Chapter 9

Conclusions

The primary objective of this research was to develop a forward dynamic model to predict kinetic outcomes in response to NMES applied to the legs of people with SCI while cycling. Individuals differed in the range of movement at each joint while cycling, however, the crank angles at which each subjects' joints began to flex and extend were relatively constant. Consequently, the optimal NMES firing patterns for single joint muscles did not vary substantially between individuals. This result was found by analysis of the measured data as well as from computer simulations. Responses by the two joint hamstring muscles, however, were not constant for all individuals; with different patterns of forces applied to the pedals and consequent variations in the peak NMES firing angles. Individual modification of the hamstring moment arms about the knee enabled a much closer match between the measured and modelled pedal forces for each subject. Furthermore, modifying the knee moment arms to match a single trial enabled much better predictions of each subjects' performance across a range of NMES firing angles. It is therefore suggested that individual moment arm estimations are important for modelling the performance of two-joint muscles.

9.1 Achievement of Research Objectives

Specific objectives for the study were identified in Section 1.3. Below are statements about the level of achievement for each of those objectives.

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

Modelling a normalised distribution of fibre and sarcomere lengths widens the whole muscle force - length curve. The distributed fibre model predicts that greater forces would be generated at <u>very</u> short and long muscle lengths than could be explained by a force - length relationship scaled up from a single sarcomere. At lengths closer to optimal, however, the distributed fibre model did not greatly change the force expected from that predicted using a single sarcomere model. The assumption of normality means that, for lengths less than optimal, any increase in force achieved by having fibres shortening towards their optimal length was balanced by an equal number of fibres at lengths already less than optimal.

The present cycling model did not require muscle forces to be predicted over extreme ranges of muscle lengths. There was, therefore, insufficient benefit to be gained from the distributed fibre model to justify the extra computational cost.

b Determine suitable series elastic slack lengths for the model in order to best fit the quadriceps' isometric torque - angle relationship.

The isometric torque - angle relationship was best fitted using a rectus femoris slack length of 0.351 m. The slack length fitted to the combined vastii muscle was 0.156 m. The rectus femoris slack length was within 6% of the estimate provided by Hoy et al.

(1990), however the vastii length differed from Hoy et al. by more than 30%. It is suggested that this difference length of the vastii series elastic component arose from differences in either the segment lengths modelled and/or from differences in the calculation of length of the whole muscle-tendon complex.

c Test the suitability of quadriceps moment arm - joint angle relationships available from previously published research.

Simultaneously fitting both quadriceps tendon slack length and the joint angle - moment arm relationship did not result in moment arms that differed significantly from those published by Kellis and Baltzopoulos (1999). The moment arms measured by Kellis and Baltzopoulos were therefore used for all subsequent modelling of the quadriceps muscles.

d Model activation dynamics during the rise and relaxation of isometric torque in response to NMES onset and cessation.

Activation dynamics were modelled using a first order differential equation adapted from Pandy et al. (1990), modified slightly to include a delay between stimulation onset and the rise of muscle force. Fitting this equation to isometric contractions performed at a knee angle of 60 deg resulted in respective Rise Delay and Fall Delay constants of 53 and 66 ms. The time constants for Rise and Fall Times were both 48 ms.

e Determine how activation dynamics alter in response to changes in knee angle and level of fatigue.

Fall Delay and Fall Time remained constant for all knee angles. Although the torque generated at each knee angle was different, Fall Rate changed proportionally to torque resulting in the constant Fall Time. Both Rise Delay and Rise Time changed significantly with knee angle, with more extended positions requiring more time to generate torque. The changes with knee angle were greatest at the more extended positions, with little change for the more flexed knee angles experienced during cycling.

For this reason, the decision was taken not to incorporate a knee angle effect on the Rise constants utilised in the cycling model.

Rise Delay, Rise Time and Fall Delay all remained constant during the 5 min fatigue protocol while Torque declined to less than 25% of rested levels. There was an increase in Fall Time, however this increase was moderate compared to the magnitude of change in torque. Much smaller fatigue induced torque changes were experienced during the cycling experiments. Consequently, the decision was taken to maintain a constant Fall Time for the cycling model.

f *Determine whether visco-elastic resistance generates significant torque during passive movements for velocities of contraction relevant to ergometer cycling.*

Passive visco-elastic resistance was measured at knee extension velocities between 10 and 240 deg s^{-1}. Passive resistance was found to be constant for all velocities except 240 deg s^{-1}. Angular velocities of the knee were never found to be above 120 deg s^{-1} during the cycling experiments. Consequently, no visco-elastic effect was considered necessary for the cycling model.

g *Determine constants predicting changes in torque with knee extension velocity using Hill's force - velocity equation.*

Fitting the model's Hill constants to torques measured during isokinetic knee extension tests resulted in respective values for constants a and b of 0.351 × isometric force and 2.250 × fibre length. These constants were within 0.3 % of those used by Pierrynowski and Morrison (1985).

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.*

Fitting tendon slack lengths for the hamstring model to isometric hip extension data from published literature resulted in a length of 0.385 m. This was identical to that reported by Hoy et al. (1990). The present gluteus maximus slack length of 0.05 m was greater than the 0.001 m found by Hoy et al. The difference between studies may have resulted either from different methods for determining gluteal moment arm, or from a difference in the overall length of the muscle-tendon complex.

3 *Measure cycling performance for SCI individuals using contractions of the quadriceps, hamstrings and gluteal muscles elicited by 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.*

Average external power output dropped by 50% over periods of 5 min continuous cycling using NMES of the quadriceps muscles. While the peak torque declined significantly with fatigue, there was no apparent change in the *pattern* of torque application to the pedal when normalised against peak torque at each time period. There was no change in the time taken for crank torque to rise or fall to 50% of peak level in response to stimulation onset and cessation.

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.*

The experimental methods employed in this study were not successful in precisely determining NMES firing angles that maximised acute power output. Changes in power output in response to firing angle changes were not large in comparison to the effects of cumulative fatigue and inconsistent power output between trials, leading to large uncertainties in the measured peak firing angles.

Although caution must be exercised because of the high uncertainties involved, results suggest that peak NMES firing angles were relatively similar between subjects for the quadriceps muscles. Peak firing angles for the hamstrings, however, exhibited substantial differences between individuals. Results for the gluteals were variable, with some subjects not applying any measurable torque to the cranks, even with maximal stimulation applied. With fewer data points available the uncertainties for gluteal results were too great to draw conclusions regarding NMES timing without considering additional information from the model.

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

Only a small range of seat positions were available for each subject that enabled a reasonable range of movement at the hip and knee. Consequently, the seat positions during the final experimental sessions were only set back an average of 3.6 cm from the initial position.

There was little change in the peak stimulation firing angles for the quadriceps and gluteal muscles when the seat position was moved back. Hamstring muscles showed a variable response between subjects, with peak firing angles being advanced for some subjects and retarded for others when the seat was moved back.

Subsequent modelling suggested that the variation between individuals in the change of peak hamstring firing angles with seat position may be explained by variation in the relative hip – knee moment arms between individuals. Gluteus maximus and the vastii muscles, being single joint muscles, were not affected by the magnitude of their moment arms in determining peak 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,*

There was no measurable change in the pattern of torque production, other than the magnitude of peak torque, over the course of a 5 min period of cycling. Therefore, maintaining a constant model with simple adjustments to the maximum strength of the model's muscles should have maintained the model's predictive ability throughout the fatigue protocol. The correlation between measured and modelled crank torques did decline slightly over the course of 5 min cycling. It is suggested that this decline in predictive ability was the result of a decreased signal to noise ratio in the measured crank torques with fatigue.

- *changes in NMES firing angles for each muscle group,*

The model was able to satisfactorily predict changes in crank torque during the generation of muscle force in response to NMES of the quadriceps muscles at different crank angles. The model was able to satisfactorily predict the delay between stimulation onset and torque generation as well as the rise of crank torque up to maximum. When stimulation commenced early, both measured and modelled crank torques exhibited a period of negative torque with similar angles where the torque changed from negative to positive values.

The decline following peak crank torque was not well predicted for any muscle. Following a peak, the measured crank torque declined rapidly and was followed by an overshoot of negative torque. It is suggested that this error was due to a reduction in force exerted by the ergometer against the pedal, rather than from errors in the model's ability to predict muscle forces.

Results for hamstring muscles were quite variable between individuals. It appears that individual variation in hip to knee moment arm ratios may account for these differences. Modification of individual models' knee moment arms to match measured pedal forces based on a single trial improved the model's predictive ability across the full range of stimulation angles measured.

The model predicts the direction of initial torques developed by gluteus maximus, however it did not predict torques accurately throughout the range of movement. This appears to be a limitation of using a single fibre to model the gluteal muscles. The inability of a single fibred model to accurately predict hip extensor torques across a range of joint angles has been found within other models (Hoy et al., 1990). It is suggested that modelling of gluteus maximus could be improved by including a number of fibres, each with individual moment arms and fibre lengths.

- *changes in seat position between experimental sessions.*

Modelled responses to NMES of the quadriceps and gluteal muscles were very similar for both chair positions; matching the findings of experimental measurements. Again, like measured data, the modelled responses to hamstring stimulation differed between individuals once the knee moment arms were adjusted for each subject.

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

- *Each muscle group*

The simulation model was able to predict the NMES onset and cessation angles measured from experimental data to within the measurement uncertainty for every muscle. For the quadriceps, each subject produced similar responses to stimulation and the model agreed with this finding.

For gluteus maximus, uncertainty in the measured peak firing angles was very high, and consequently this could not be used for verification of the model. The model accurately matched the pattern of torque production immediately following NMES onset; predicting those stimulation angles that would initially produce negative crank torques. This gives confidence in the model's ability to predict peak onset angles because these are determined by the initial direction of torque production.

Results for the hamstring muscles were variable, with each subject producing different peak stimulation onset angles. The standard model failed to account for this inter-subject variability, most likely because of individual differences in the ratio of hip to knee moment arms. Modifying individual subjects' knee moment arms to match a single experimental trial enabled the model to more accurately predict individual variation in peak firing angles.

It was suggested in Chapter 1 that peak NMES firing angles would vary between individuals because of differences in anthropometric measurements. This hypothesis, however, was not supported by the present study. Anthropometric variations, although affecting the range of movement at each joint, did not greatly affect the crank angles at which each subject reached maximum flexion and extension of the hip and knee. Therefore, the range of crank angles through which single joint muscles shorten was consistent across individuals, with consequential consistency in firing angles. Inter-subject differences in peak firing angles were greater for the hamstring muscles, however it is suggested that these differences arose from variation in moment arm sizes, rather than simple anthropometrics like leg length. The standard model with consistent moment arms for all subjects predicted constant firing angles across subjects. Only when moment arms were adjusted for individuals was the measured variation in peak firing angles satisfactorily predicted by the model.

- *The effect of changing seat position for the current ergometer*

Modelled responses between the two seat positions were similar to measured responses. The quadriceps and gluteus maximus muscles produced similar peak firing angles at each chair position because, although the chosen chair positions affected the range of motion, they did not change the crank angles at which each joint commenced flexing and extending. Firing angles for the hamstring muscles however, did change with chair position; with both the magnitude and direction of change differing between individuals.

Fitting hamstring moment arms across the knee to individual subjects showed only a moderate correlation between the two sessions (r = 0.71). The difference between sessions suggests that the fitting procedure was not completely accurate in modifying individual moment arms and it is recommended that an experimental method be developed for this purpose. The difference between sessions does not definitively mean that the fitted moment arms were incorrect, because the ratio of knee to hip moment arms was determined by matching to experimental data, rather than directly measuring the magnitude of knee moment arms. Therefore, any change in hip range of motion between sessions (for example, from a different sitting posture) would have resulted in different hip moment arms for each session, thus requiring different knee moment arms to best fit the ratio of hip to knee moment arm.

9.2 Delimitations

- Muscle contractions for SCI individuals were elicited by percutaneous NMES via self adhesive electrodes. The contractions thus produced differ from voluntary muscle contractions in the order of recruitment of individual fibres and the rate of force development in response to activation. By delimiting the project to this type of contractions, this limits the extent to which the model can be extended to cover other types of movement.

- Only a single set of NMES parameters was examined within this study (mono-phasic square waves, 35 Hz frequency, 250 ms pulse width). Stimulation parameters are known to affect the muscles' response, particularly the amount of force generated, the rate of force rise and the rate of fatigue. The present model can not predict how muscle responses would alter with changes stimulation parameters.

- The model includes no physiological parameters that would predict a change in muscular performance with fatigue over time. The experimental procedures have therefore been delimited to acute periods of only 10 s cycling. While the model was used to predict NMES firing patterns that would maximise power output during an acute period, it is possible that different patterns may be needed to maximise the average power over an extended period of cycling.

- The present model includes only two dimensional measurements of kinematic and kinetic variables. While the ergometer bracing constrains the cycling movement to two dimensions and thus all propulsive movements can be considered, no understanding is gained of the lateral stresses involved. If, for example, it was desired to balance lateral moments between various muscles of the quadriceps group, a three dimensional model of the movement would be required.

- The present study investigates cycling only at a fixed cadence of 50 rpm. Cadences other than this would certainly affect the power output of cycling, but the effects would not be obvious. Much higher cadences are associated with maximum power output by voluntary cyclists (van Soest and Casius, 2000). With SCI individuals, however, it is not clear that the usual advantages of higher velocity muscle contractions would have as much benefit when compared to the decreased rest period associated with faster cadence.

9.3 Limitations of the study

- The current study modelled hamstring and gluteal muscles as if they were two individual muscles. The quadriceps were modelled as another two muscles, rectus femoris and the vastii, with equal activation levels for both muscles. While this modelling procedure reflects the degree of stimulation control available with most NMES cycle ergometers, it does not provide any understanding of the possibilities for differential activity between muscles in each group. Where different muscles within a group spend a different portion of the crank cycle shortening (eg if the moment arm differences are significant between individual hamstrings), it is possible that alternate stimulation strategies having differential timing between muscles within these groups could provide a more effective outcome.

- Each muscle was modelled assuming that all fibres within a muscle had the same length, the same pennation angle and shortened by exactly the same amount during whole muscle shortening. These assumptions, while simplifying the model, would have reduced the range of joint angles over which the muscles could generate significant torques. Chapter 3 demonstrates that the assumption of constant fibre length and pennation angle had little effect on the force – length relationship of a fusiform muscle like rat semimembranosus except at extreme muscle lengths. For muscles with a broad origin like gluteus maximus, however, this point is not so clear. It is possible that fibres within the muscle have different moment arms and hence would change length by differing amounts during hip extension. While there is insufficient anatomical information available to improve the model at this time, it seems likely that this explains the model's failure to match in-vivo joint torques reported in Section 6.2.2.

Similarly, modelling the vastii and hamstrings as two single muscles would have affected the patterns of joint torque - angle curves. Modelling these muscles separately would have enabled each muscle within the group to achieve optimum fibre length at a different joint angle, thus widening the torque - angle curve for each group. The model was not adapted in this way, however, because there would have been no way to decide the angle at which each individual muscle should generate peak torque. Assumptions about the angle of peak torque had to be made in order to set the tendon slack length for each muscle within the model.

- No allowance has been made in the present model for reflex arcs that may exist at the spinal level. Reflexes may change the activation of the target muscles or lead to the recruitment of additional muscles. Furthermore, muscle spasticity may occur for SCI individuals resulting in forces generated from contractions not under the control of stimulation. As explained by Schutte et al. (1993), the presence of spasticity is likely to be variable and is unlikely to occur in a predictable point within each pedal revolution. Therefore, such spasticity is an indeterminate signal that cannot be modelled satisfactorily. Reflex spasms did occur periodically during each experimental period, resulting in force measurements that differed dramatically from other trials. These periods were detected both by the subjects and from visual inspection of the data, and were eliminated from any subsequent analysis.

- There was substantial disagreement between crank torques predicted by the model and those measured experimentally during both the declining phase of active torque development and the subsequent passive phase. Section 7.2.2 discusses this in detail and concludes that changes in the measured torques are likely to have resulted from torques generated by the ergometer motor control system while attempting to maintain constant velocity, and/or from damped oscillations within the ergometer's elastic drive belt.

The cycling model was designed for use with an isokinetic ergometer. Changes in cycling cadence were recorded within each revolution, however, violating one of the modelling assumptions. The error resulting from violation of this assumption was quantified in Section 7.2.2 and found unlikely to be a major cause of disagreement between the measured and modelled crank torques.

- Chapter 6 developed parameters for the quadriceps model that were derived from experimental measurements of isometric and isokinetic knee extension contractions by SCI individuals. No isolated contractions were performed by the hamstring or gluteus maximus muscles for use in the development of those models. Rather, activation parameters measured from the quadriceps were used for all muscles within the model. This appeared not, however, to affect the model's ability to predict the dynamics of torque rise in response to the stimulation of these muscles. Previously published measurements of joint torques generated by voluntary contractions by able-bodied individuals were used to fit tendon slack lengths to the hamstring and gluteus maximus muscles. This may have contributed to errors in the model's ability to predict the patterns of torque generated by the hamstring and gluteal muscles across a range of crank angles.

- Only seven subjects completed the two experimental components of this research. Furthermore, for the cycling experiments, not all subjects' data were useable for all analyses. Small subject numbers are common in this type of research, where there are only a limited number of suitable subjects available who are willing to devote the necessary time for participation. With such small subject numbers, however, one must be cautious when extrapolating findings to the general population of SCI individuals. The cohort of subjects within this study were not specifically trained on NMES ergometers in preparation for this experiment. Consequently, it is possible that more highly trained individuals may have produced different magnitudes of power output variation in response to changes in NMES timing.

9.4 Areas for Further Research

A number of possibilities for further research have been brought to light during the course of this investigation. These include:

- Distribution of fibre and sarcomere lengths. Modelling a normal distribution of fibre and sarcomere lengths changed the whole muscle force-length relationship only at extreme muscle lengths. Further measurements of the statistical distribution of lengths within muscles would assist in the development of more realistic models.
- Effect of muscle velocity on rate of force development. Isokinetic contractions could be performed at different velocities using a controlled starting length for each contraction. This would provide statistical verification for the suggestion in Section 6.1.9, that the rate of activation rise is not dependent upon the velocity of contraction.
- Measurement of in-vivo hip and knee torques through a larger range of movement. The in-vivo data taken from literature reported hip and knee torque only through a 90 deg range. Recumbent cycling produces more flexed joint angles than this; creating difficulties in validating the model throughout the range of joint angles employed. It is recommended that torque angle curves be measured across a wider range of joint angles.
- Anatomical measurements of different fibres within the gluteus maximus muscle. In order to improve the gluteal model, a range of fibres should be used; each with a different length and/or moment arm pattern. Anatomical details are required before such a model could be developed.
- Measurement of model parameters for the hamstring and gluteal muscles. Chapters 5 and 6 employed measurements of quadriceps performance during isometric and isokinetic knee extension tests to validate parameters used within the muscle models. Measurement of these parameters in the hamstring and gluteal muscles is required to verify that the same parameters are suitable for these muscles.
- Measurement of relative hip and knee moment arms for the hamstrings. Section 7.3 hypothesised that differences between stimulation firing angles for individual subjects may be the result of differences in the ratio of hip to knee moment arm magnitudes. The design of an experiment to measure isometric hip and knee torques through a range of angles for each joint may enable these moment arms to be measured in-vivo.

- Prediction of crank torques during the declining phase after stimulation cessation. The present cycling experiments could be repeated using an ergometer that controlled crank velocity more precisely. This would determine whether the mismatch between measured and simulated data after the time of peak torque was an error inherent in the model, or caused by the ergometer control system.
- Stimulation timing to maximise power output over longer duration cycling periods. Cycling experiments could be repeated using a longer duration of activity for each firing angle. The purpose of this would be to investigate the rate of fatigue induced by changes in stimulation firing angles. This would overcome the present limitation of only measuring power outputs during an acute period of 10 s for each NMES firing angle.
- Modelling of physiological processes leading to fatigue. The current model predicts performance only under acute conditions, with no allowance for fatigue within a single simulation. Modelling of physiological processes is necessary to enable predictions regarding performance over an extended period where cumulative fatigue would be an important determinant of power output.
- Effect of cadence on power output. The present study investigates cycling only at a fixed cadence of 50 rpm. Cadences other than this would certainly affect the power output of cycling, but the effects would not be obvious. Much higher cadences are associated with maximum power output by voluntary cyclists (van Soest and Casius, 2000). SCI individuals, however, fatigue rapidly unless sufficient time is given between repeated contractions. The effect of pedalling cadence is therefore an area of interest for further study.
- Recruitment of ankle plantarflexors. All ergometers used to date by SCI individuals have employed a rigid ankle orthosis permitting no movement of the ankle. It is possible that changing the ankle orthosis to allow flexion/extension movements, together with stimulation of the triceps surae muscles, might enable increased work to be performed by SCI individuals.