutes but a complete recovery may takes up to hours. High frequency fatigue occurs at the time when the frequency of the stimulus is in the range of 20 to 100 Hz. It happens due to the propagation failure at the axon branch points, neuromuscular transmission failure and propagation failure of the action potential and is accompanied by the accumulation of K^+ in extracellular space (Mizrahi, 1997). In a number of experiments which is conducted for the purpose of this thesis and presented later in this

chapter it is shown that higher frequencies will result in greater force, however the output force is declined in time and it decays faster at higher frequencies.

5.3.2 Pattern of stimulation

Another important factor affecting muscle fatigue when the muscle is electrically stimulated is the pattern of stimulation. The frequently used pattern of stimulation consists of continuous train of pulses with fixed stimulus parameters. When it comes to issue of muscle toning and preparation phases for a FES-base activity the constant train of pulses can be administrated with some rest time however when the purpose of stimulation is of a functional nature another set of stimulation patterns which results in less fatigue should be looked into. Variable firing rate, intermittent stimulation, doublets and N-lets are some of the commonly used patterns of stimulation.

When variable firing rate is used the firing rate is gradually increased, and it is shown in animal testing that it reduces the fatigue (Binder-McLeod and Baker, 1991). While using this method has been proved to reduce fatigue, on the downside it will result in the decline of maximum achievable force. To further improve this pattern it is shown that when instead of a constant stimulation frequency a variable stimulation frequency is used the fatigue is delayed (Mizrahi, 1997). It is known that muscle phenotype can be manipulated via chronic electrical stimulation to enhance fatigue resistance at the expense of contractile power (Duan et al., 1999). This manipulation can be further expanded by a pause during muscle stimulation which in return is called an intermittent stimulation. The intermittent stimulation can be divided into regular and irregular intermittent stimulation patterns. In regular intermittent stimulation duty cycle is of importance as by decreasing the duty cycle fatigue is reduced. In irregular intermittent stimulation the durations of the stimulation and the rest phases can vary without following any set pattern. Although it is known that irregular intermittent stimulation pattern results in less fatigue and due to its nature it is perfect for practical FES application, but the effect of fatigue and recovery are highly dependent on the prearranged stimulation parameters and allocated resting time (Giat et al., 1996). Doublets and N-lets are another type of pattern for stimulation. Stimulation trains using doublets, two closely spaced stimulation pulses, are thought to reduce the rapid fatigue seen in electrically stimulated muscle (Routh and Durfee, 2003). For FES and muscle monitoring applications, there are advantages to doublet stimulation over a range of activation levels that modulate motor unit recruitment; the very famous one is the reduction in muscle fatigue (Freeman and Durfee, 2006). The twitch force resulting from doublet stimulation is shown to be more than

twice the amplitude of twitches produced by normal single stimulation (Mizrahi, 1997). N-lets are the expansion of doublets meaning a set of N closely spaced stimulation pulses as a means of producing contractions with improved fatigue characteristics. In a study by Karu et al. (1995) on 27 able bodied and SCI individuals it is shown that on average, optimal N-let stimulation resulted in 36% increase in isometric torque tracking when compared to traditional singlet stimulation. The results have immediate implications for alleviating the problem of premature fatigue during FES. In this study they conclude that there is considerable subject to subject variation in the results, but specific recommendations can be made for generic "optimal" stimulation pulse train parameters which will minimize fatigue.

5.3.3 Other factors affecting fatigue in muscle stimulation

Looking into the issue of the factors affecting muscle fatigue during electrical stimulation some more general aspects such as the task to be performed, muscle and fibre composition and electrodes and methods arise.

As mentioned earlier a separation has to be made whether the purpose of stimulation is for rehabilitation purposes or functional tasks. Before putting an SCI individual on a FES-task for the first time the muscles should go under extensive training before performing a functional task which might be highly energy demanding. For the purpose of building muscle tone which is also known as endurance training, a set of parameters and stimulation patterns are used while for the purpose of task performing another different set which requires sophisticated method of training is utilised. As another example when an SCI subject is trying to stand using FES the knee extensors are stimulated with high intensity to ensure safety. This will result in high energy consumption and rapid fatigue (Mizrahi 1997).

As explained extensively in chapter 3 human muscles consist of two major fibres known as slow or type I and fast or type Π. Slow fibres produce low forces with little fatigue for short periods of time. Fast fibres produce large forces for short periods of time. Based on the time of the post injury and whether the SCI individual has been receiving FES treatment the proportion slow to fast fibres differs, (Martini, 2007). Low frequency stimulation will transfer fast fibres into slow fibres which in return will result in an increase in the fatigue resistance.

Last but not the least, the choice of electrodes used in stimulating muscles is very important. As they are non-invasive invasion surface electrodes are mostly the product of choice when it comes to the FES. They are easy to apply and employ but the negative side of them is that they literally stimulate any fibre within their reach resulting in some of the fibres

which are no use in performing a task to be extra stimulated and get fatigued. Also for the stimulation signal to travel from the skin to the bone of the muscle and stimulate the relevant fibres and motoneurones, a higher level of stimulation must be used. These issues can greatly get sorted by using implanted electrodes but one must pay the price for invasion in this way.

5.4 Stimulation parameters to be controlled

Electrical stimulation may be delivered through either open or closed-loop control systems. In open-loop control, the patient sends out a command that delivers a certain amount of stimulation, regardless of the actual response of the muscle using visual or auditory feedback. In closed-loop control, electrical stimulation is initiated with the user's command and then modified based on some feedback measurement such as force or position. With closed-loop control, the delivery of electrical stimulation is continuously modulated to control the quantity being measured by the sensors (Sisto, 2008). Using a closed loop system will result in less energy transferred into the muscles inducing less fatigue. However, the pattern of stimulation as well as the parameters to be controlled plays a major role.

The effect of stimulation parameters on fatigue has been investigated extensively. However, the findings have often proved contradicted to each other. This is usually due to the fact that some of the studies have been performed on healthy limbs. Binder-Macleod and Snyder-Mackler (1993) that stimulation frequency and pulse intensity have a supreme impact on muscle fatigue. In 1999 Binder-Macleod and Russ concluded that muscle fatigue was greater at lower frequencies in intermittent stimulation while opposite results have been obtained during continuous stimulation (Kesar and Binder-Macleod, 2006). Furthermore, several studies have suggested different ways to overcome muscle fatigue including using random modulation (Granham et al., 2006; Trasher et al., 2005), N-let pulse trains (Karu et al., 1995) and variable frequency pulse trains (Mourselas and Granat, 1998).

Trasher et al. (2005) and Granham et al. (2006) concluded that random modulation of the stimulation frequency, amplitude and pulse-width did not have any effect on the muscle fatigue rate. However, if this technique gives opposite results, then having this kind of signal is not practically viable technique for muscle fatigue reduction. Moreover, this signal is difficult to control and apply in closed-loop FES control applications. Routh and Durfee (2003) and Bigland-Ritchie et al. (2000) proposed doublet stimulation signal to reduce muscle fatigue during FES. However, Routh and Durfee (2003) wrap up that doublet stimulation signal did worsen fatigue reduction and singlet stimulation signal led to 33 more

cycles than doublet, with the same stimulation parameters. Moreover, doublet stimulation produces more than twice the amplitude of twitches force and tends to make the muscle fatigue quicker. Karu et al. (1995) reported that N-let pulse trains reduce 36% muscle fatigue rate compared to singlet stimulation signal. They used 1 to 6 pulse trains in their study and found the optimum stimulation train as doublet and triplet stimulation signal. This finding is conflicting with findings from Routh and Durfee (2003). This is possibly because of different protocols used in the studies. Yet, Bigland-Ritchie et al. (2000) proposed that using doublet stimulation signal for the first 2 minutes and then keeping on with singlet stimulation will cut muscle fatigue drastically as compared with constant rate trains. Again, this technique is difficult to implement practically in closed-loop FES control applications.

Kesar et al. (2007) studied the effect of stimulation frequency and pulse-width on muscle fatigue and found that stimulation with frequency modulation gives less muscle fatigue rate compared with stimulation with pulse-width modulation. However, stimulation with frequency modulation is impossible with currently available off the shelf programmed stimulators since most of the stimulators allow pulse-width modulation with fixed frequency. Moreover, in their paper, they found that constant stimulation signal gives the least muscle fatigue compared to the others. Previously, Kesar and Binder-Macleod (2006) also concluded that using the lowest stimulation frequency and longest pulse duration could maximize muscle performance if the stimulation frequency and intensity are kept constant. In both papers, they used healthy subjects in their experiments with very high pulse-width. Stimulation pulse-width more than 450μsec and stimulation frequency more than 50Hz are not suitable for paraplegic use (Riener, 1998), therefore these results are only valid for healthy subjects. Earlier, Mourselas and Granat (1998) performed the same experiment with five healthy subjects and one SCI subject with which the results of Kesar et al. (2006, 2008) agree and conclude that stimulation with frequency modulation increases muscle performance, but this effect is very small and almost non existent for many subjects. Therefore, the effect of stimulation with frequency modulation is not significant in improving muscle fatigue rate.

All FES systems have multiple parameters including waveform, amplitude, current duration, frequency, ramping, and duty cycle which need to be adjusted to optimize the response to electrical stimulation. The waveform is a stimulus pulse consisting of a phase (positive or negative), a shape (sine, rectangular, etc.), and amplitude. The effectiveness of FES depends on the type of waveform, which affects both the excitability and the fatigability of the muscle. The amplitude or current intensity is the amount of electrical current that is

delivered which by increasing it more motor units are recruited. The current duration which is also known as pulse width or pulse duration, is the amount of time that the electrical current is delivered to the muscle. The current duration affects the intensity that is required to generate a motor response resulting in comfortableness of the stimulation. The frequency is the rate of individual electrical pulses delivered in trains of pulses at a specified frequency. Ramping is the rate of rise and fall of the current. Rise time allows for sensory accommodation to the stimulation and thus affects the comfort of stimulation. Fall time is less significant since the muscle will stop contracting as the stimulus decreases during fall time. Finally, the duty cycle is the cycle of stimulation, characterized by the time the unit is on versus the time the unit is off. This is also referred to as on or off time. This ratio is important because a muscle needs adequate rest to avoid fatigue. Adjustment of these various stimulation parameters will affect the quality of the muscle contraction (Sisto, 2008, Kralj et al., 1983, Kirshblum, 2004).

In building the stimulation signal some of the mentioned parameters will be fixed while the rest will be adjusted in time by the controller. In order to select the right controlled parameters when dealing with developing the controller section of any FES facility the parameter should be selected in a way that they have the most influence yet induce less fatigue. Since the literature has extensively concentrated on reduction of muscle fatigue and not an examining the effect of choosing controlled parameters on inducing fatigue it is therefore of vital importance that the effect of frequency modulation and pulse width modulation in intermittent stimulation is examined in SCI subjects.

5.4.1 Fatigability of the quadriceps muscles using different control parameters

In order to overcome the fatigue in paralysed muscles that are activated by electrical stimulation, the control parameter or parameters need to increase in time to compensate for the movement. Normally these controlled parameters are chosen because of their great influence on the output torque and their easiness to control in practical applications. It is attempted here to see the effect on fatigue induction on the stimulated muscle by modulating the two very favourable controlled parameters, frequency and pulse width.

Fatigue of FES induced muscle contraction is measured in 3 modes, namely isometric, isotonic and isokinetic measurements. Jailani et al. (2009) measured the fatigue in isometric format arguing that this mode of measurement allows easier control of testing parameters, however it was noted that since most of the FES activities result in a constant movement the result of their study on isometric measurement would not do a complete justice

on the issue of fatigue induced by different controlled parameters. Due to the fact that there is no limit set for identifying the level of irreversible muscle fatigue and considering the fact that in the isometric test the lower end of the shank is fixed and the movement of the shank is not allowed the probability of tissue damage and bone multiple bone fracture will arise highly. Moreover, their study is comprised of testing one paraplegic subject which will make their result substantially subject based. Last but not the least, the range and the step of elevation of the controlled parameter in Jailani (2008) study is not of adequate levels and does not consider the fact that in many cases higher levels of stimulation is required, especially when the subject is left without exercise and proper rehabilitation program for a long period of time.

Based on the above hypothesis, adjustments to the Jailani (2008) experimental procedure is made. The new excitation protocols are designed for greater stimulation levels, as well as for more stimulation levels. Protocol 1 employs 5 different stimulation frequencies varying from 10Hz to 50Hz with step size of 10 Hz and keeping the rest of the parameters fixed (current at 40mA, pulse-width at 300 μsec and pulse duration at 3 sec on and 7 sec off). Protocol 2 employs 5 different stimulation pulse widths varying from 200μs to 400μs with step size of 50 μs and with other parameters fixed (current at 40mA, frequency at 30Hz and pulse duration at 3 sec on and 7 sec off). Intermittent stimulation is used because of the fact that the patterns are more realistic in practical use of FES.

Electrical stimulation was delivered via two MultiStick™ gel surface electrodes (Pals platinum, Axelgaard Mfg. Comp, USA, 50mm x 90mm). A RehaStim Pro 8 channels (Hasomed GmbH, Germany) stimulator received stimulation pulses generated in Matlab software through USB connection for application to the muscle. A dynamometer was used to record the peak torque during the period of stimulation. Since the objective was not to achieve a certain target in angular or torque terms a variety of angular movements in different subjects under different setting of stimulation was observed. The maximum angular extension of 60 degree was allowed.

The performance of the muscle under stimulation was assessed by measuring the decline in peak torque for different stimulation frequencies and pulse-widths. Throughout the experiment on 7 subjects the results showed that fatigue occured generally faster at higher frequencies as the percentage of the decline of the peak force linearly increased with stimulation frequency. Also the peak torque was greater at higher frequencies initially but the decline of the output torque occurred faster. It was noted that in subject 3 and 4 the initial torque was the highest among all and that the percentage of the decline was much less

comparing to the average peak torque decline of all subjects. Subject 3 and 4 were those whose time of injury were shorter than those in experimental group and had active life receiving proper physiotherapy treatment and subject 4 had used FES training before. Table 5.1 summarises the demography of the 7 subjects that took part in this study.

Table 5.1: The demography of the subjects included in this study

	S1	S2	S3	S4	S5	S6	S7
Age	20	55	29	34	43	40	48
Gender	Male	Male	Male	Female	Male	Female	Male
Height	168 cm	168 cm	168 cm	171	158 cm	163 cm	164 cm
Weight	70 Kg	75 Kg	75 Kg	77 Kg	64 Kg	54 Kg	70 Kg
Level of injury	L4	T12	T12	L2	L2	T10	L1
Post injury time	5	32	4	5	17	7	20
Exercise history before the injury	acive(3)	Rare(1)	very good-hicker(5)	very good(4)	Rare(2)	Active workoholic(5)	Rare (1)
Exercise and treatment after injury	basketball-active(3)	physio-rare(2)	physio-active(4)	active house wife-active(4)	physio-rare(2)	Sport-activist(5)	rare(0)
FES experience	No	no	no	Yes	no	no	no
Extras				Very active			

In subjects who have not been receiving appropriate training after injury the decline of the initial force was proved to be higher, this is exceptionally seen in subject 5. Figure 5.1 shows the averaged pick torque for all 7 subjects using different frequencies while Figure 5.2 compares the averaged results at 30 Hz with the results obtained from subjects 4 and 7.

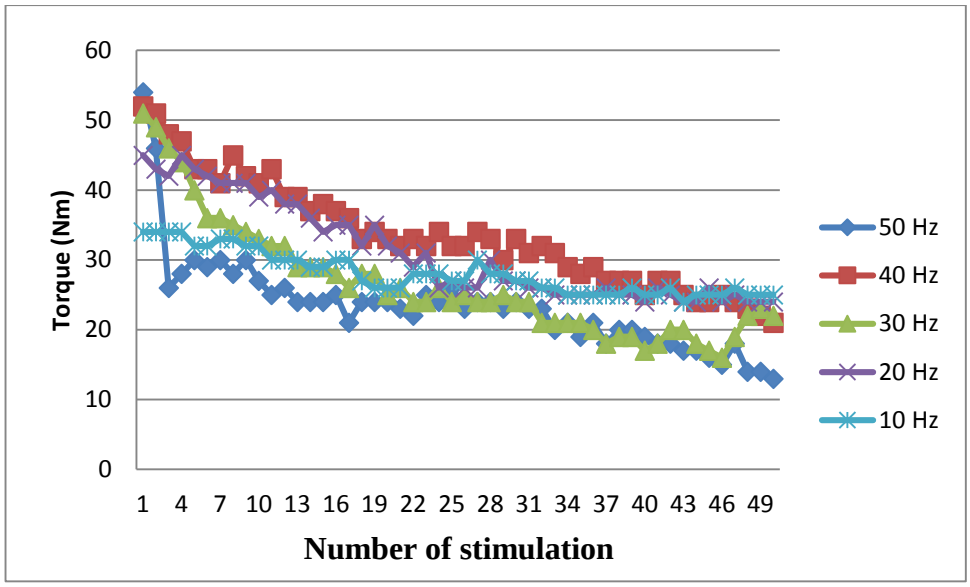


Figure 5.1: Averaged results for all 7 subjects at different frequencies with current fixed at 40mA, pulse width at 250µs, signal on for 3 seconds and off for 7 seconds.

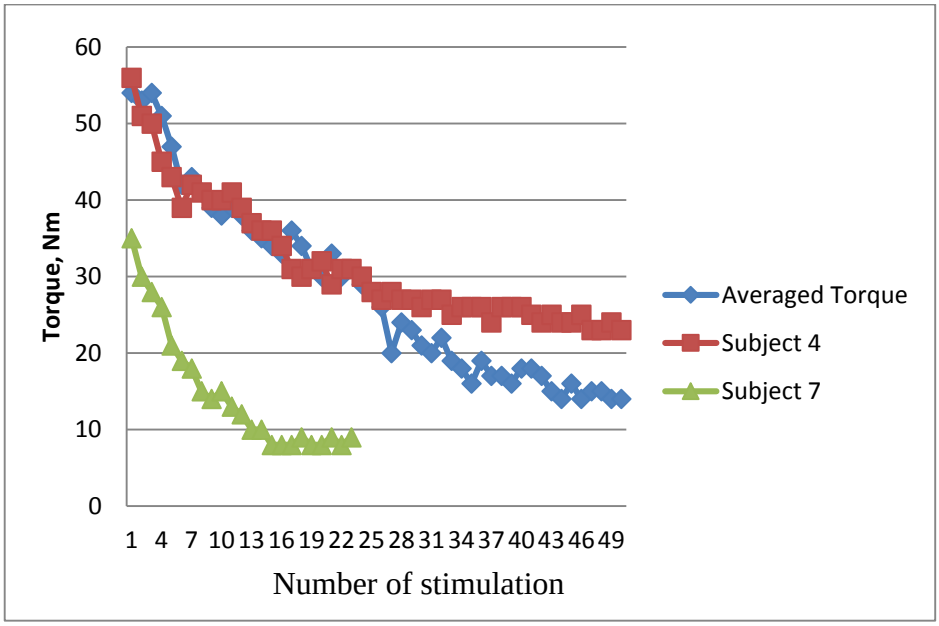


Figure 5.2: Comparison between the averaged torque results at 30 Hz with the results obtained from subjects 4 and 7

Table 5.2 summarises the individual results for each subject. It is shown that frequency modulation resulted in decline or incline of the output peak torque though it is shown that the healthier paraplegics with greater attention to their in-time rehabilitation program produced a higher peak torque.

Table 5.2: Individual results obtained from all 7 subjects

		10 Hz	20 Hz	30 Hz	40 Hz	50 Hz
S1	Torque Decline %	39%	43%	56%	60%	67%
	Maximum Torque (Nm)	46	51	56	58	61
S2	Torque Decline %	33%	53%	74%	76%	80%
	Maximum Torque (Nm)	39	49	54	52	55
S3	Torque Decline %	20%	37%	44%	45%	54%
	Maximum Torque (Nm)	73	83	86	89	92
S4	Torque Decline %	39%	55%	58%	60%	67%
	Maximum Torque (Nm)	46	55	56	58	61
S5	Torque Decline %	20%	46%	74%	78%	80%
	Maximum Torque (Nm)	39	49	54	55	57
S6	Torque Decline %	10%	35%	42%	48%	75%
	Maximum Torque (Nm)	67	78	80	84	103
S7	Torque Decline %	68%	67%	74%	77%	86%
	Maximum Torque (Nm)	25	31	30	40	50

By employing protocol 2 with all the parameters fixed but the pulse width, it is shown that the stimulation pulse width affects the output force but there is no significant effect on muscle fatigue as the decline of the peak torque is not of any significant importance. The quadriceps of subjects with less post injury complications are shown to be able to exert more torque throughout the experiment when compared to that of subjects who have received less rehabilitation attention after their injury. Figure 5.3 illustrates the average of normalised output torque of the subjects in the entire experimental spectrum while Figure 5.4 shows the comparison of this average with subject 3 who received in-time rehabilitation attention and subject 7 who did not receive appropriate rehabilitation attention at 300µs.

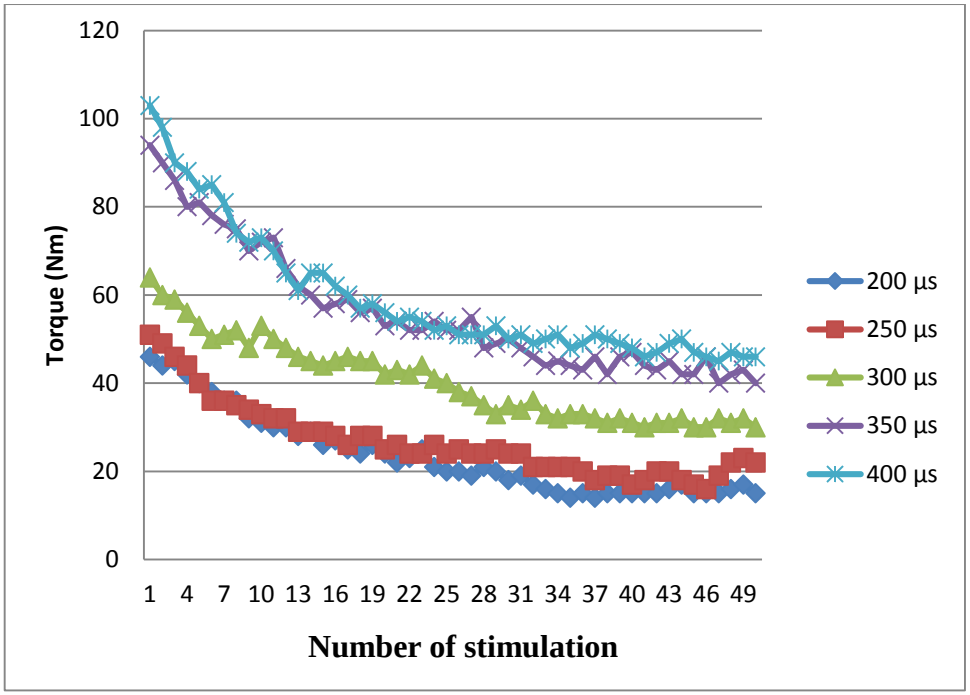


Figure 5.3: Averaged results for all 7 subjects at pulse widths with current fixed at 40mA, frequency at 30 Hz, signal on for 3 seconds and off for 7 seconds

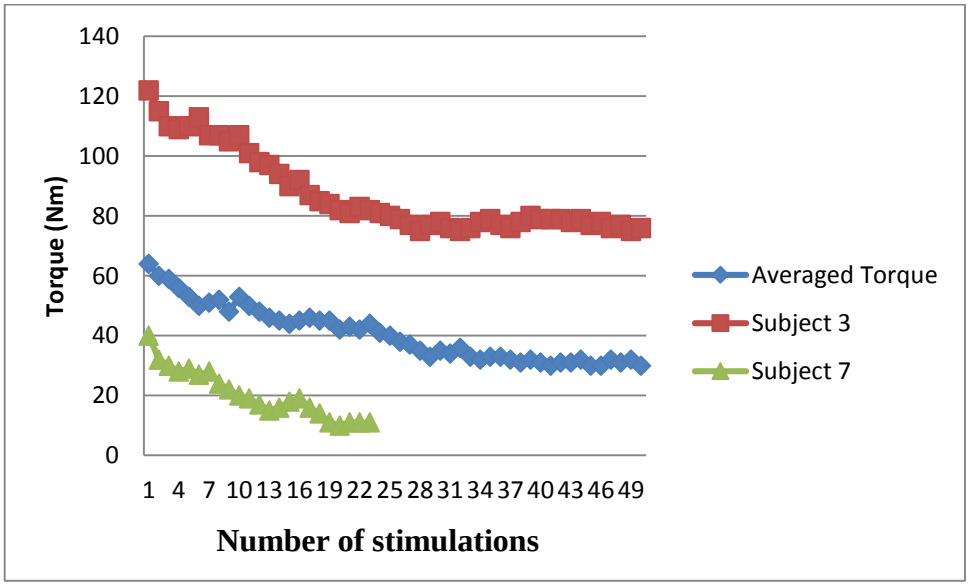


Figure 5.4: Comparison between the averaged torque results at 300 µs with the results obtained from subjects 3 and 7

Table 5.3 summarises the individual results for each subject which clearly shows that the subject with less experience of FES produced faster decline in output torque. The same is arguable for subjects with unattended injuries.

Table 5.3: Individual results obtained from all 7 subjects

		200 µs	250 µs	300 µs	350 µs	400 µs
S1	Torque Decline %	61%	58%	53%	59%	55%
	Maximum Torque (Nm)	52	56	77	105	112
S2	Torque Decline %	69%	74%	60%	60%	56%
	Maximum Torque (Nm)	49	54	75	100	108
S3	Torque Decline %	33%	44%	37%	44%	46%
	Maximum Torque (Nm)	74	86	122	144	164
S4	Torque Decline %	62%	58%	56%	64%	62%
	Maximum Torque (Nm)	50	56	79	104	110
S5	Torque Decline %	68%	74%	58%	61%	52%
	Maximum Torque (Nm)	58	54	70	102	104
S6	Torque Decline %	56%	42%	51%	50%	46%
	Maximum Torque (Nm)	79	80	119	154	171
S7	Torque Decline %	81%	74%	72%	58%	51%
	Maximum Torque (Nm)	27	35	40	48	60

Based on these experiments it is evident that the controlled parameter has an absolute effect on inducing fatigue in the electrically stimulated muscle. Thus, it is essential to implement these findings in the design of the appropriate controller to control a FES-based activity. It should be noted that the fatigue discussed here is mainly the premature fatigue. Fatigue is normally the natural reaction of the muscle fibres and nerves to overworking which might lead to damages of tissues and bones. In the case of FES where the sensory and command passages to the brain are disconnected, the controller should be designed in a way to be able to recognise premature fatigue from one of natural fatigue. The controlled parameter should therefore be chosen in a way to increase in time to overcome the natural fatigue and stop the workout when the muscle reaches its limit with regard to the produced output torque to limit excessive and unnecessary workout. It is however of essential importance to note that the health condition of the stimulated muscles' of SCI individual is a determining factor therefore implementing a sensory feedback such as output muscle torque

in the control design. Subjects who have started a rehabilitation program straight away after their injury or those who have recently got injured have shown that the choice of the controlled parameters does not have a significant effect on their muscle performance. However, subjects with longer injury time or those who have not worked out their paralysed muscles extensively after their time of injury have shown that the performance of their muscle is highly dependable on the controlled parameter.

5.5 Summary

In this chapter muscle fatigue, premature fatigue and natural fatigue have been defined and the factors affecting the fatigue in electrically activated muscles have been discussed in detail. It is noted that after paralysis the muscles and their fibres tend to lose their characteristics rapidly resulting in a greater nonlinearity and greater challenges controlling them under electrical stimulation. While the literature show an extensive review of the effect of pattern of stimulation on muscles it was noted that most of these studies have been performed on a small sample of SCI subjects or that the study compromise a combination of healthy and paralysed muscles. Moreover, there was no such study found to study the effect of controlled stimulation parameter on the muscle fatigue. Therefore, a study was carried out to analyse the effect of stimulation parameters, frequency and pulse width, on the issue of muscle fatigue on 7 SCI individuals. It has been concluded that in the SCI subjects who have had a weakness in their muscles due to not having been active enough muscle fatigue occurs faster by frequency modulation while the pulse width modulation does not affect the fatigue extensively and higher pulse widths produce higher output muscle torque. For the subjects who have been actively using and training their muscle since the time of the injury or the ones recently got injured it has been shown that the frequency modulation have a limited effect on inducing muscle fatigue while the pulse width modulation does not affect the fatigue and yet produce greater muscle torque.

Chapter 6

Control Methods for Dual Leg Movement

6.1 Introduction

In order to control a paralysed limb electrical stimulation is delivered to the muscles using certain patterns. Due to reverse recruitment order the muscles fatigue far more rapidly under electrical stimulation which limits the use of FES in many applications. In order to limit this effect, choosing a suitable control strategy as well as an appropriate stimulation regime is essential. Applying an appropriate control technique is thought to be an effective way of reducing muscle fatigue. Different control strategies have been proposed in the literature each addressing some of the difficulties and recommending alternative methods to overcome them.

In this chapter the VN4D model developed in Chapter 3 combined with the ANFIS muscle model developed in Chapter 4 is controlled separately using PID, fuzzy, adaptive neuro-fuzzy and iterative learning control (ILC). The performances of these controllers are evaluated by comparing their outputs to a reference trajectory. The controlled parameter which is pulse width is recorded during the control process and is analysed later. It is known that by inducing energy to the muscles they will start to fatigue therefore in this chapter the energy induced in the stimulated muscles using the 4 different control methods is measured and its effect on the muscle fatigue is studied.

Moreover, In order to evaluate the difference between employing different models in the outcome of whole simulation the VN4D model performance is compared to the performance of the pendulum model which was earlier developed in Chapter 3. The advantageous and disadvantageous of employing each of these models are discussed and the selection of VN4D as the modelling technique is justified.

6.2 Control strategies in FES

As discussed in the previous chapters, one of the limitations of electrical stimulation for restoration of functional movements, which is mainly a muscle physiology matter, is the issue of rapid fatigue. It is a known fact that muscle fatigues far more rapidly when artificially

stimulated than when excited by the central nervous system. As a result, successful implementation of FES paradigms for rehabilitation has been greatly limited by premature fatigue (Karu et al., 1995). In order to address this concern it is apparent that choosing a suitable control strategy as well as an appropriate stimulation regime in an assisted FES leg extension exercise will play an effective role to reduce fatigue in stimulated muscles. Applying an appropriate control technique is thought to be an effective way of reducing muscle fatigue. Different control strategies have been proposed in the literature each addressing some of the difficulties and recommending alternative methods to overcome them. Among these open loop (Buckett et al., 1988, Hoshimiya et al., 1989, Adamczyk and P. E. Crago, 2000), closed-loop (Crago et al., 1980, Chizeck et al., 1991, Watanabe, et al., 2002) and rule-based controllers (Kostov et al., 1995) have been employed.

Knowledge based control such as neural networks (Chen et al., 2004; Graupe and Kordylewski, 1994), fuzzy logic (Chen et al., 2004; Feng and Andrews, 1994; Davoodi and Andrews, 2004; Sau Kuen and Chizeck, 1994) and genetic algorithm (Davoodi and Andrews, 1999) have been employed in many research activities. Adaptive fuzzy logic have also been effectively used by Feng and Andrews (1994) to control FES for swinging leg. They found that the controller can customize a general rule based controller and adapt to the time-varying parameters due to muscle while Yu-Luen et al. (2004) found that fuzzy control solves the nonlinear problem by compensating for the motion trace errors between neural network control and actual system (Jailani, 2011). Every controller needs to adjust the controllled parameter(s) in order to perform a specific task. Based on the findings presented in Chapter 5 the controlled parameter throughout this chapter is chosen to be pulse width.

The purpose of this chapter is to evaluate the effect of the controlled parameter(s) on muscle fatigue through analysing the changes in the stimulation patterns and studying the amount of energy transferred to the muscles throughout the stimulation. In this section

different control strategies ranging from conventional strategies to inelegant control methods adopted throughout this thesis are discussed. The model is required to follow a predefined sinusoidal trajectory presented below by adjusting the control parameters.

$$\theta_r = 81.7 - 30\sin(4t) \tag{6.1}$$

This signal aims to extend the knee joint from 81.7° to 111.7°, and flex it back to the start point with a frequency of 0.636 Hz (4 rad/sec) in a smooth manner. The cyclical movement of the leg can be divided into 5 general phases: Main extension, resistance extension, full extension, main flexion and resistance flexion. The main extension is where quadriceps muscle is fully activated to allow the knee extension from the ready flexed position. The resistance extension is the phase where hamstring muscle is activated and quadriceps muscle is relaxed. This phase is necessary to slow down the shank extension just before the knees are fully extended. The full extension phase is the phase where the knees are maintained at fully extended position for a very short period of time just before flexing back to the ready flexed position. The main flexion phase is the phase where the hamstring muscle is again activated to drive the shank back to its initial position. In the resistance flexion phase the shank is required to slow down just before the knee is fully flexed by activating the quadriceps muscles.

6.2.1 PID control

Two sets of PID controllers are developed in order to generate the stimulation pulse widths necessary to maintain the predefined reference signal. The first PID controller (PID quadriceps) is used as knee extensor and the second one (PID hamstring) is used as knee flexor. To incorporate the phases recognised earlier for smooth control of cyclical motion of the shank a muscle selection block is designed to recognise the phase switching. The muscle selection block is consists of two main switches and a series of input and output. The switches are used to select one of the muscle groups, quadriceps or hamstring, to be activated depending on the phase they are in; the phase is derived from angular position of the shank.

The input to the PID controller is the knee trajectory error which is calculated as the difference between the reference and the actual trajectory obtained from the model. The output of the controller for quadriceps and hamstring muscles are the regulated stimulation pulse widths which by feeding to the muscle model generates the essential torque to drive the FES cyclical motion of the plant. The controller parameters are manually tuned to track the predefined reference trajectory. The tuned K_p, K_i and K_d gains for the quadriceps and hamstring controllers are reported in table 6.1.

Table 6.1: The tuned PID parameters for the controller

Control	K_p	K_i	K_d
Quadriceps controller	2.5	3.65	0.9
Hamstring controller	1.74	4.55	0.08

6.2.2 Fuzzy logic control

In order to track the predefined reference signal using fuzzy logic control (FLC), two FLC controllers, one to control the quadriceps and one to control the hamstring are developed. Unlike PID control of the cyclical leg motion there is no need for the muscle selection block and the selection is done using the fuzzy inference mechanism depending on the FLC input, membership functions and the fuzzy rules. The inputs of the controllers are the knee angle, the error, $e(t)$, and the change of error, $\Delta e(t)$, normalised by scaling factors G_E and $G_{\Delta E}$. The

outputs of the controllers are normalised stimulation pulse widths generated for both quadriceps and hamstring muscles. The scaling factors G_Q and G_H are used to denormalise the outputs. Similar to the PID control the generated stimulation pulse widths are then fed into the muscle model which in return produces the essential torque to drive the shank in a cyclical motion.

The inputs are fuzzified by using a fuzzy set of five equally distributed variables that are defined by Gaussian-shaped membership functions: negative big (NB), negative small (NS), zero (Z), positive small (PS) and positive big (PB). The modified knee angle trajectory is fuzzified by using a fuzzy set of two variables that are defined by sigmoidally-shaped membership functions: higher (H) and lower (L). The two fuzzy outputs obtained from the fired fuzzy rules of the FLCs are changes to crisp values using the centre of area defuzzification method. They are defuzzified by using a fuzzy set of five equally distributed variables that are defined by Gaussian-shaped membership functions: very low (VL), low (LW), middle (MD), high (HG) and very high (VH). The fuzzy outputs are generated by using the five by five fuzzy rules table. Tables 6.2 and 6.3 show the fuzzy rules used in the FLC control technique. Table 6.2 is the fuzzy rules table used for quadriceps muscles stimulation in main extension, full extension and resistance flexion phases, and selected when the modified knee angle is positive in value, whereas Table 6.3 is the fuzzy rules table used for hamstrings muscles stimulation in main flexion and resistance extension phases, and selected when the modified knee angle is negative in value.

Table 6.2: Fuzzy rules for quadriceps muscle stimulation

(Selected when the modified knee angle is positive)

Δe / e	NB	NS	Z	PS	PB
NB	VH	VH	HG	MD	MD
NS	VH	HG	MD	MD	LW
Z	HG	MD	MD	LW	VL
PS	MD	MD	LW	VL	VL
PB	MD	LW	VL	VL	VL

Table 6.3: Fuzzy rules for hamstrings muscle stimulation

(Selected when the modified knee angle is negative)

Δe / e	**NB**	**NS**	**Z**	**PS**	**PB**
NB	LW	LW	MD	MD	MD
NS	LW	MD	MD	MD	HG
Z	MD	MD	MD	HG	HG
PS	MD	MD	HG	HG	VH
PB	MD	HG	HG	VH	VH

6.2.3 Adaptive neuro fuzzy control

Adaptive control comprises adjustable parameter and a mechanism for automatic adjustment in real time in order to maintain a desired system performance. This technique can present an automatic tuning process in closed-loop for controller parameters and permitting online identification of critical parameters of the plant (Astrom et al., 1995; Landau et al., 1998). Adaptive control is chosen to cope with the disturbances appearing due to the issue of muscle fatigue. It does have the ability to improve the performance and permitting an automatic online tuning process. The control is designed to track the predefined reference trajectory. The performance output trajectory can be improved by including the adaptive control system as shown in Figure 6.1. This structure consists of plant, fuzzy logic controller and artificial neural network (ANN) as adaptive mechanism. The control parameters can be regulated real-time in a closed-loop structure. It is essential to develop a neural identifier of the plant for suitable network architecture such as the number of layers and neurons per layer. Moreover, appropriate inputs that distribute the network with adequate information about the plant have to be determined (Remy and Weck, 1995).

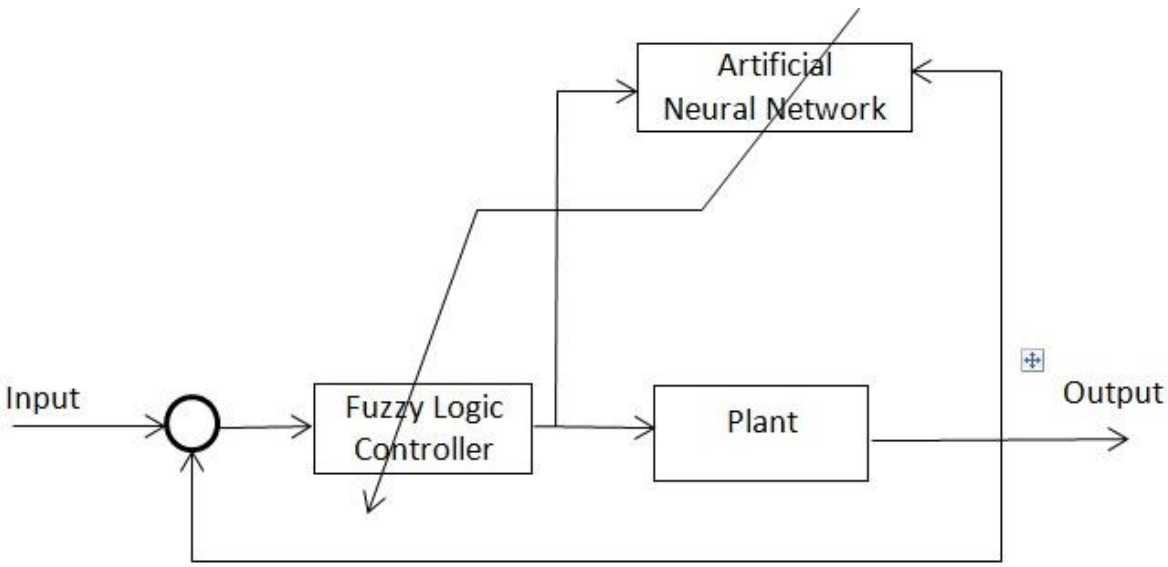


Figure 6.1: Adaptive Control Systems

Fuzzy Logic is applied as main controller. Unfortunately, due to muscle fatigue, advanced controller is still required to improve the control performance. Two fuzzy controllers which were introduced in section 6.2.2 are employed here to control knee extensor and flexor of the leg. Adaptive control is selected to deal with the fatigue disturbance whereas

ANN is used as adaptive mechanism. Brown and Harris (1994) have selected ANN for online adaptive control of FES-assisted rowing exercise because of its ability to learn significant information in real-time, with fewer restrictions, provable learning convergence and stability properties.

At the initial stage of the ANN model development, several parameters of data are acquired by simulating the process control with different fitness function rates for both muscles; quadriceps and hamstrings. The pulse width of FES as output of fuzzy logic controller needs to be tuned manually or by optimization to achieve the best pulse intensity which can produce better output trajectory. These make up a total of 30 data sets for training objective. It is fundamental for the learning rate η, the number of epochs (iteration), the hidden nodes (the number of neurons) and the number of hidden layers to be tuned perfectly in order to achieve rapid learning during the teaching process. Several combinations of these parameters can be applied to attain the convergence of error and also the respective optimal values. From the learning and training process, the learning rate, the number of epochs and the number of neuron are tuned within 0.1 to 0.9, 2000 to 20000 and 2 to 52 respectively to determine the best tuned set for optimal performance. The target is that the backpropagation error must meet at least 0.0001 and the number of hidden layers is fixed to unity to avoid the complexity of the ANN model. Therefore, it can reduce computation time consumption. Gradient descent training is used as the learning rule while the sigmoid function is employed as the transfer function of activation function. During training sessions, it was found that the best performance of this model can be achieved with the optimum values of learning rate, the number of epochs and the number of neuron equal to 0.3, 10000 and 52 respectively with one hidden layer. These parameters present acceptable result for the given test data. Fitness function rates at quadriceps and hamstring (MFq and MFh) are measured and processed by ANN model as input, and then the output of ANN will maintain the pulse width as output of fuzzy controllers to both muscles (PWq and PWh).

6.2.4 Iterative learning control

Iterative learning control (ILC) is an effective control tool for improving the transient response and tracking performance of uncertain dynamic systems that operate repetitively.

Systems typically treated under the ILC framework are repetitively operated dynamic systems, such as a robotic manipulator in a manufacturing environment or a chemical reactor in a batch processing application. Due to the excellent results it provides with nonlinear systems it has been tried to control FES systems in the past few years especially in upper extremities and ankle used control in stroke subjects (Nguyan, 2007).

The idea of ILC is straightforward: use the control information of the preceding trial to improve the control performance of the present trial. This is realised through memory based learning. Fig. 6.2 shows one such schematic diagram, where the subscript i denotes the i-th control trial.

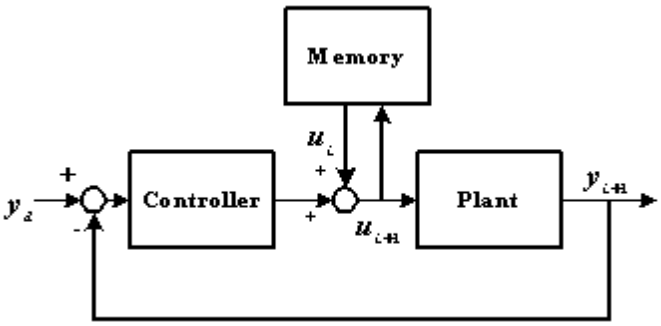


Figure 6.2: Memory based learning (Xu and Tan, 2003)

Assume that the controller is memoryless. It can be seen, in addition to the standard feedback loop, a set of memory components are used to record the control signal of the preceding trial, $u_i(t)$, which is incorporated into the present control, $u_{i+1}(t)$, in a pointwise manner. The sole purpose is to embed an internal model into the feed-through loop. To see how this can be achieved, assume that the target trajectory, $y_d(t)$, is repeated over a fixed time interval, and the plant is deterministic with exactly the same initialization condition. Suppose that the perfect output tracking is achieved at the i-th trial, i.e. $y_d(t)-y_i(t) = 0$, where $y_i(t)$ is the system output at the i-th trial. The feedback loop is equivalently broken up. $u_i(t)$ which did the perfect job will be preserved in the memory for the next trial. In the sequel $u_{i+1}(t) = u_i(t)$, which warrants a perfect tracking with pure feedforward. A typical ILC, shown in Figure 6.3, is somehow still different from Figure 6.2.

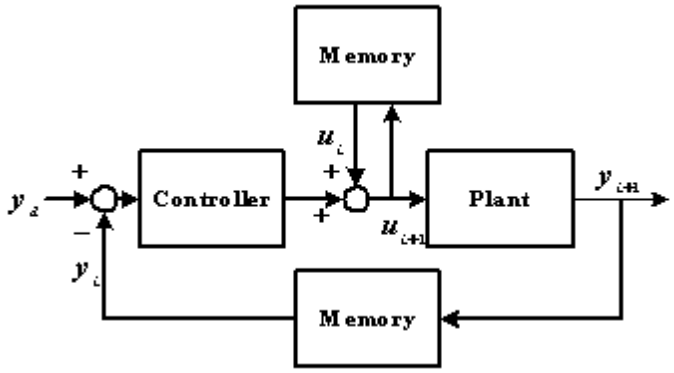


Figure 6.3: A typical ILC arrangement (Xu and Tan, 2003)

The interesting idea is to further remove the time domain feedback from the current control loop. Inappropriately closing the loop may lead to instability. To design an appropriate closed-loop controller, much of the process knowledge is required. Now suppose the process dynamics cannot escape to infinity during the tracking period that is always a finite interval in ILC tasks. It may not even be needed to design a stabilizing controller in the time domain, as far as the control system converges gradually when the learning process repeats. In this way an ILC can be designed with the minimum system knowledge (Xu and Tan, 2003).

ILC is a well-established area of study in control theory. The general principle of ILC is thought to be mentioned first in a patent by Garden (1967) under the title"Learning control of actuators in control systems" (Dijkstra, 2003). The first academic publication on ILC is dated back to 1978 in Japanese by Uchiyama (1978) where Arimoto et al, (1984), Casolino and Bartolini (1984) and Craig (1984) published the first series of papers on learning control in English on the topic of the use of learning control for robotics applications (Dijkstra, 2003). Arimoto (1984) used the term Iterative Learning Control (ILC) first and it has been widely used since then. A unified formulation of linear iterative learning control has been presented by Phan, et al. (2000) in AAS/AIAA Space Flight Mechanics Meeting. ILC, which can be categorized as an intelligent control methodology, is an approach for improving the transient performance of systems that operate repetitively over a fixed time interval. Although control theory provides numerous design tools for improving the response of a dynamic system, it is not always possible to achieve desired performance requirements, due to the presence of unmodeled dynamics or parametric uncertainties that are exhibited during actual system operation or the lack of suitable design techniques. ILC is a design tool that can be used to overcome the shortcomings of traditional controller design, especially for

obtaining a desired transient response, for the special case when the system of interest operates repetitively. For such systems, ILC can often be used to achieve perfect tracking, even when the model is uncertain or unknown and there is no information about the system structure and nonlinearity. For the purpose of design in this section a schematic working diagram as presented in Figure 6.4 is employed.

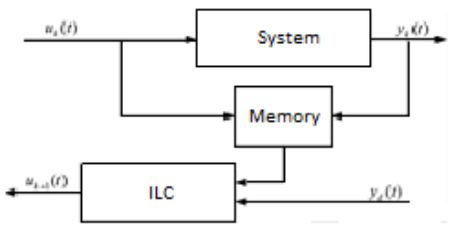


Figure 6.4: Parameter definition of an ILC used throughout this section

Consider the following linear continuous-time system:

$$\dot{x}_k(t) = Ax_k(t) + Bu_k(t) \tag{3.1}$$

$$y_k(t) = Cx_k(t) \tag{3.2}$$

The control task is to servo the output y_k to track the desired output y_d on a fixed interval $t \in [0, T]$ as the iteration k increases. If the system has relative degree one or less, an iterative learning control is given by $u_{k+1}=u_k+\Gamma \dot{e}_k$ where $e_k(t)=y_d(t)-y_k(t)$ and Γ is diagonal gain matrix, ensures that $\lim_{k\to\infty}(t) \to yd(t)$ for all t€[0,T], if $\| I-CBT\|_I < 1$

A number of more general expressions such as a "PID-like" update law can also be developed and given as $u_{k+1}=u_k \Phi e_k+\Gamma \dot{e}_k+ \Psi \int e_k dt$ where Φ,Γ, and Ψ are learning gain matrices.

6.3 Performances of the controllers

The performances of the 4 proposed control methods are evaluated using recorded trajectories from the VN4D leg model. The controlled parameter is pulse width while the rest of the main stimulation parameters are current at 40mA and frequency at 30Hz. The evaluation is done by comparing the leg movement trajectories obtained from the model to the earlier proposed reference trajectory of $\theta_r = 81.7 - 30\sin(4t)$. The behaviour of the system under all 4 control schemes proved to be of great acceptance when the maximum steady error was the studied parameter. The maximum steady error was recorded until the torque produced by the model is increased by 50% comparing to the initial torque required to move the modelled limb. The largest error seen among the proposed control methods during this period was 11.23 degree belonging to the PID controller and the least steady error of 5.4 degree for the fuzzy controller. The maximum average error for the same period of time for controlled movement in all 4 control methods was 5.20% for the PID controller while the least average error belonged to the ILC controller with 3% degree. Figure 6.5 graphically illustrates the tracking of the reference trajectory by the 4 control methods while Figure 6.6 shows the tracking error. Table 6.4 presents the maximum steady state and the maximum average error until the torque reaches 50% increase from its initial value in a numerical format. Twenty cycles were chosen due to the fact that all the 4 control methods achieved a 50% increase in their pulse width modulation and the emergency switch was switched on so that if this was an actual experiment the subject would not suffer from tissue and bone injury as explained in Chapter 5.

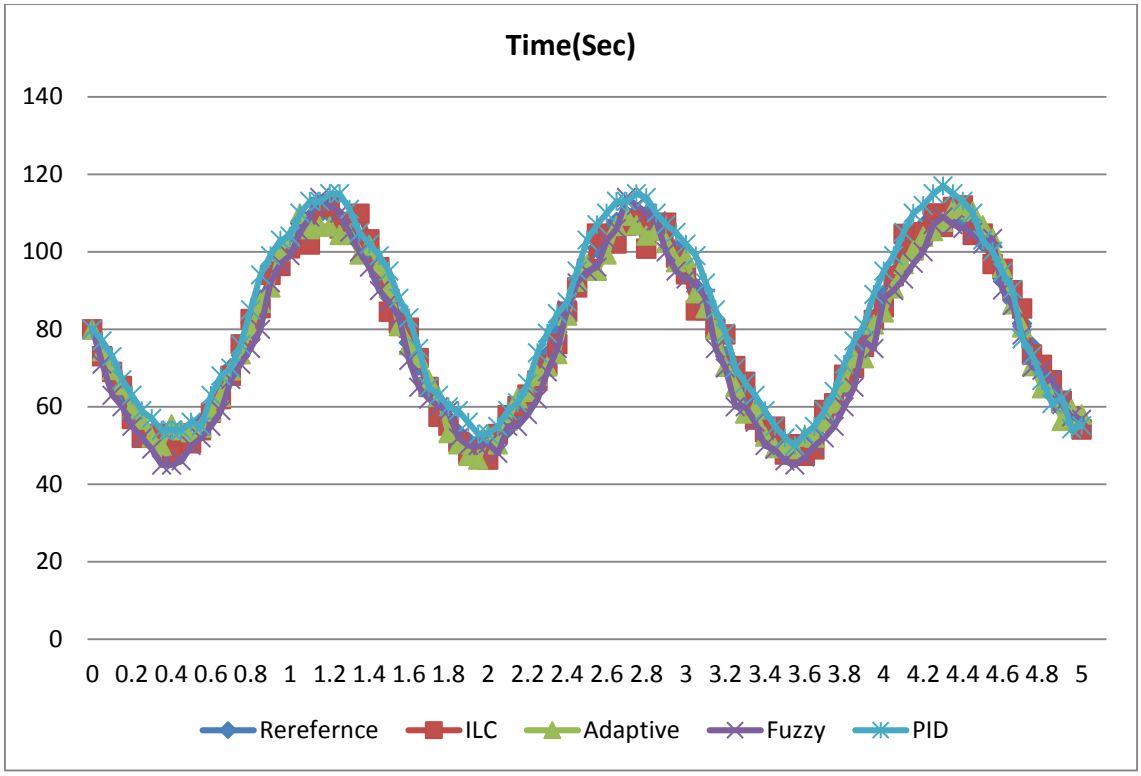


Figure 6.5: Tracking the predefined sinusoidal signal using the 4 proposed control methods

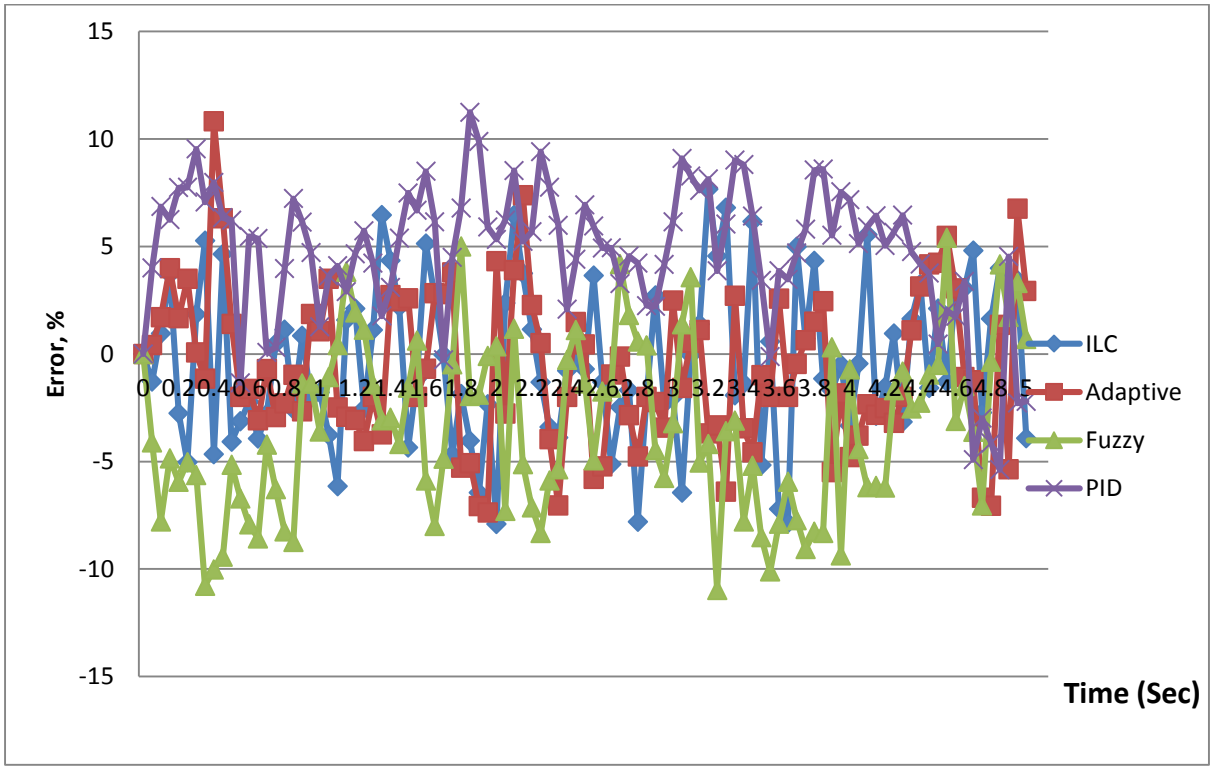


Figure 6.6: The tracking error of the 4 proposed control methods

Table 6.4: Numerical presentation of the steady state and averaged error until reaching 50% increase from the initial torque

Control Method	Maximum Steady State Error (degree)	Minimum Steady State Error (degree)	Averaged Error %
PID	11.23	0.27	5.17%
Fuzzy	5.40	0.08	4.20%
Adaptive	10.81	0.06	3.42%
ILC	7.66	0.06	3.02%

It can be noticed that maximum error occurs at the extremities of the trajectory at which point the phases switch from one to the other. At these points because of the velocity and the acceleration that the model gained prior to reaching the extremities the controller needs to react at the precise time to slow down the movement and make it ready to settle in the target position.

In order to exercise the effect of choosing different modelling methods the test, the same control methods which were applied to control the shank movement using VN4D software were applied to the simplified mathematical model proposed in section 3.3. The comparison between the steady-state and averaged error during the same cycles of the controlled shank movement for both modelling methods is presented in Table 6.5.

Table 6.5: Numerical presentation of the steady-state and averaged error for VN4D and pendulum models

		Maximum Steady State Error (degree)	Minimum Steady State Error (degree)	Average Error (%)
PID	VN4D	11.23	0.27	5.17%
	Pendulum	10.24	0.27	4.86%
Fuzzy	VN4D	5.40	0.08	4.20%
	Pendulum	6.17	0.17	4.97%
Adaptive	VN4D	10.81	0.06	3.42%
	Pendulum	12.24	0.19	4.87
ILC	VN4D	7.66	0.06	3.02%
	Pendulum	10.44	0.09	4.29%

As it can be noticed in Table 6.5 the steady-state error with all 4 control methods employed was higher in the mathematical modelling, and the averaged errors in the different control methods, except PID control, were higher using mathematical modelling. Although the implementation of the mathematical modelling proved to be much easier and less time consuming when compared to the VN4D modelling of the lower extremities, given the fact that using mathematical modelling proved to result in a higher level of error in tracking the reference trajectory mainly due to the assumption made when the modelling is done and also due to the capabilities of the VN4D in modelling the detailed sections of the lower extremities. The employment of physiological based modelling techniques can be of use in order to evaluate the effect of control techniques used in this chapter with much confidence as it will be discussed in detail in future work section in the Chapter 8.

6.4 Measurement of fatigue induced by using different control techniques

Four control approaches have been developed and used to separately control a modelled leg complex. The techniques were employed to control free swing cycle of the modelled leg in VN4D. This cyclical movement is designed so that in the real world an exercise can be performed employing FES to enhance the muscle health and prepare the subject for more complex and energy demanding FES exercises such as FES cycling and FES rowing. However, before adopting the actual exercise regime by patients, a study was needed to address the issue of muscle fatigue induced in the quadriceps and hamstrings due to the control methodology employed. Often in the process of industrialising a product when different approaches deliver almost the same accuracy the one with the lower cost of manufacturing will grab the most attention. However, in some cases other decision making factors should be taken into account as well. In this case the issue of fatigue induced in the muscles due to the chosen control approach has been investigated.

All four control approaches have shown their ability to keep the actual trajectory close to the predefined reference until the torque reaches 50% increase from the initial torque. Due to the high nonlinearity of the system the maximum error occurred during PID control of the VN4D plant in comparison to the other 3 methods which were more suitable for highly nonlinear models. It should be noted that the whole system was simulated to the best of knowledge and the incorporated muscle model is a highly valued one. It should be also noted that the controllers have been designed to follow a predefined trajectory with a 30 degree extension and flexion margins only. The high nonlinearity of the muscle complex and muscle model might have been shown extensively if the shank would have been extended up to 90 degrees. In this case a more sophisticated control technique can be employed. However, in reality an extension of 90 degree is not that essential because of the complexities arising and also due to the fact that this is not fundamental for most of the other FES activities. It should be also noted that the maximum error does happen at the extremities of extension and flexion. This is because the shank is trying to settle down from a moving position with a known velocity and acceleration raising a challenge for the controller to identify the phase and adjust the controlled parameters in order to complete the tracking of the predefined trajectory.

The main purpose of this section is not to discuss the performance enhancements of the controllers employed but to examine how they logically overcome the issue of muscle fatigue. As it has been discussed earlier one of the major FES limitations is that stimulated

muscles tend to fatigue very quickly because of the reversed recruitment order of the artificially stimulated motorneurons which limit the role of FES in certain applications (Rabischong and Guiraud, 1993). In this case increasing the pulse width of stimulation is employed to enable the targeted muscle torque to be maintained during this functional task. Other stimulation parameters such as frequency could have been modulated instead of pulse width to overcome the fatigue but pulse width has been chosen as mentioned earlier as the higher pulse width would not result in fatigue itself and only produce greater torque at the knee joint so it is a perfect fit to tackle the fatiguing muscle. Muscle fatigue is a natural element in daily human life and the human body does have a natural mechanism to overcome it and keep the body going but in the case of spinal cord injured individual adopting FES as a technique to provide them with functional movement, it is the task of the FES controller to replace this natural response.

In order to study the level of induced fatigue in the muscles using the developed model two separate methods are employed. The first method compromises of studying the trend of changes in pulse widths of all four controllers while the second method employs the use of measuring the amount of delivered energy to the muscles by employing all four different controllers.

6.4.1 Pulse width analysis to determine the effect of muscle fatigue

In order to follow the reference trajectory the controlled parameter, which is pulse width in this case, is adjusted in the course of time by the controllers. The changes in the pulse width for a typical cycle of the leg extension for four different control methods are plotted in Figure 6.7. It is essential for the pulse width to be adjusted during the time for the modelled leg to maintain the desired trajectory otherwise if a constant trend of pulse width is used muscle fatigue will occur and decay will be seen in the cyclical leg movement.

For each cycle the overall changes in the pulse width were averaged and plotted. By studying the trend in changes of these averaged pulse widths a sense of how the controllers try to overcome muscle fatigue can be established. It will also be made possible as such to see differences in pulse widths applied by different controllers in time, which in return will induce different levels of muscle fatigue. Figure 6.7 illustrates a randomly selected period of time in which the controller is adjusting the pulse width in order to control the movement of the modelled leg in Vn4D. This illustrates graphically how different control approaches has

employed different levels of pulse width in order to track a same reference trajectory. It can be seen that different control strategies have used different levels and pattern of stimulation.

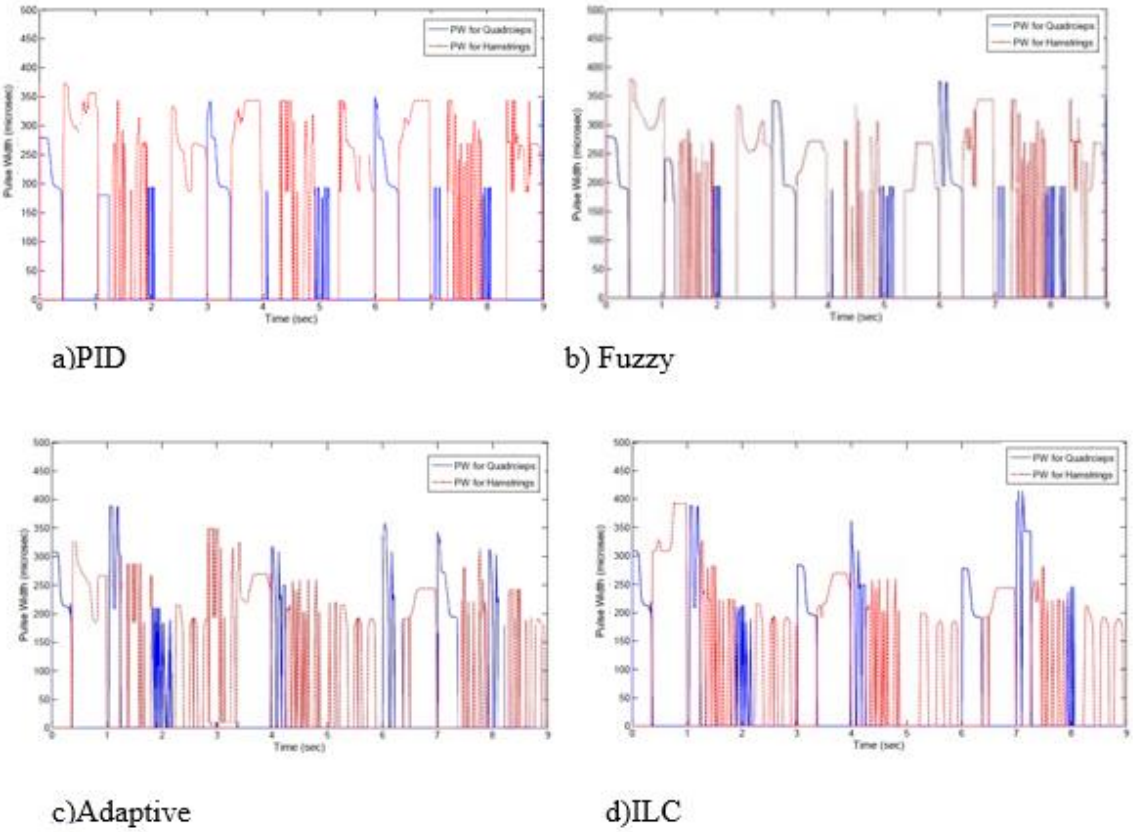


Figure 6.7: A randomly selected period of time in which the controller is adjusting the pulse width to control the lower extremities model. Vertical axis is the pulse width (µs) and the horizontal axis is the time (s). The blue line illustrates the level of pulse width used to stimulate the quadriceps while the red line illustrates the level of pulse width used to stimulate the hamstrings. Figure 6.7a presents the changes in the PID control and Figure 6.7b presents the changes in the fuzzy control, Figure 6.7c presents the changes in adaptive control and Figure 6.7d presents the changes in ILC

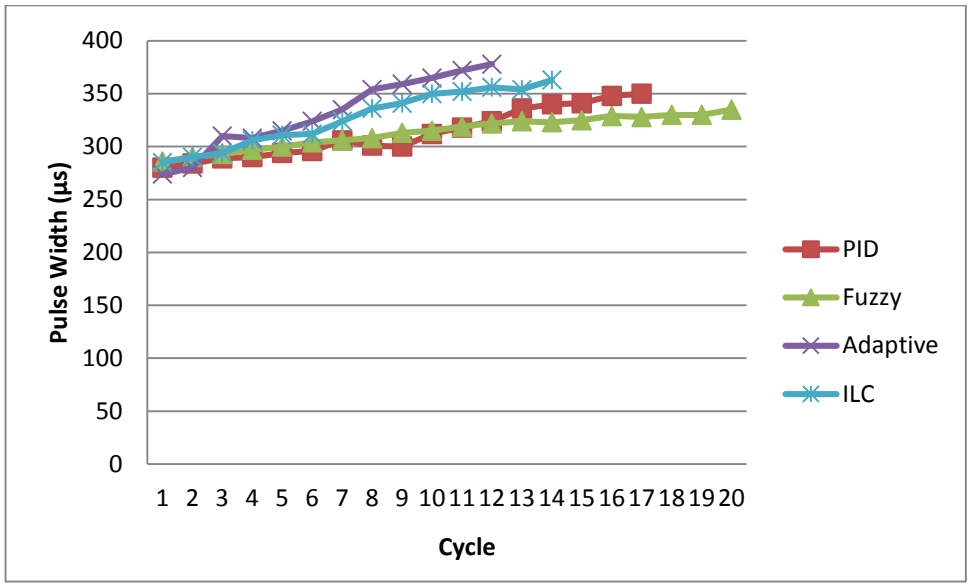
The averaged and maximum level of pulse width applied to the quadriceps and hamstrings are summarised in Table 6.6. The trend of the changes is of more interest than the one to one study of the pulse width changes.

Table 6.6: The trend of the changes in stimulation levels for the 4 control methods

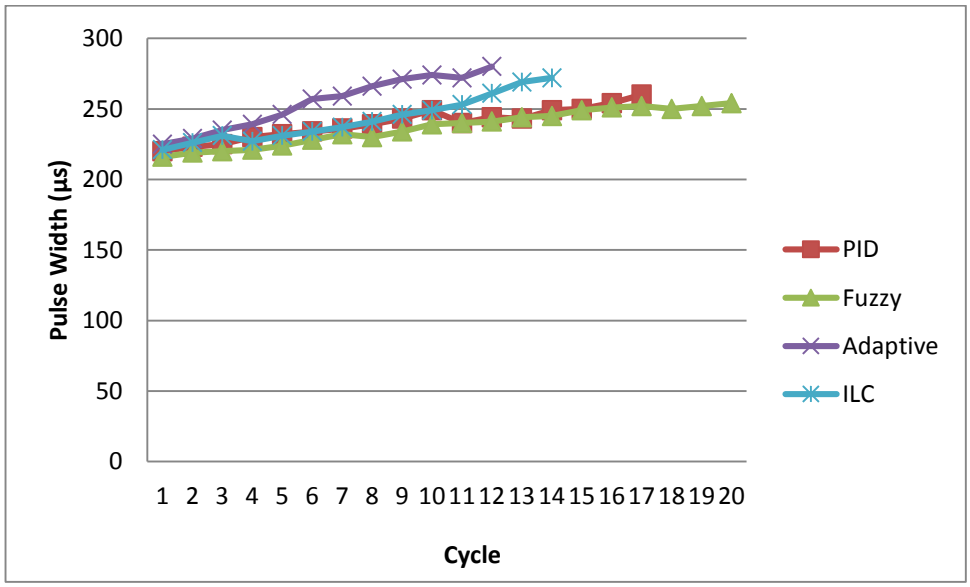
		Maximum Level of stimulation (µs)	Averaged Level of stimulation (µs)
PID	Quadriceps	406	303
	Hamstring	367	247
Fuzzy	Quadriceps	388	315
	Hamstring	432	240
Adaptive	Quadriceps	446	324
	Hamstring	443	254
ILC	Quadriceps	363	328
	Hamstring	451	246

As it has been discussed earlier one of the major FES limitations is that stimulated muscles tend to fatigue very quickly because of the 'reversed recruitment order' of the artificially stimulated motorneurons which limit the role of FES in certain applications. In this case increasing the pulse width of stimulation is employed to enable the targeted muscle force to be maintained during this functional task. Other stimulation parameters such as frequency could have been modulated instead of pulse width to overcome the fatigue but pulse width has been chosen as mentioned earlier as the higher pulse width would not result in fatigue itself and only produce greater torque at the knee joint so it is a perfect fit to tackle the fatiguing muscle. In this case by increasing the level of pulse width applied to the quadriceps muscles from 280 µs at the start of the task for PID, 293 µs for the fuzzy, 274 µs for adaptive and 285 for the ILC to the peak of 406 µs for the PID controller, 388 µs for fuzzy controller, 446 µs for adaptive and 373 µs for the ILC up to the point that the muscle model shuts down the process due to the torque reaching to 50% of the initial torque. The same method applies to the hamstrings with the parametric numbers stated in Table 6.6. While Figure 6.6 shows the pulse width modulation for random duration of the time for the 4 control methods, Figure 6.7 shows

the trend of averaged pulse width changes per cycle until it reaches to the 50% of the initial torque.



a) Changes in pulse width employed to control quadriceps



b) Changes in pulse width employed to control hamstrings

Figure 6.8: The trend of changes in the averaged pulse widths for each cycle employing different controllers. Figure 6.7a presents the changes in the pulse widths used to control the quadriceps muscles while figure 6.7b presents the changes in the pulse widths used to control the hamstring muscles.

While the initial pulse width recruited to produce the first cycle differs by less than 2% among all the controllers yet as it can be seen from Table 6.6 the pulse width applied to the quadriceps rose by 20% using the PID controller while it rose by 14.62% by fuzzy controller, 27.51% by adaptive control and 21.48% by ILC. The applied pulse width to the hamstring raised by 15.38% using the PID, while it rose by 14.9% by fuzzy controller, 19.64% by adaptive control and 18.75% by ILC.

The tracking of the pattern and the trend of pulse width can be used to compare different control techniques used in FES based activities to evaluate how they affect muscle fatigue. In adaptive and ILC methods the pulse width at hamstring decreased due to the weakness of the muscle. In contrast, pulse width at quadriceps was amplified by both adaptive and IL controllers in order to increase the torque of the muscle extension as well as to fix the output performance.

When the purpose of a sporting activity is just static endurance training and building muscle then it might not be too important how soon the muscle goes into fatigue as it is known when the muscle fatigues it does start building up if given enough resting time. There is a price to be paid for having the least error as the more sophisticated controllers intended to employ pulses with bigger pulse widths for longer times so that the error would be minimised. This will result in faster fatiguing of the muscles under ILC and adaptive control methods as it is seen from Figure 6.7 that the muscles reached to the defined irreversible fatigue point (50% increase in the output torque). If the purpose of training is of the endurance training then the price to be paid for 2% decrease to the trajectory following error is not justifiable as at least the subject will receive 5 cycles less in training for performing the same exercise. But if the purpose of stimulation is functional movement such as FES walking then the issue of lessening the error is becoming of the highest priority as the subject needs to be stabilised first and the external accessories such as wheel walker and balance hoist can be used to compensate for the cycles lost due to the effect of fatigue.

6.4.2 Induced energy analysis to determine the effect of muscle fatigue

Muscle fatigue is related to the energy consumption in the muscle, which in turn depends on the time integral of both the activation of a muscle (the stimulation input) and the actively generated muscle force (output of the muscle) (Peckham, 1981) and (Khang, 1989) . Under stimulation conditions the activation of the muscle is directly related to the number of

stimulation pulses and the level of recruitment u of every pulse. This leads to the following criterion for muscle fatigue (Veltink, 1992):

$$C_u = \sum_i u_i \tag{3.3}$$

The energy induced in the joints can be measured by $E = \int P.dt$ in which P_i is the power input to the joint that can be achieved by multiplying the external torque to the angular velocity of the joint as:

$$P_i = \tau_i \times \omega_i \tag{3.4}$$

A similar approach is employed in this section to evaluate and compare the performances of the four control approaches in terms of fatigue induction. With every pulse delivered to muscles certain amount of energy is induced into the stimulated muscle. The less the muscle is stimulated, the less energy transferred to the muscle resulting in less muscle fatigue. To overcome muscle fatigue the controller increases the pulse width in time resulting in more energy being transferred into the muscles. By studying the pattern of the pulse widths employed using the four control techniques in each identical cycle and also considering how the pulse width is raised from one cycle to another it is apparent that the time under which the muscles are stimulated is playing a major role in fatiguing the muscles.

Overall in the performed cycles using PID quadriceps were under stimulation for 77% of the time while the hamstrings were under stimulation for 65% of the time. This percentage is 71% for quadriceps and 60% for hamstrings using fuzzy, 80% for quadriceps and 69% for hamstrings using adaptive and finally 78% for quadriceps and 63% for hamstrings using ILC. This means that due to the fact that the muscles have been stimulated for shorter periods of time with less energy induced to them, the target muscle is less fatigued. This is summarised in Table 6.7. It should be noted that Table 6.7 illustrates the percentage of the time in which the muscle is stimulated for each control method.

Table 6.7: The percentage of the whole exercise time in which the muscle is under stimulation

		Time under which the muscle is under stimulation %
PID	Quadriceps	77
	Hamstring	65
Fuzzy	Quadriceps	71
	Hamstring	60
Adaptive	Quadriceps	80
	Hamstring	69
ILC	Quadriceps	78
	Hamstring	63

The cumulative effect of previous stimulations means that as time goes on to overcome the fatigue that has already been induced to the muscle, the larger pulse widths are employed and more energy is transferred into the muscles resulting in a jump in the pulse width modulation greater than before. The energy levels for one random cycle of the movement for all four control approaches are presented in Figure 6.8. Table 6.8 shows the averaged energy induced in the muscles for all cycles of the movement.

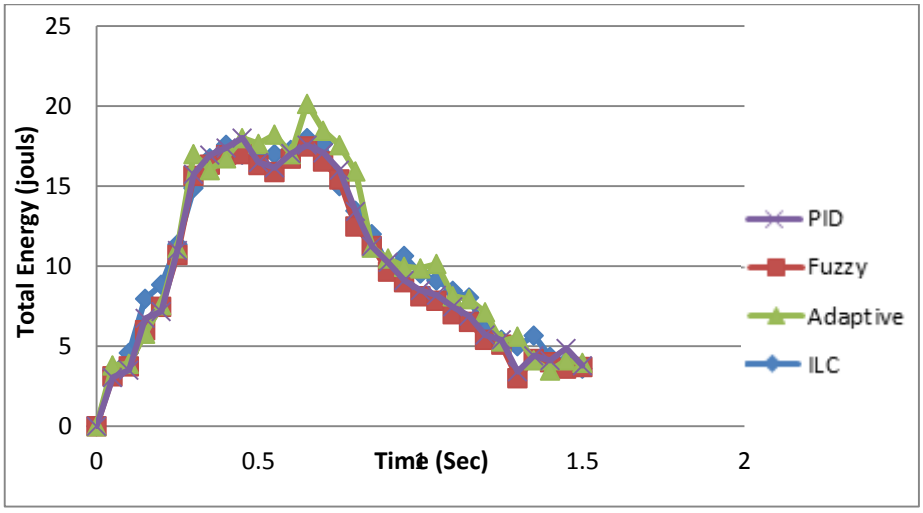


Figure 6.9: Total energy in the muscle throughout one cycle

Table 6.8: Averaged energy induced in the muscles for all cycles of the movement

	Number of cycles	Average energy in each cycle (joules)	Total energy for 12 cycles (joules)	Total Energy for the whole exercise (Joules)
PID	17	12.68	152.16	215.56
Fuzzy	20	11.84	142.08	236.8
Adaptive	12	14.93	179.16	179.16
ILC	15	14.83	177.96	222.45

It can be therefore concluded that while all the four control methods are keeping up with the predefined reference trajectory, the adaptive control needs to employ higher levels of pulse width to ensure that VN4D plant tracks the trajectory perfectly which in return will result in more energy employed at the knee joint, larger and constant levels of stimulation being employed resulting in less cycles of movement. Fuzzy and PID both are tracking the trajectories reasonably well while the level of energy is 20.69% and 15.07% less respectively compared to the adaptive controller. Using the same logic the level of energy induced in the muscles are 14.49% and 20.16% lower using fuzzy and PID when compared to ILC.

6.5 Summary

Due to the nature of different control methods dissimilar levels of stimulation are employed in order to control the movements of the limbs under electrical stimulation. These diverse levels of control will result in different levels of fatigue induced in the muscles under stimulation. In this chapter, four different control methods have been employed to adjust the pulse width in a way that enables a lower extremities model to follow a predefined trajectory until it reaches 50% increase in the torque. The controllers attempted to control a VN4D and a pendulum model of the legs separately in order to compare the tracking of the predefined trajectories by them. It has been shown that while the pendulum modelling simplifies the representation of the lower extremities and therefore saving time in the process of modelling, but due to the fact that it is not able to model the lower extremities in detail, it is not a suitable strategy to model a highly nonlinear mechanism such as the leg and foot complex. The four control methods employed proved to be able to track the reference trajectory

reasonably well but in order to do so they employed different levels and patterns of stimulation which in return resulted in different levels of fatigue. The fatigue induced in the muscles by different control methods have been studied using the analysis of changes in the pattern of pulse width and also the overall amount of energy induced in the targeted muscle. The more energy transferred to the muscle resulted in greater levels of muscle fatigue.

Chapter 7

Pulse Width Modulation and Energy Consumption Analysis During Practical FES Activities

7.1 Introduction

Due to the high nonlinearity of the musculoskeletal system especially in people suffering from SCI and the lack of information on some of the physiological matters of the human nervous and musculoskeletal systems the results of modelling and simulation do not always repeat themselves when the control approach is applied on an actual subject. The lack of movement resulted from the SCI will affect the condition of nerves and muscles in dissimilar ways in different individuals. Although simulation and modelling of the limb's movement under electrical stimulation help extensively in design and fine tuning of the control parameters and also save the patient from some unnecessary experimental procedures, but the results are not always as accurate in practical and clinical applications.

In this chapter the effect of some of the previously built controllers in Chapter 6 is evaluated on a group of subjects with regard to the trend of changes in pulse width, the amount of energy transferred to their muscles and more importantly the number of cycles they can achieve.

Two practical FES exercise activities including FES leg extension exercise and FES rowing are designed and controlled through simple yet practical feedback control to study the effect of employing external mechanical support on the issue of muscle fatigue. The fatigue element is extracted and based on the previously identified mapping strategy the fatigue which would have been induced to the muscle by the rest of the control approaches appeared in Chapter 6 is evaluated and the number of cycles the exercise could have continued under the other control schemes is predicted. The fatigue induced in the muscles practically is also compared to when some mechanical support parts is added to the exercise facility.

7.2 Fatigue profiling in practical applications

In healthy individuals fatigue is considered a natural mechanism to defend the musculoskeletal system from over working which may result in temporary or long term injury to the muscles and the bones. The physiological fatigue can be categorised as the inability to produce force due to a decline in sensitivity of troponin to calcium, meaning the strength of the muscle has been exhausted and the muscle no longer responds to brain signals, or fatigue due to the build-up of lactic acid which usually occurs from aerobatic activities and the body need for oxygen (Vigres, 2007). The muscles of those suffering from SCI or stroke lose shape and function much faster compared to those of healthy individuals with low levels of exercise activity. One of the reasons premature fatigue occurs much earlier than the normal fatigue is the result of muscles not been worked out for a period of time and is reversible if attended up to a certain stage (Allon et al., 2001).

When FES is recruited it should be noted that the muscle group is not over-worked as the connection to the brain is lost and it is the duty of the supervising technician or the controller to be able to recognise the premature fatigue from actual fatigue and dealt with it. The goal here is to overcome premature fatigue and fatigue the muscle as if it is done by the brain signals. Here, with pulse width as the control parameter, the controllers stop the cycle if the torque recruited reaches an increase of over 50% from its original set point to protect the musculoskeletal system from damage of any kind.

In order to compare the amount of energy induced in the muscles via different control methods a correspondence table between the pulse width and the energy consumed in the muscles during the performed cycles of exercise using VN4D model is developed. Table 7.1 represents these relationships.

Table 7.1: The PW and consumed energy relationship

Cycle	PID		Fuzzy		Adaptive		ILC	
	Average PW (µs)	Average Energy (j)	Average PW (µs)	Average Energy (j)	Average PW (µs)	Average Energy (j)	Average PW (µs)	Average Energy (j)
1	250	11.74	251	8.47	249.5	11.92	253	10.64
2	253.5	12.34	254.5	8.97	254.5	12.66	258	10.89
3	257	12.87	256.5	8.99	272.5	13.87	262.5	11.88
4	260	12.99	259	9.27	273.5	14.22	266.5	11.96
5	263	13.37	262	9.87	280.5	14	271	12.39
6	265	13.86	266	10.16	290.5	14.67	273	12.83
7	271	14.29	269	10.74	297	15.06	280.5	13.62
8	270	14.82	269	10.92	310	15.87	288.5	13.84
9	271.5	15.64	273.5	11.36	315	16.43	293.5	14.31
10	280.5	15.97	277	11.84	319.5	16.99	299.5	14.67
11	279	16.13	279.5	12.64	322	18.27	302.5	14.63
12	284	16.64	281.5	13.04	329		308.5	15.52
13	289.5	17.47	284	13.49			311.5	15.78
14	294.5	18.69	284	13.97			317.5	16.03
15	295.5	19	287	14.87			321	16.46
16	301	18.24	290	14.88				
17	305	19.67	290	15.24				
18			290	15.89				
19			291	16				
20			294.5	16.44				

Based on the direct relationship between the pulse width modulation and average energy recruited, as in Table 7.1 the number of cycles that each controller made possible for the VN4D model to complete can be used as a factor to relate these control approaches with each other with regard to the fatigue they induced in the muscles. Pulse width modulation and energy induced are both related and their final effect is to induce higher levels of fatigue which will eventually result in lower number of completed VN4D cycles. In order to establish a simple relationship table between these different control methods and number of cycles they each can make the VN4D model to perform, the number of cycles each of these methods can perform is divided by the number of cycles PID control was able to move the model and therefore, Table 7.2 is created.

Table 7.2: Energy relationship factor

	PID	Fuzzy	Adaptive	ILC
PID	1	0.85	1.41	1.13
Fuzzy	1.17	1	1.66	1.33
Adaptive	0.7	0.6	1	0.8
ILC	0.88	0.75	1.25	1

To apply these relations to practical applications of FES successfully and be able to adjust the pulse width accordingly two sets of FES exercises are designed. These exercises include the FES leg extension and FES rowing which will be done once by purely electrical stimulation of appropriate muscles and once by extra mechanical support combined with electrical stimulation.

7.3 FES dual leg extension exercise and its benefits

It was known from early years of practically employing FES that in paralysed limbs the muscles frequently become atrophied by disuse and the muscle response is both weak and rapidly fatiguing. Muscle, however, is a plastic structure which responds to the demands placed upon it as in exercise. Electrically induced 'exercise' has been employed to modify the physiologic and metabolic characteristics of both the disuse atrophied and normal muscle. The results of these studies have demonstrated that the chronically exercised muscle tends to assume properties which are similar to those of the slow motor unit. That is chronically stimulated fibres are fatigue resistant slow contracting and slow relaxing and have approximately the same cross sectional area as their unstimulated slow counterparts. Furthermore, stimulated fibres have a metabolic profile which is similar in many respects to the normal slow fibre which suggests that the internal energetics of the muscle have been modified to provide more efficient energy utilisation in situations of prolonged activity. Although the optimal stimulus regime has not been defined, it is known that these changes can occur in less than two weeks with continuously applied stimulation and can be maintained with substantially less stimulation (Peckham, 1981).

Weight training or resistance training as an organised exercise in which muscles of the body are made to contract in response to external weights in order to stimulate growth and strength is a perfect starting point for anyone suffering from SCI and wants to employ it together with FES to build muscles and train against premature or fast muscle fatigue before going on to more complex FES activities such as rowing, cycling and walking. The leg extension exercise is a very popular activity that can be easily combined with FES to train lower extremities muscles including quadriceps and hamstrings. A typical exercise machine uses a lever machine with a padded front bar which is pushed upward with extended legs. The exercise is widely used in knee-thigh rehabilitation with light weights and monitored repetitions. The exercise can be performed on right and left legs alternatively or can be performed by moving both legs at the same time.

The FES leg extension exercise will help to visually reshape the disabled muscles, reduce their fatigability, and improve their performance by increasing bulk, strength, power and endurance to prepare them for more complex FES activities. It will also result in secondary improvement to the musculoskeletal system which include better blood circulation, increase in bone mineral density and prevent lifestyle diseases such as diabetes, osteoporosis and obesity (Rodgers, 1991, Ezenwa, 1991, Belanger, 2000). The contraction of muscles, and hence the contraction of veins will result in a better circulation of blood in the lower extremities which will result in better nourishment of the muscles and bones and in return will reverse most of the primary and secondary complications that arise from SCI. The subjects who have been suffering from SCI for a long time and their injuries have not been through suitable rehabilitation programs will benefit the best from FES leg extension exercise as they need to build muscles and overturn premature and fast occurrence of fatigue before going on any other FES activity such as FES rowing or FES assisted walking.

7.3.1 The design of the exercise facility

The design of the exercise facility consists of a specially constructed exercise chair with padded seat and back support, as well as mechanisms for adjusting the back support position for various femur lengths, stabilizing the knee joint, applying load weights to the lever arms, and for obtaining feedback signals via limb angular position sensors to be used during knee extension exercise. As the controller pulse width ramps up, the paralyzed quadriceps muscles contract concentrically to lift the lower leg of the subject

and the attached load weights against gravity within a predetermined zero to 60 degree angle range. To return the limb and the weight stack to the resting position, a combination of quadriceps and hamstring muscles stimulation is employed via ramping down the quadriceps stimulation's pulse width and ramping up the hamstring stimulation's pulse width.

This system utilizes limb movement/position feedback sensors in conjunction with adaptive control methods to adjust stimulation parameters in order to achieve the desired performance as the muscles fatigue. During each contraction, the controller acquires the angular movement of the leg complex as well as the pulse width used for that cycle of movement and adjusts this variable for the next contraction. To maintain performance with fatigue, the pulse width is progressively increased nonlinearly by control circuitry. Resistance exercise can be performed by placing additional weights on the lever arms as muscle performance improves. Logic circuit detects muscle fatigue, spasticity, open electrode circuit and short circuit, and initiates appropriate system responses for subject safety.

Mechanical dampers are attached to the weight levers for two main purposes. First, to provide protection for subject's tissues should the system fails and the shank is dropped rapidly. Second, to control the system via stimulating quadriceps muscles only by ramping up the pulse width in extension and ramping it down during the flexion phase and using the damper it is better to control the shank's movement in flexion phase.

7.3.2 Control of the FES leg extension exercise by electrical stimulation only

A PID controller is used for the purpose of this section. In the absence of knowledge of the underlying process, a PID controller has historically been considered to be the best controller (Bennett, 1993). As explained earlier PID controller is a generic control mechanism which is used extensively in industrial control systems. This type of controller calculates the error which in the case of FES control of the leg extension exercise is the difference between the reference trajectory (a certain set point at a certain time) and the angular movement of the subject's leg under stimulation. The controller attempts to minimize the error by adjusting the controlled parameter which in this case is the pulse width.

The PID control strategy which was previously proposed in section 6.2.1 is employed. The input to the PID controller is the knee trajectory error which is calculated

as the difference between the reference and the actual trajectory obtained from the model. The outputs of the controller for quadriceps and hamstring muscles are the regulated stimulation pulse widths which by feeding to the actual muscle generate the essential torque to drive the FES leg extension exercise. For this purpose a controller-setbox is developed which can be used to adjust different parameters and develop different predefined trajectory signals. A semi-automatic controller was designed and implemented on an 8 bit PIC® microcontroller with maximum frequency of 8 MHz. Various parameters of the system including pulse width, frequency, delay time and reference velocity can be tuned online throughout the external remote device. The controlled stimulation parameter is the pulse width. The controller is connected to a PC via RS232 protocol and gets updated. Also it can use its internal memory chip to save the exercise history and send the data sets to a computer for displaying on monitor, printing a hard copy or for future mathematical analysis.

A closed loop control system was designed for this system. The main reference signal is the angular movement of the shank which can be adjusted by a skilled and trained physiotherapist through the control device but for the purpose of this exercise is fixed to extend no more than 60 degrees. This signal helps the controller force the system to provide the most possible similar movement for the patient to that of a normal person. In order to apply the best references signal possible 9 normal subjects have completed a 10 minute leg extension exercise and the recorded data was basically trained providing the expert physiotherapist with 6 exercise reference programs which can be chosen according the physical situation of each paraplegic subject through a digital knob placed on the controller box. The position of the shank which is measured using a 150mm Biopac Twin-Axis Goniometer is fed back to MATLAB through an analogue to digital converter (ADC) card and serial connection. Electrical stimulation is delivered via two MultiStick™ gel surface electrodes (Palsplatinum, 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 signals through the entire trial. A RehaStim Pro 8 channels (Hasomed GmbH, Germany) stimulator generates the stimulation pulses for application to the muscle. The setup of the experiment is presented in 7.1.

Figure 7.1: The setup of the experiment

By recording each session of the paraplegics' exercise and training the acquired data offline a more accurate reference signal can be developed for future use. In order for the original developed signals to be as accurate as possible, all the 9 non paraplegic participants' height, weight and level of daily activity have been chosen to lie within those of paraplegic subjects. Figure 7.2 shows the schematic of the controller while Figure 7.3 shows the actual controller-set box. The controller is connected to a RehaStim 8 channels stimulator which produces the exact stimulation signal to be delivered to the muscles.

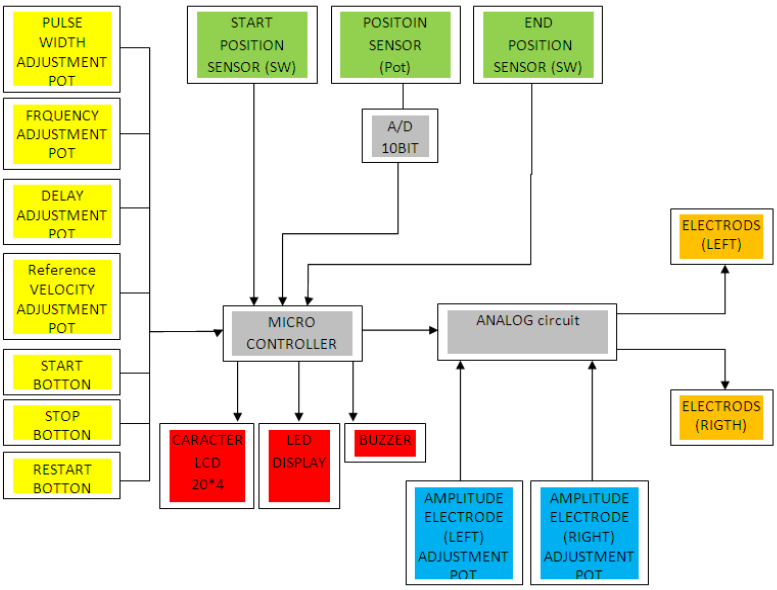


Figure 7.2: Schematics of the controller set box

Figure 7.3: Actual controller set box design

While observing the subjects performing the task using the PID control method it was apparent that although the practical controller managed to track the desired trajectory well giving the circumstances, the performance of the controller differed to what was expected compared to simulation. This is mainly due to the fact that there are some physiological matters that had not been taken into account during the modelling of the system as well as some of the extra activities the subject had prior to taking part in the exercise.

7.3.3 Control of the FES leg extension exercise with mechanical support

Combination of mechanical and electrical instruments often results in greater and smoother movements especially in robotics and mechatronics. Rasmussen et al. (2005) described efforts to subsequently optimize the ergonomic design of product based on computer model of human musculoskeletal system exemplified by the optimization of a spring-loaded bicycle crank. In this study two springs attached to the bicycle frame are used which allow the storage of elastic energy that help to overcome the dead centre of the cycle. This leads to uniform cycle allowing the rider to produce an even crank torque. It was also noted that the maximum muscle activity was less than half of the case with no spring mounted.

The use of physics laws and mechanical apparatuses together with the electrical stimulation has extremely helped in FES based activities. For example energy storage actuators which store the exceeded kinetic energy as potential energy and release it

again when needed, have been implemented in hybrid powered orthosis (HPO) (Gharooni et al., 2001, Durfee et al., 2005). Massoud (2007) has implemented a spring as an energy storage device in FES cycling while Gharooni et al., (2007), and Hussain (2009) implemented a spring in rowing exercise with quadriceps stimulation only as a mechanical storage device for the recovery phase.

In the case of FES leg extension exercise the same setting as section 7.3.1 is used in order to repeat the exercise. The only difference this time is adding a one way rotary damper (Ace controls UK, FYN-LA3-L) to the joint where the weight lever is connected to the seat. The maximum applicable torque for this model is 6 Nm. The additional mechanical support, in this case the one way rotary damper, will result in smoother movement of the system in flexion phase which will help specially at the extremities by engaging with the system and help in moving from one phase to another which will result in a much more smooth movement compared to when the stimulation was the only available resource. It will also help in less stimulation being delivered to the muscles and avoid the fatigue occurring fast.

7.3.4 Experimental results

The result from implementing the feedback control method for both modes of exercise, with and without mechanical support, is shown to be satisfactory when compared to the reference trajectory. The performances of the system in both modes with regard to following a predefined reference trajectory are presented graphically in Figure 7.4 and compared to the performance of the VN4D model under the same controller scheme.

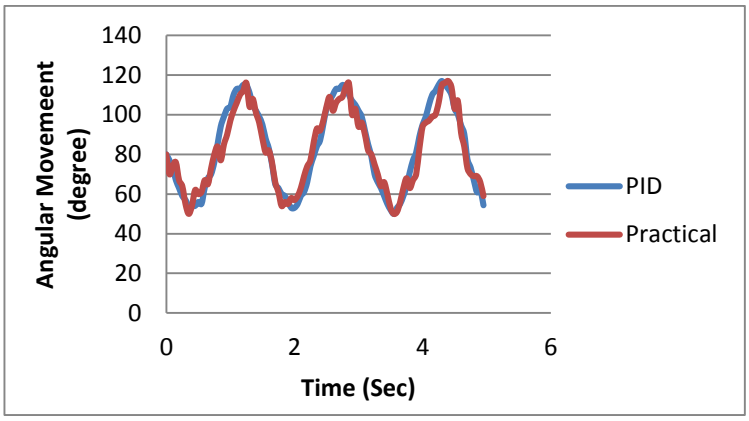


Figure 7.4: The performances of the controller with regard to following a predefined reference trajectory compared to the performance of the VN4D model

Although the result from the practical application of these control methods appears to be different to what was previously achieved in simulating the lower extremities using VN4D software, it is justified by considering the fact that the system is highly nonlinear and any surrounding and prior factors during and before experiment can affect the results. Also during the simulation it was considered that the only moving joint was the knee joint and the rest of the body was solidly attached and no extra movement existed, however in practice it was observed that hip joints tends to move constantly during the exercise, and a twitch was observed in the ankle joint during the extension and flexion of the shank which transformed the system into almost a double pendulum which in return resulted in extra degrees of freedom which was not considered during the simulation period. Moreover, although the subject was firmly tightened to the experimental rig it was observed that he tried to restore his position on the bench during and after each cycle which heavily effected and interrupted the recording, feedback and control of the lower extremities under stimulation.

The unexpected movements of the subject which added to the high non linearity of the lower extremities and resulted in extra degree of freedom compared to the simulation in VN4D resulted in different pattern of pulse widths delivered to the muscles and had the effect of fatigue shown 3 cycles earlier compared to the simulation. The energy patterns induced in the muscle by the same 2 modes of experiment, with and without mechanical support, is shown in Figure 7.5 for a typical cycle and compared to the energy patterns produced during the VN4D stimulation.

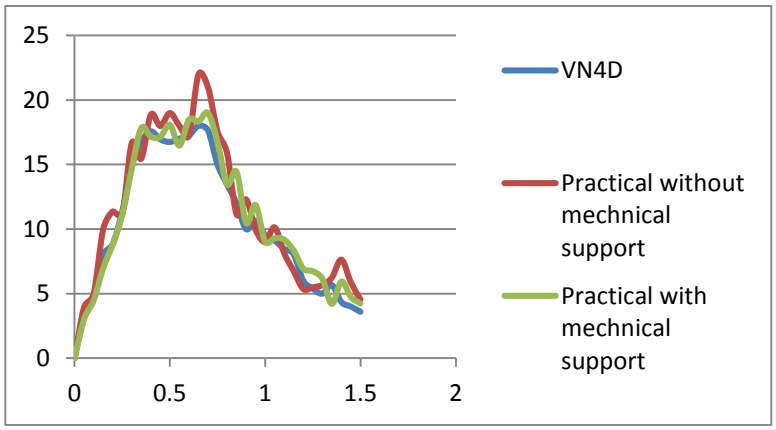


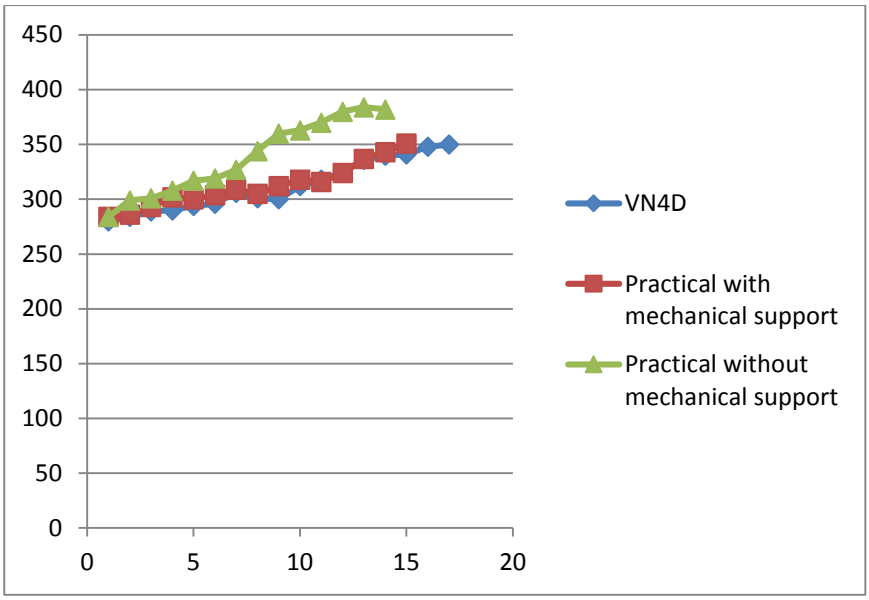
Figure7.5: Energy consumption for a typical practical test with and without mechanical support compared to the VN4D

The average energy induced in the muscles during the full exercise is noted in Table 7.3. As it can be seen from this table the stimulation only control mode induces almost 11% extra energy to what the same mode of exercise with mechanical support will induce in the muscles.

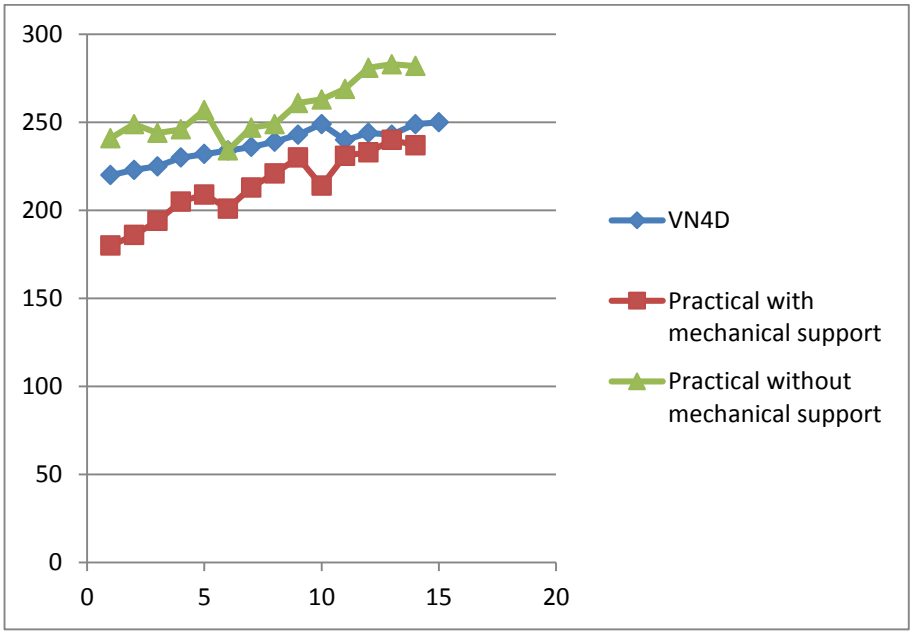
Table 7.3: The average energy induced in the muscles during the full exercise for the 2 modes of the exercise

	Number of cycles	**Average energy in each cycle (joules)**	**Total energy for 14 cycles (joules)**	**Total Energy for the whole exercise (Joules)**
VN4D	17	14.83	207.62	252.11
Practical (with mechanical support)	15	15.91	222.74	238.65
Practical (without Mechanical support)	14	17.84	249.76	249.76

Initially the objective was to perform the leg extension exercise with the stimulation of quadriceps muscles for the extension phase and the use of mechanical apparatuses to control the flexion phase where hamstring flex voluntarily by the gravitational force. However, it was made obvious after a few strokes of the movement that both the quadriceps and hamstring muscles specially rectus femoris and semitendinosus, showed irregular spasms which in one of the trials resulted in locking of the test subject knee joint. It was therefore decided that the hamstring muscles gets stimulated as well but obviously it was done with much lower levels due to the help it was receiving from the one way rotational damper. It is believed by the author that the more well trained the subject's muscle the less complication arises when stimulation is delivered to the quadriceps only. This is going to be investigated in a follow up study after this thesis. Figure 7.6 compares the averaged pulse width recruitment levels for the movement for both modes of exercise.



a) Quadriceps



b) Hamstrings

Figure 7.6: The averaged pulse width recruitment levels for in 2 modes of the exercise compared to VN4D modelling for quadriceps and hamstrings

7.4 FES rowing exercise and its benefit

Enhancing exercise opportunities for persons with SCI have become increasingly important as evidence stated in Chapter 2 shows an increased risk of diseases in this population. One possible opportunity to overcome these primary and secondary effects is enhancing physical activity through application of FES assisted lower extremities exercise combined with voluntary upper extremities exercise, referred to as hybrid FES exercise. The rationale of combining voluntary effort of upper extremities and FES-assisted lower extremities exercise for persons with SCI has been described by Hooker et al. (1992).

While the exercise opportunity for upper extremities has been looked into extensively initially when FES was firstly employed however due to small sizes of the upper extremities muscles the secondary benefits of the exercise was not significant. Employment of FES in the lower extremities of SCI individuals has enabled researchers to address this issue by exploring FES exercise facilities such as FES-evoked cycling, standing, rowing, leg extension, or stepping over the past 20 years. Some of the advantages of such exercise include augmented cardio-respiratory fitness, promotion of leg blood circulation, increased activity of specific metabolic enzymes or hormones, greater muscle volume and fibre size, enhanced functional exercise capacity such as strength and endurance, and altered bone mineral density.

The common hybrid FES exercise for persons with SCI combines arm cracking with FES cycling. However, this hybrid FES mechanism although effective, is somewhat cumbersome, expensive, immobile, and cannot be adapted for use by able-bodied. Over the years FES rowing has been explored thoroughly by academics, addressing a range of issues surrounding the physiology and controller development. Andrews et al. (1995, 2002), Wheeler et al. (2002) and Davoodi et al. (2001, 2002a, b, c, 2004, 2008) contributed immensely to FES rowing, resulting in development of a sophisticated FES rowing machine. They have reported increased levels of oxygen consumption during graded FES-assisted rowing and hybrid exercise. Their results compared favourably with levels achieved in subjects with similar lesion levels during knee extension-flexion exercise, ERGYS cycling studies, upper extremities crank ergometry and hybrid exercise combining upper extremity crank and ERGY cycling (Wheeler et al., 2002).

In this section development of a prototype FES rowing machine together with development of a feedback control approach is discussed. The controller is firstly designed to develop the stimulus needed for both the quadriceps and hamstrings without any added mechanical apparatus. Secondly the controller attempts to make possible the rowing exercise by applying electrical stimulation mainly to the quadriceps muscle and the recovery phase to the set position realised by providing a slope on the rail of the machine and attempting to recover from the extended position by using gravitational force. The smoothness of the movement as well as the pulse width trend employed are compared to when both quadriceps and hamstrings where stimulated.

7.4.1 Design of the rowing machine

The rowing sequence for paraplegic may differ from normal able-bodied rowing sequence. There are three phases of rowing stroke involved in FES assisted indoor rowing exercise: flexed ready position, drive phase and pull phase, as shown in Figure 7.7. Flexed ready position is the return phase from the maximal lower extremities extension to maximal flexed of the legs at knee and hips. This is followed by the drive phase which allows bilateral lower extremities extension of the legs and hips and body lean. Finally, the pull phase allows maximal lower extremities extension of the legs at knee and hips and upper extremities pull before returning to the flexed ready position (Wheeler et al., 2002).

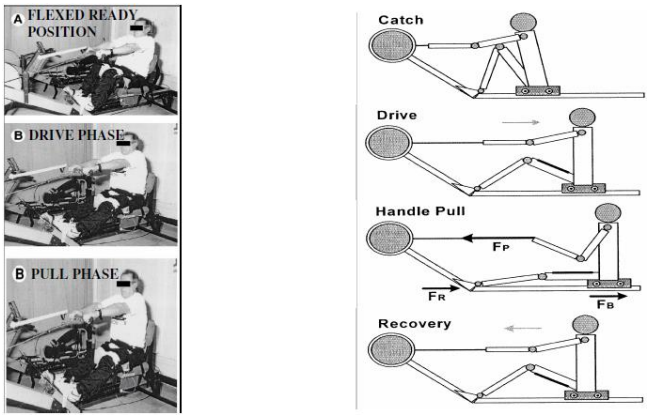


Figure 7.7: Rowing sequences for paraplegics (Wheeler, 2002)

Planning to develop a combined FES exercise facility in long run, a customised rowing exercise machine was developed. Doing so it was made sure that the ergonomics for rowing exercise are precise and in line with all the other commercially available

rowing machines. Being able to design the equipment in house there was no need to redesign some of the futures of a standard rowing machine. However it should be acknowledged that this took two month effort which is justified only because of the alternative usage of the facility that is explored for future projects.

Inspired by high end child car seat design, the sit provides sufficient postural support for paraplegic rowers. It offers an inclined high back, lumbar support and 3-point lap and diagonal seat strapping. The seat design constraints the trunk's involuntary movements and together with a telescopic two bar mechanism attached to the body of the machine helps in reducing the degree of freedom of the body parts below the level of injury almost to one. The mechanism is restricting the lower extremities' movement in sagital plane by attaching velcro straps around the thighs at one end and the other end to the frontal section of the rowing machine. Inside the velcros are cushioned with soft cotton velour which is breathable and does not cause skin irritation. The insider tube of the telescopic mechanism is attached to a special damper at the bottom allowing the extension of the mechanism without any resistance but providing a constant damping while it flexes back in order to provide hamstrings with adequate mechanical support to resist the joint locking as explained earlier from findings of FES leg extension exercise and passively training them using body mass.

The rowing machine's rail can be given a small slope in order to assist with the recovery phase using the gravitational force when the hamstrings are not stimulated; therefore in the design the seat and backrest were inclined to compensate for the slope. The seat movement on the rail is restricted in order to better control the system and to protect the lower extremities' muscles and joints at the marginal points of motion. To limit this movement two adjustable bars equipped with micro switches in their inner sides has been assembled on the rail. The bars can be adjusted based on anthropometric measurement of each subject. By reaching the extremities of the motion and the initial contact of the seat with the micro switch the phase of the motion is identified for certain. In order to smoothly start the movement in the catch phase, the controller smoothly raises the pulse width from zero to the point that the seat is dispatched and the phase changes from catch to drive. This is identified by the frontal micro switch turning off resulting in initiation of stimulation to the quadriceps. When the seat reaches the fully extended position the end switch is triggered, the final state is identified and appropriate adjustments to the stimulation signals are made. The structure of the machine is presented in Figure 7.8.

Figure 7.8: The design of the rowing machine

7.4.2 Control of the FES rowing exercise by electrical stimulation only

The same semi-automatic controller presented in section 7.3.2 is used for the purpose of controlling the rowing exercise for an SCI subject. A closed loop control system was designed for this system. The main reference signal is seat's velocity, which can be adjusted by a skilled and trained physiotherapist through the control device. This signal helps the controller to force the system to provide the most possible similar movement for the patient to that of a normal person. To track the reference signal, the controller uses two signals, seat position and direction of movement, provided by a multi turn position sensor that sends data sets to the microcontroller 32 times per second. The microcontroller uses these data sets to calculate the velocity as well. Two micro

switches placed at the start and end point of the machine rail mark the beginning and end of movement.

By pressing the start button in the catch position the controller applies the stimulation signal to the quadriceps muscles with the initial values. The extension of lower extremities starts till the seat reaches the end point of the track where it activates the rear end micro switch that results in microcontroller making a buzz sound to make the user aware of the new state. During the drive phase, the microcontroller checks the seat position regularly and adjusts the pulse width of the output signal to follow the reference velocity. To avoid muscle spasm, the controller can increase pulse width gradually from 0-500 µs. No coordination between the handle position and the leg extension is implemented at this stage.

In order to evaluate the effect of the presence of a mechanical supporting device on the possibility of muscle fatigue reduction, the rowing exercise was performed by 2 different methods. Once the subject was asked to perform the exercise with quadriceps and hamstring stimulation without any mechanical adjustments and once he was asked to perform the exercise with quadriceps and hamstring stimulation together with the support of the gas damper and the adjusted 2% slope. This experiment studied both approaches and results were analysed. Recovery phase starts automatically after the buzz is heard. To provide a safe exercise, microcontroller checks the seat's speed against the reference velocity and applies the stimulation signal to quadriceps and hamstrings muscles regularly in order to keep the movement going. To make the application safe if for any reason such as unexpected spasm or fatigue the user cannot follow the reference signal and the produced stimulation signal has reached the defined critical value, immediately the signal goes off by controller making a longer buzz to prevent any injury to the user.

7.4.3 Control of the FES rowing exercise with mechanical adjustment

Most of the indoor rowing machines are designed for healthy people. For FES-assisted indoor rowing machine designed for paraplegics with higher levels of muscle weakness, some modifications have been made. Wheeler et al. (2002) introduced a simple spring that was attached to the rowing frame and seat to assist the recovery phase of rowing cycle. Pull-out pegs were placed on the rowing tract at the limits of flexion and extension that compress the two gas struts for shock absorbance and energy return. The

return to flexion is assisted by the spring that overcomes the body's inertia in the extended system. The gas spring described by Davoodi et al. (2002c) also helps in phase to phase momentum transfer by storing the seat momentum energy in impact and releasing it when the seat moves in the opposite direction.

In this case the subject is asked to perform the exercise with quadriceps and hamstring stimulation together with the support of the gas damper and the adjusted 2% slope. The adjusted slope will result in a natural return from the fully extended position using gravity providing the hamstring with both passive and active treatment. The gas damper located inside of the two-bar mechanism helps in slowing down the descending of the seat, preventing any shock during the recovery phase and ensuring the smoothness of the recovery phase. The slope adjustment is highly important as setting it at a greater slope will result in difficulties during catch and drive phase and fatiguing the quadriceps at a greater level.

7.4.4 Experimental Results

The results from implementing the feedback control method for both mode of exercise, with and without adjustment to the slope and adding gas damper, have proven that the rowing exercise using both modes is possible. The performances of the system in both modes with regard to following a predefined reference trajectory are presented graphically in Figure 7.9 and compared to the performance of the VN4D model previously built by Sareh (2007).

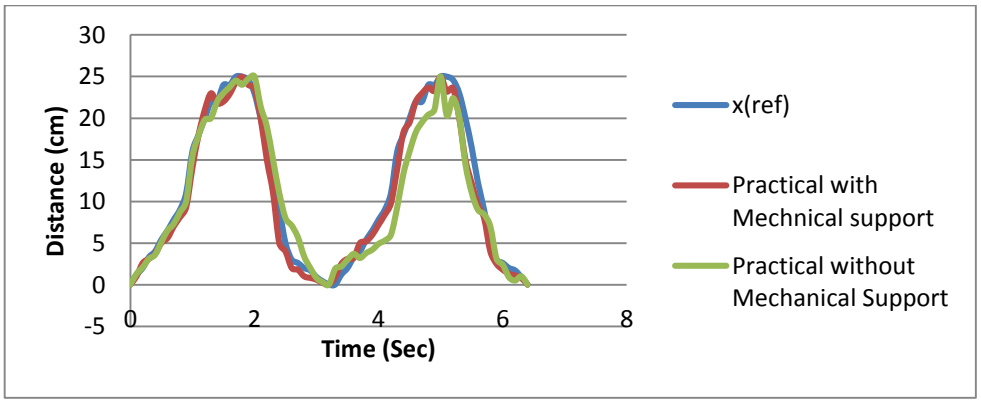


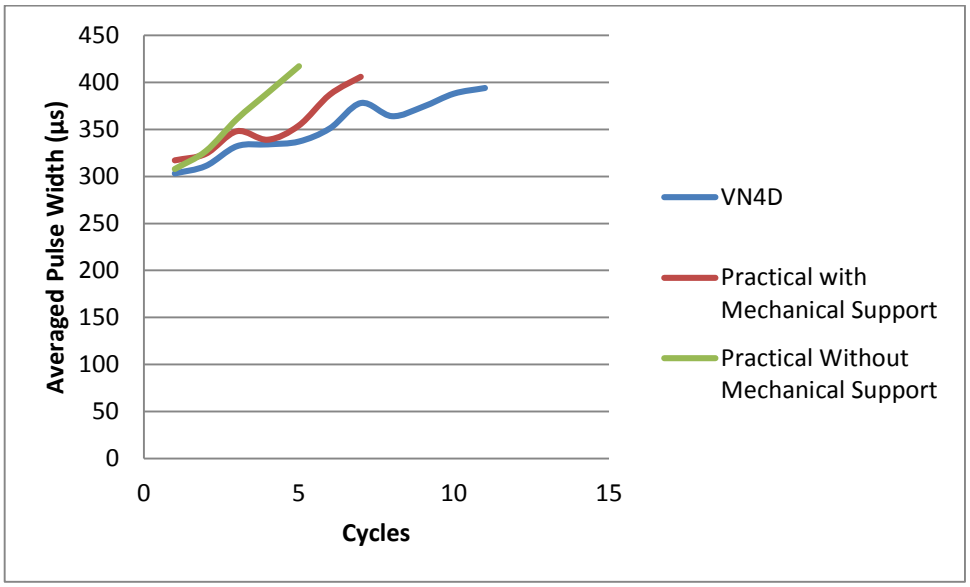
Figure 7.9: The performances of the controller with regard to following a predefined reference trajectory compared to the performance of the VN4D model

Again the results from the practical application of these control methods appeared to be quite different to what was expected from the modelling. The effects which were seen by simplifying assumptions for the modelling of leg extension exercise were much more of an issue here. Not only during the rowing simulation it was assumed that hands are moving in a linear form but also it was assumed that the only moving joints of the lower extremities was the heap and the knee joint. It was noted during the FES rowing exercise that the movement of the ankle joints plays a role in defining the rest of the trajectories for the lower extremities joints. Moreover, although the subject was fastened tightly to the seat yet any slight movement resulted in different trajectories for the rest of the joints. These complications have resulted in different patterns of pulse widths delivered to the muscles and had an early effect on muscle fatigue. This resulted in 4 cycle cut to what would have happened under Sareh (2007) simulation of FES rowing in VN4D. The averaged energy patterns induced in the muscle by the same 2 modes of experiment, with and without mechanical support, is shown in Figure 7.10 and compared to the energy patterns produced during the VN4D stimulation. The average energy induced in the muscles during the full exercise is noted in Table 7.4. As it can be seen from this table the stimulation only control mode induces almost 21% extra energy to what the same mode of exercise with mechanical support will induce in the muscles.

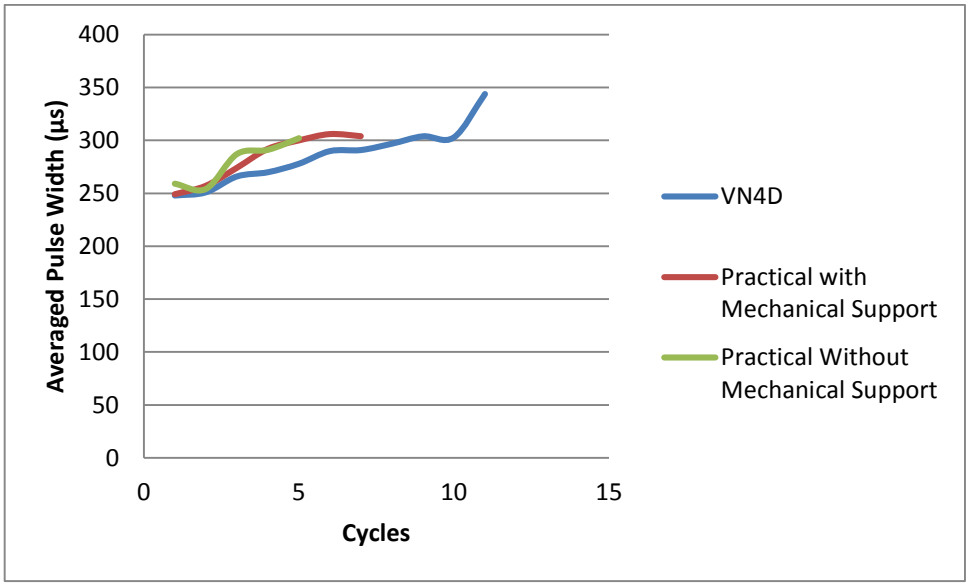
Table 7.4: The average energy induced in the muscles during the full exercise for the 2 modes of the exercise

	Number of cycles	**Average energy in each cycle (joules)**	**Total energy for 5 cycles (joules)**	**Total Energy for the whole exercise (Joules)**
VN4D	11	18.61	93.05	204.71
Practical (with mechanical support)	7	21.19	105.59	148.33
Practical (without Mechanical support)	5	26.74	133.7	133.7

Figure 7.10 compares the averaged pulse width recruited during VN4D model control, the practical with mechanical support and the practical without mechanical support.



a) Averaged pulse width recruited to stimulate the quadriceps



b) Averaged pulse width recruited to stimulate the hamstrings

Figure 7.10: the averaged pulse width recruitment levels for VN4D and two modes of exercise. a) Averaged stimulation level for quadriceps, b) Averaged stimulation level for hamstrings

7.5 The effect of cycle factors on other control methods

So far in this chapter the results from implementing a feedback control method in 2 different FES exercise facilities with and without mechanical assist have been studied. By developing a map between the energy induced in the muscles during simulation and practical exercises it is made possible to study the effect of control method on the fatigue during practical applications. FES leg extension and FES rowing exercise have been chosen for the purpose of this section so that they could represent the simplest and the most demand forms of FES exercise training.

As it was noted in practical application with FES leg extension the number of cycles the subject was able to perform with and without mechanical support were respectively 15 and 14 cycles while this was 7 and 5 for the FES rowing exercise. Therefore the number of cycles the Vn4D model was able to perform can be related to the number of cycles a subject is able to complete the exercise using PID control. These factors are respectively 0.88 (15/17) for leg extension and 0.63 (7/11) for the rowing in the mode of receiving mechanical support. The same ratios can be calculated for the exercise mode in which no mechanical support has been employed which will result in the cycle factors of 0.82 and 0.45 for leg extension and rowing exercise respectively.

To have the findings in a more general terms the cycle factors for leg extension and rowing exercises can be averaged and used to predict how many cycles one can actually do based on the simulation result and the choice of control method. These cycle factors are summarised in Table 7.5.

Table 7.5: Adjusted cycle factors

Existence of mechanical support	**Cycle Factors**
With Mechanical Support	0.85
Without Mechanical Support	0.54

These factors can be then multiplied by that proposed in Table 7.2 and therefore the number of realistic cycles an exercise is repeatable based on the choice of control method can be achieved.

7.6 Summary

The effects of some of the previously built controllers in Chapter 6 have been evaluated on SCI subjects with regard to the trend of changes in pulse width and the energy induction to the muscles. A mapping table relating the fatigue elements to the stimulation element have been derived. Two practical FES exercise activities including FES leg extension exercise and FES rowing have been designed and controlled in a feedback control setting. The effect of employing mechanical facilitators in a FES exercise activity has been studied. The fatigue element from the designed exercise activities has been extracted and based on the previously identified mapping strategy the fatigue which would have been induced to the muscle by the rest of the control approaches has been predicted.

Chapter 8

Conclusion and Future Work

This chapter presents relevant conclusions drawn from the outcome of this research and attempts to place these in the context of the effect which muscle fatigue will have on the choice of controlled parameters and control methods when it comes to employing functional electrical stimulation (FES) as an exercise facility in spinal cord injury (SCI). The achievements made throughout this thesis with regard to mapping muscle fatigue to controlled stimulation parameter and the control method employed are highlighted. Finally, recommendations for future work in these studies are made.

8.1 Summary and concluding remarks

This research has primarily been motivated by the need for understanding the effects that different control strategies and different control parameters will have on muscle fatigue. Research has always looked into applying new control methods to the control of FES powered facilities with regard to smooth and well controlled movements. However, it has generally lacked in understanding of the effect of applying these methods on muscle fatigue. Studying the fatigue induced in the muscles by different control approaches is essential when it comes to commercialising the FES equipments as the cost, easiness to use and cosmetics are the factors playing a major role in the process of equipment design.

In order to achieve this aim successfully, every aspect from modelling to control have been investigated and analysed vigilantly. Furthermore, it has been attempted that models and guidelines provided in this thesis will provide the research and industry with footpath on how to design the system to minimise the effect of fatigue. It should be noted that throughout this thesis muscle fatigue has not been referred to as a negative matter. Muscle fatigue is the natural response of the musculoskeletal system when it gets over worked. Certain levels of muscle fatigue are of therapeutic nature and help the overall health of the muscles and the body which can be induced by different forms of endurance muscle training. The effect of premature and fast muscle fatigue are

also of importance when the purpose of the activity is not endurance training but daily function such as locomotion. Therefore choosing the right controlled stimulation parameter as well as the right control method is suddenly becoming of a high priority.

For the purpose of this thesis several stages have been completed. First and prior to entering the engineering phase of the thesis a comprehensive review of what is SCI, what primary and secondary complications arise from it, what are the financial burdens on individuals and the whole economy and the way these complications can be handled with optimum costing, has been provided. FES has been introduced as an innovative rather recent methodology in providing the SCI subjects with functional movement and savings.

Second, an accurate model of humanoid leg extension exercise that replicates the real environment has to be developed in order to simulate the FES-leg extension activity. This is one of the crucial stages as the model has to represent the system in the most accurate way. In this study, the humanoid model has been modelled with anthropometric measurement based on the studies performed on 15 randomly chosen SCI subjects. These studies allowed the model to represent a complete demographic spectrum of SCI subjects using MSC.visualNastran4D software environment (VN4D) and also by employing a simplified model of the lower extremities using pendulum mathematics. The knee joints passive elements especially stiffness and viscosity, have been extracted from the mentioned 15 SCI subjects through Waterberg's methodology and incorporated into both modelling techniques. The performances of both sets of modelling techniques have been evaluated by free pendular movement test and VN4D model has proved to replicate the actual lower extremities better. The VN4D allows the model to be simulated as a virtual physical environment with the ability of measurement made in real-time. The VN4D proved as impressive and highly competent software that can replace conventional approaches involving complex mathematical modelling. Moreover, the ability of VN4D to be integrated with Matlab/Simulink environment has given additional advantages in this study where the interaction of the humanoid model with the muscle model and control system has shown real world conditions visually and measurably.

Third, the importance of having a model representing the knee's musculoskeletal system as a function which predicts the output of the muscles with

regard to torque and therefore the angular movement, has argued. This is a crucial part in this thesis as the human muscle is known to be a highly complex, time-varying and nonlinear dynamic system. Therefore, it is necessary to develop a muscle model by a suitable approach to cope with the complexity and uncertainty of the model and yet represent the actual system accurately. Extensive study has been done on different muscle models developed experimentally or physically and the majority of them have been found not to be suitable to use in control applications. This is because these models characterise each muscle features alone and there is no relation between the model features which may prevent them from combining as one complete model. Also the issue of muscle fatigue has not been taken into account in some of these studies. A very good recent muscle model which includes the function of fatigue has been developed by Riener. The Riener's muscle model is a physiological based muscle model that comprises calcium dynamics and muscle fatigue. Unfortunately, it needs many of the muscle parameters to be identified and that requires customised experimental procedures and special facilities which restrict the development and implementation of this model.

In this thesis, two new experimental based muscle models have been further investigated. Primarily a NARMAX-OLS black box based model has been further studied. This model proved to be able to track a predefined trajectory exceptionally well but since it was noted that the model is fit only to track a non-loaded (free swing) leg model a secondary approach which had been developed initially by Jailani (2011) using ANFIS technique was upgraded. In her study Jailani (2011) used just one subject for the purpose of data acquisition and she used isometric force measurement as a method of evaluating the muscle output under different rays of stimulation. Since, all the FES activities are performed with the extremities under stimulation moving the isokinetic torque measurement has been employed to replace the isometric force measurement proposed. Moreover, the model has been developed by employing 15 different SCI subjects with different levels of injury and health conditions to build a more global model. Also a greater spectrum of the stimulation parameters has been utilised.

Fourth, the issue of fatigue in muscles activated by functional electrical stimulation has been looked into in detail comparing the difference between voluntary and electrically stimulated muscle contraction, and further studying the factors affecting fatigue in electrically induced contractions including stimulation parameters and stimulation pattern. Furthermore, a series of experiments aimed at understanding the

parameters affecting the muscle fatigue in quadriceps muscles have been performed and the results have then been used in determining the controlled parameters to control the paralysed limbs. For the purpose of this study pulse width has been chosen as the only controlled parameter which must be adjusted in time to overcome the muscle fatigue.

Fifth, it has been noted that applying an appropriate control technique is thought to be an effective way of reducing muscle fatigue since different control strategies have proven to adjust the stimulation pattern in certain manners. The VN4D model developed has been combined with the ANFIS muscle model and controlled separately using PID, fuzzy, adaptive neuro-fuzzy and iterative learning control (ILC). The performances of these controllers have been evaluated by comparing their outputs to a reference trajectory. The controlled parameter (pulse width) has been recorded during the control process and is analysed. Since it is known that muscle fatigue starts after energy is cumulatively induced in the muscles a relationship between the energy induced in the muscles, pulse width modulation and the control method has been realised.

Sixth, based on the patterns recognised between the pulse width modulation, energy induction and control method a relationship table has been developed. The relationship table has been built based upon the simulation. To make this relationship more in line with the practice application and to see the effect of some of the developed controllers in reality with regard to the trend of changes in pulse width and the amount of energy transferred to the muscles, two practical FES exercise activities including FES leg extension exercise and FES rowing have been designed and controlled through a feedback control method. The fatigue element has been extracted from these exercises and based on the previously identified mapping strategy the fatigue which would have been induced to the muscle by the developed control approaches has been predicted. The fatigue induced in the muscles practically has also been compared to when some mechanical support parts is added to the exercise facilities and a separate relationship matrix has been developed.

8.2 Recommendations for further work

With respect to the targets, guidelines, methodology and time restrictions for different aspects of this research, it is noted that the targets have been met, deliverables have been produced and footpath has been set for future work. The research presented throughout this thesis has answered many of the previously unresolved questions. However, a number of questions have been raised as well, and some of them need detailed attention by the research community.

1. Although, the use of visual Nastran (VN4D) software as well as the pendulum modelling of the lower extremities proved to be able to highly replicate the thigh, shank and foot complex the possibility of extending the search to replace the modelling software should not be left out. Some of the differences we've noticed between the simulation and practical results arise from the way the system is mechanically modelled under VN4D. Having a biomechanical based software such as SIMM will further improve the lower extremities model and will result in a more accurate modelling of the body.
2. The process of NARMAX-OLS modelling proved to be highly accurate and robust. However, further investigation on how to make the model stable when different load sizes are added to the frontal part of lower shank is essential and can be undertaken as a part of another project.
3. When choosing which control approaches to take on to control the FES leg extension it was tried to have a selection of basic to complicated well known control techniques to cover a whole range of possibilities. However, having developed a complete set of controllers for the purpose of FES exercise and mobility will add value to the relationship matrices developed in Chapter 7. It will also extensively help in further developing the relationship table if the control approaches which have been employed in simulation can be built in a practical manner and the performance of the relationship matrix can be further studied.
4. It was tried to minimise the use of non-tuned parameters in this work. However, offline tuning of some of the parameters employed throughout this thesis will further provide fine robust controllers.

5. While it was tried to investigate the effect of controlled parameters on the muscle fatigue by mimicking it to controlling the pulse width only the possibility of having a few more adjustable stimulation parameters to control the plant and actual system should be explored.
6. It was tried originally to show controllability of the practical section of the work by combining mechanical apparatuses with quadriceps stimulation. However, after having seen the effect of quadriceps stimulation only on the subject leg it was decided to go with stimulation of both quadriceps and hamstrings. The effect of readiness of the subject to go under electrical stimulation should be further investigated as in the author's experience if the subject would have been well endurance trained before taking up the exercise the performance of the system with quadriceps stimulation would have been made possible.
7. Having worked with different SCI subjects throughout this thesis the author believes that the training protocols for those subjects with a long history of paralysis and those who have not been treated through therapeutic programs right away after the injury should be altered and a new protocol should be advised for these individuals.

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