Chapter 1

Introduction

This thesis describes the development of a forward dynamic model to predict the performance of Spinal Cord Injured (SCI) individuals cycling using Neuromuscular Electrical Stimulation (NMES) elicited contractions of their denervated leg muscles. This research was conducted with two goals in mind. Firstly, it was intended that the use of NMES elicited contractions would provide a means for validating the model, and would thus contribute to the field of muscle modelling in general. Secondly, it was anticipated that this research would further our understanding of the relationship between NMES timing and cycling performance by SCI individuals. It was planned to use the model to customise NMES timing parameters for individual subjects cycling a novel isokinetic ergometer, adapted for use by individuals with SCI within the Faculty of Health Sciences at The University of Sydney.

1.1 Background

Cycling Exercise for Individuals with Spinal Cord Injury

Spinal Cord Injury is a condition affecting many aspects of the health and lifestyle of those affected. While the incidence of new SCI cases has declined in Australia in recent years, the population of existing cases continues to increase because of increasing survival rates (Blumer and Quine, 1996). There are a number of secondary complications arising from SCI, apart from just paralysis, including higher risks of cardiovascular disease, pressure sores, osteoporosis, urinary tract infections and respiratory system disorders (Petrofsky and Smith, 1988). Exercise of the denervated leg muscles using NMES has been proposed as a partial solution to many of these secondary complications.

Paraplegics cycling using NMES to elicit muscle contractions was first trialed by Petrofsky and colleagues from Wright State University in the early 1980s (Petrofsky et al. 1983, 1984).

A number of groups have developed similar methods since then using ergometers of different levels of complexity (eg Bremner et al., 1992; Mulder et al., 1989; Popp, 1983). Popp (1983) measured EMG patterns from able-bodied cyclists using his ergometer to determine the range of crank angles when muscles should be stimulated. This method did not achieve an effective cycling action and hence "trial and error" was used to determine stimulation angles producing effective propulsion for each subject.

It is interesting to speculate why Popp's use of able-bodied EMG did not produce suitable stimulation patterns for NMES cycling. Some possible explanations may be:

- Electromechanical delay and muscle activation responses may differ between voluntary and electrically evoked contractions. Therefore, providing stimulation at the same time as recorded EMG signals would not necessarily result in the same pattern of force production.
- Electrically stimulated cycling involves relatively crude muscle activation patterns with only a few muscle groups able to be activated independently. If able bodied cyclists recruit many synergistic muscles with slightly different firing times for each muscle, then patterns based on these individual muscle activations may not be suitable when whole muscle groups are activated simultaneously with electrical stimulation.
- Differences in leg geometry may cause the appropriate firing angles to be so different between individuals that taking measurements from one person would not predict suitable angles for another individual.

The Ergys 2 (Theraputic Alliances Inc, Fairborn, Ohio) is a commercially available ergometer that has been used widely for both home rehabilitation and scientific experiments (eg Hooker et al., 1992; Raymond et al., 1999). Sinclair et al. (1996) measured the torque applied to cycle cranks by SCI individuals cycling an Ergys cycle ergometer with stimulation applied to the quadriceps, hamstring and gluteal muscles. When compared with able-bodied subjects cycling against identical resistances, SCI individuals produced significantly higher peak torques (Figure 1.1.1). Able-bodied subjects increased overall work load by increasing torque relatively consistently across all crank angles. By contrast, SCI individuals could only apply torque during the relatively short periods each muscle was active. Consequently, the peak torques had to be higher for SCI subjects in order to achieve the same average workload.

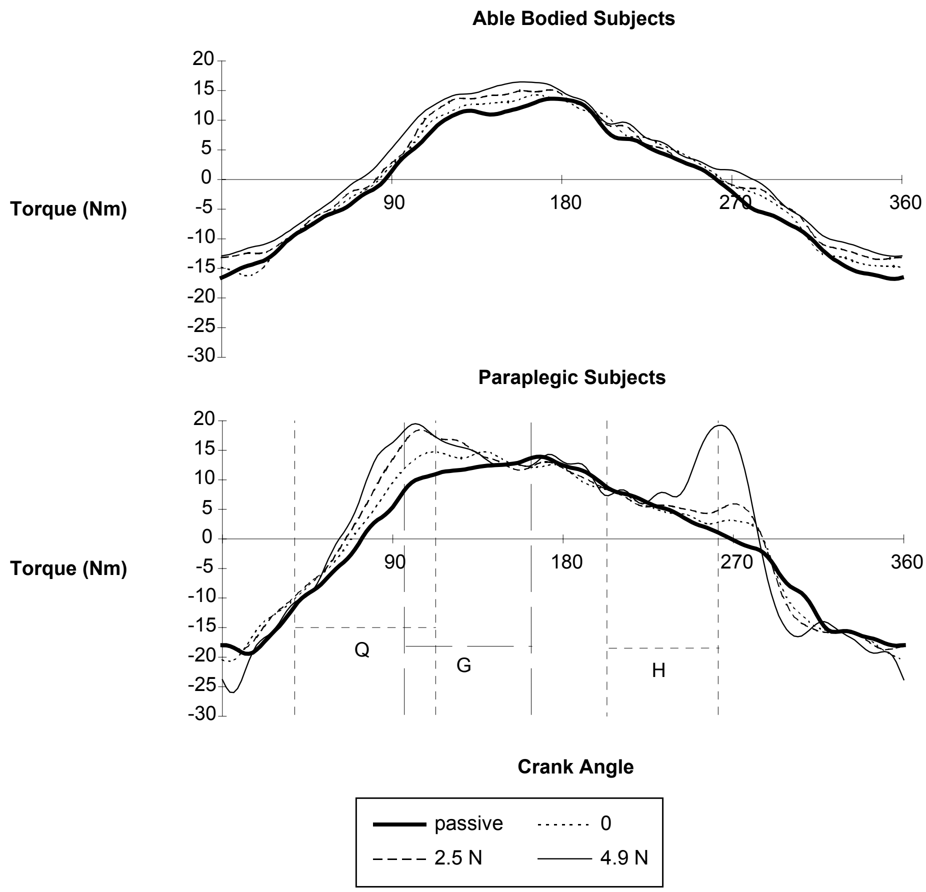

Figure 1.1.1 **Crank torques generated by SCI and able-bodied subjects cycling against four different resistances. From Sinclair et al. (1996), p54. Crank angles differ to those used in the present study, with TDC and BDC occurring at 90 and 270 deg respectively.**

It seems likely that increasing the range of crank angles over which stimulation is applied would decrease the peak torque required to achieve the same workload. The Ergys stimulates the quadriceps, hamstring and gluteal muscles for durations of 75, 59 and 67 deg respectively (Ergys-1 Service Manual, Theraputic Technologies Inc), and it is not obvious why such a narrow range of stimulation angles was chosen. One possibility is that different subjects may have different ranges of crank angles through which stimulating the muscles will provide positive propulsion. If this were so, then the range of angles suitable for all subjects would be much narrower than the available range for any specific subject (Figure 1.1.2).

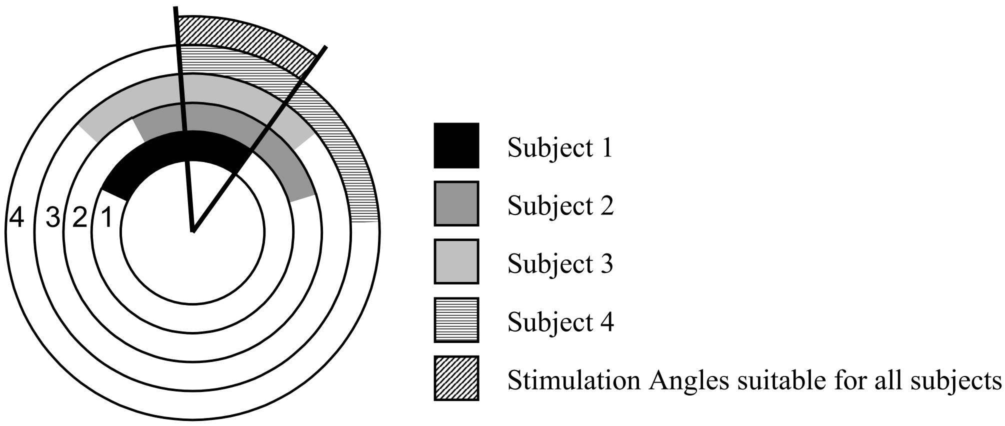

Figure 1.1.2 **Schematic illustrating that only a narrow range of stimulation firing angles may be suitable to provide a common pattern for all subjects.**

Different optimum stimulation patterns for each subject may arise from differences in anthropometric characteristics of individuals. During pilot testing for the study by Sinclair et al. (1996), it was difficult to standardise the range of knee angles experienced by each subject. If all subjects were positioned so that their legs reached the same maximum angle of knee extension, then they achieved very different peak knee flexion angles. This effect is illustrated by Figure 1.1.3. Even if two subjects had exactly the same leg lengths, differences in crural index (ratio between shank and thigh length) would affect the range of angles experienced by each subject. The original intent of this research was to investigate whether computer modelling of NMES cycling would provide a method to develop optimum stimulation patterns, based on the anthropometric characteristics of individual subjects.

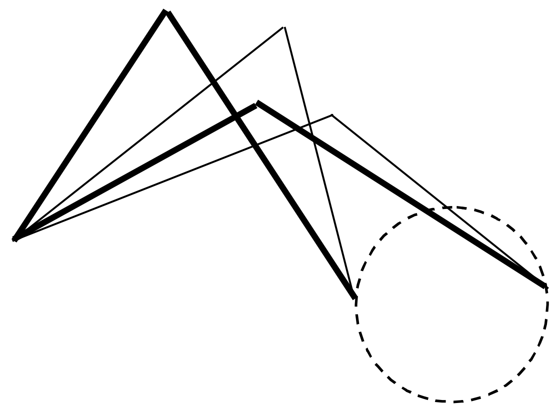

Figure 1.1.3 **Illustration of the effect of crural index on knee angles during cycling. The illustration shows two legs, both with the same overall length, but with differing crural indexes.**

Optimum stimulation patterns may also be affected by subject specific muscular characteristics such as fibre length, series elastic component length, moment arms and fibre type morphology. While this study will not measure these variables for individual subjects, the effect of differences will be discussed where they affect the ability to predict individual stimulation patterns.

Forward Dynamic Simulations of Human Movement

Forward (or direct) dynamic modelling involves the prediction of motor performance from muscle forces used as inputs to a model (van den Bogert, 1994). This is in contrast to inverse dynamic modelling where muscle forces are deduced from measurements taken after a movement has been completed. The advantage of forward dynamics is therefore that it can be applied to novel situations without the requirement for experimental measurements.

While forward dynamic modelling has been popular in recent years for predicting mechanical responses resulting from specified muscle recruitment patterns, the opportunities to experimentally confirm predictions have been limited by the difficulty in directly measuring outcomes (Herzog and Leonard, 1991). During voluntary contractions, complex interactions are present between muscles and, while modelling allows us to predict the force in various muscles, the opportunities to experimentally confirm these findings are limited. Recently, there have been attempts to confirm the predictions of forward dynamic modelling using experimentally measured forces from animals (eg Cole et al., 1996; Herzog and Leonard, 1991). Such invasive measures are not applicable for human subjects except for a few distal muscles with tendons suitable for the attachment of in-vivo transducers (eg Fukashiro et al., 1993; Schuind et al., 1992). Therefore, validation of models for human movement must be based on the measurement of external forces, rather than those developed within individual muscles.

Redfield and Hull (1986) have performed comparisons between measured and predicted pedal forces for able-bodied individuals. While their model's performance was generally quite good, it was not possible to identify the origins of errors because of the synergistic activation of many muscles at varying times during each revolution. Errors resulting from inappropriate partitioning of effort between muscles could not be separated from those caused by more fundamental assumptions of the model such as muscle moment arms, fibre lengths, etc. Cycling performed by SCI individuals using NMES, however, offers several advantages for the validation of a model. The muscle activation patterns are under direct "a priori" control during the experiment, so the timing of muscle activity is known for comparison with the model[1]. Furthermore, individual muscle groups can be stimulated without simultaneous activation of other groups, allowing external measurements to be confidently allocated to the effect of a specific muscle group. This study therefore provides a unique opportunity to evaluate the performance of a forward dynamic model by comparing direct measurements of pedal forces during cycling with those predicted by the model.

Gföhler and Lugner (2000) and Schutte (1992) have previously developed forward dynamic models of cycling for SCI individuals. Both studies predicted that wider stimulation firing angles would enhance performance beyond that provided by the standard Ergys stimulation angles. The validation process for Schutte (1992), however, consisted only of measuring the cycling cadence for a single subject using a variety of seat positions with constant stimulation parameters. There was no attempt to measure the model's ability to predict pedal forces or how changing stimulation timing would affect performance. Gföhler and Lugner (2000) provided no experimental measures to support their conclusions.

[1] In practice, the influence of spinal reflex arcs may modify muscle recruitment somewhat from the activation under direct control of electrical stimulation. Other muscles may be recruited or the activation of the target muscle may change in response to feedback from the stimulation of sensory nerves. This effect is assumed to be minor and is acknowledged to be a limitation of the present research design.

Significant changes occur in the structural and biochemical make-up of the muscles of SCI individuals (Round et al., 1993). These changes are known to change the performance of these muscles when compared to muscles of able-bodied individuals acting under voluntary control (Gerrits et al., 1999). Therefore, when adapting existing forward dynamic models of human muscle performance, it may be necessary to change model parameters such as the force-length relationship, the force-velocity relationship or the rate of force development as a muscle becomes activated. A preliminary component of this research will therefore be to measure performance of SCI individuals during isokinetic contractions in order to develop appropriate muscle model parameters.

1.2 Statement of the problem

The purpose of this study was to develop a forward dynamic model to predict cycling performance by SCI individuals using NMES. The initial component of the project consisted of measuring the performance during isometric and isokinetic knee extension contractions at various joint angles and velocities. These initial tests were then used to refine model parameters specifically for the SCI muscles being studied, as well as to test the model's performance under more controlled conditions. The second component of this research is to validate the model using an isokinetic cycle ergometer, adapted within this Faculty for use by SCI individuals. Validation of the model by comparing measured with predicted pedal forces during cycling is seen as the most important component of this research. Without such validation, it is impossible to have full confidence in predictions made by a model for performance under novel conditions.

1.3 Specific Objectives

The following steps were identified to accomplish the goals of this research.

1 Develop a model predicting knee extension torque generated in response to NMES of the quadriceps during isometric and isokinetic knee extension contractions. A number of specific objectives were targeted in the development of this knee extension model. These were to:

- a Investigate the effect of a normalised distribution of fibre and sarcomere lengths within a muscle on the whole muscle's force - length relationship
- b Determine suitable series elastic slack lengths for the model in order to best fit the quadriceps' isometric torque - angle relationship.
- c Test the suitability of quadriceps moment arm - joint angle relationships available from previously published research.
- d Model activation dynamics during the rise and relaxation of isometric torque in response to NMES onset and cessation.
- e Determine how activation dynamics alter in response to changes in knee angle and level of fatigue.
- f Determine whether visco-elastic resistance generates significant torque during passive movements for velocities of contraction relevant to ergometer cycling.
- g Determine constants predicting changes in torque with knee extension velocity using Hill's force - velocity equation.

2 Develop models for the hamstrings and gluteus maximus muscles, similar to that for the quadriceps. Rather than performing experimental tests, previously published research was used in the development of the hamstring and gluteus maximus models. In order to develop these models, it was necessary for series elastic slack lengths to be estimated by fitting the proposed model to isometric torque - angle curves available from the literature.

3 Measure cycling performance for SCI individuals using contractions of the quadriceps, hamstrings and gluteal muscles elicited by NMES. This testing was performed using a MOTOmed viva cycle ergometer, adapted "in-house" for use by SCI individuals using NMES. The following objectives were identified for this component of the study:

a Measure changes in activation dynamics and magnitude of torque applied to the ergometer crank by SCI individuals over a period of 5 min continuous cycling with NMES of the quadriceps muscles.

b Measure the external power output generated in response to changes in NMES timing for the quadriceps, hamstring and gluteal muscles for the purpose of identifying firing angles that maximised power output for each muscle group.

c Determine the effect of changing ergometer seat position on the measurement of peak NMES firing angles.

4 Test the ability of the model to predict the cycling performance of SCI individuals using NMES elicited contractions. Again, specific objectives were identified to provide this test.

a Test the model's ability to predict patterns of crank torque resulting from:

- fatigue over the course of 5 min continuous cycling,
- changes in NMES firing angles for each muscle group,
- changes in seat position between experimental sessions.

b Test the model's ability to predict stimulation firing angles that maximise acute power output. Test this predictive ability for:

- NMES applied separately to each muscle group
- the effect of changing seat position using a single cycle ergometer

1.4 Significance of the study

The present study makes an important contribution to the existing literature in two areas. To the field of musculo-skeletal modelling, this study provides a valuable opportunity to evaluate the ability of models to predict kinetic outcomes from known muscle activation patterns. Previous models have studied either single joint movements where single muscle groups can be assumed to be active, or multi-joint movements where a number of synergistic muscles act at different times. In the latter case, precise measurements of the activation of each muscle can only be inferred from EMG measurements, which may be difficult when muscles are close together or not adjacent to the skin. This has hampered the ability to evaluate such models because it is not clear whether model errors were the result of inaccurate measurement of muscle activation patterns or from errors within the model itself. NMES cycling provides an opportunity where multi-segment movements are occurring, yet the muscle activation patterns are simple and under direct experimental control (See Section 9.3 for discussion of the limitations of this assumption).

The second contribution of this thesis is to the field of NMES exercise for SCI individuals. A number of researchers have previously performed forward dynamic modelling of NMES exercise (eg. Riener et al., 1996; Yamaguchi and Zajac, 1990) including similar studies to the present thesis looking at cycling (Gföhler and Lugner, 2000; Schutte et al., 1993). These previous models, while contributing greatly to our understanding NMES exercise, have been limited in their degree of experimental validation. The present study provides direct comparisons between kinetic outcomes measured during cycling and those predicted by the model. With this information, we have a better understanding about the limitations of subsequent predictions made by the model.

1.5 Structure of the thesis

This thesis is arranged into nine chapters, briefly described below.

- Chapter 1 is this introduction.
- Chapter 2 contains a review of relevant literature. This review describes the present understanding of how NMES exercise may affect secondary complications arising from SCI. It then gives an overview of the development of forward dynamic models of human movement with a particular emphasis on the development of muscle models. The aim of this overview is to identify those modelling methods best suited to the present study.
- Chapter 3 shows the development of a multi-fibred muscle model with each fibre having a different length and number of sarcomeres in series. While not used in further modelling, this multi-fibred model illustrates the effect of fibre heterogeneity on whole muscle modelling.
- Chapter 4 outlines the methods used for this project. A forward dynamic model is developed, using sections from Chapter 2 to justify the choice of certain components. This chapter then goes on to describe the experimental methods used to identify appropriate model parameters and to test the performance of the model.
- Chapter 5 gives results from isokinetic knee extension experiments.
- Chapter 6 uses results from chapter 5 as well as a variety of literature sources to determine various parameters for use in the model.
- Chapter 7 gives results from the cycling experiments and compares model predictions with those measured experimentally.
- Chapter 8 gives an overview of the model's performance and discusses the implications of these findings.
- Chapter 9 reviews the original objectives of the study and provides statements regarding the accomplishment of these objectives. This chapter also provides recommendations for further study.
- Appendix 1 contains documents related to approval for this research from The University of Sydney Human Ethics Committee and is bound at the end of this thesis. Appendices 2 to 4 contain experimental data and may be found on the accompanying Compact Disk.

1.6 Definition of symbols

Kinematic and Kinetic Models.

$\dot{\theta}_c$	Crank angular velocity
$\ddot{\theta}_c$	Crank angular acceleration
x_t	Horizontal displacement of thigh centre of mass
$\ddot{x}_t$	Linear acceleration of the thigh centre of mass (horizontal component)
F_{hx}	Horizontal joint force acting across the hip
m_t	mass of thigh
r_t	distance from hip to thigh centre of mass
I_t	mass moment of inertia of thigh
M_h	Moment of force about the hip.

Muscle Model

qs[t]	shank angle	lforf	resting (optimal) fibre length
aas[t]	angular acceleration of shank	sltrf	slack length of tendon
asx[t]	linear acceleration of shank centre of mass (horizontal)	ktrf	tendon stiffness
dhk[t]	moment arm for the hamstring muscle at the knee	arf	Hill's constant a
dhh[t]	moment arm for the hamstring muscle at the hip	brf	Hill's constant b
Tris	rise time of activation	pennangorf	resting angle of pennation
Tfall	fall time of activation (time for muscle to lose force)	lbrf[t]	belly length
Amin	minimum muscle activation	pennangrf[t]	angle of pennation
qo[t]	whether quadriceps muscle is switched on or off	lfrf[t]	fibre length
qa[t]	activation level of quadriceps muscle	lf'rf[t]	fibre velocity
fforf	max isometric force at optimal fibre length	ffolrf[t]	max isometric force at given fibre length
eccrf	ratio of maximum eccentric force to maximum isometric force of a muscle	ftrf[t]	tendon force
		ftomrf[t]	force that tendon pulls on the fibre

All variables expressed as a function of time have the suffix [t]. Anything without this suffix is a constant. Variables ending with “rf” are applied to muscle rectus femoris. The same symbols may be used for muscles vastus, hamstrings or gluteal by replacing “rf” with respectively “v”, “h” or “g”.

Variables reported as abbreviations in Chapters 5 and 6.

Calculation of these variables is defined in Section 4.3.6

RT_{50}	Rise Time 50%
RR_{50}	Rise Rate 50%
FT_{50}	Fall Time 50%
FR_{50}	Fall Rate 50%

Other Abbreviations

EMG	Electromyography
NMES	Neuromuscular Electrical Stimulation
RMS	Root Mean Squared
rpm	revolutions per minute
SCI	Spinal Cord Injury
SD	Standard Deviation

Common Terms

Concentric Contraction	Condition where a muscle develops force and the length of the muscle between the origin and insertion decreases.
Eccentric Contraction	Condition where a muscle develops force and the length of the muscle between the origin and insertion increases.
Isometric Contraction	Condition where a muscle develops force, but the length of the muscle between the origin and insertion remains unchanged.
Isokinetic Cycling	Cycling at a constant angular velocity of the ergometer crank. This does not imply constant velocity of shortening of the muscles involved.
Muscle Fatigue	Condition characterised by a decline in force of a muscle over time. For the present study, no tests extended beyond five minutes duration, hence the physiological mechanisms of fatigue were limited by that duration.
NMES Firing Angles	The range of crank angles through which stimulation is applied.
Peak Firing Angles	NMES firing angles that result in maximum power output applied to the ergometer crank.
Tendon Slack Length	The resting length of a tendon when there is no force applied to the tendon.
Top Dead Centre (TDC)	Position where the right ergometer crank is pointing vertically upward.
Bottom Dead Centre	Position where the right ergometer crank is pointing vertically downward.