Chapter 5

Knee extension experiment results and discussion

This chapter contains results from isolated knee extension experiments performed using a Biodex isokinetic dynamometer. General results are presented here to describe the resulting patterns of torque production. Data from this chapter will then be used in Chapter 6 for developing muscle model parameters specific to contractions by SCI individuals elicited using NMES.

5.1 Isometric Contractions

5.1.1 Effect of knee angle on muscle performance.

Knee extension torque varied as a near parabolic function of knee angle. While there was considerable variation in the maximum torque produced by each subject (Range 10.5 - 22.8 N m), the pattern of torque production was more consistent when normalised against the maximum torque generated by each subject (Figure 5.1.1.1). All subjects produced their maximum torque at either 45 or 60 deg of flexion.

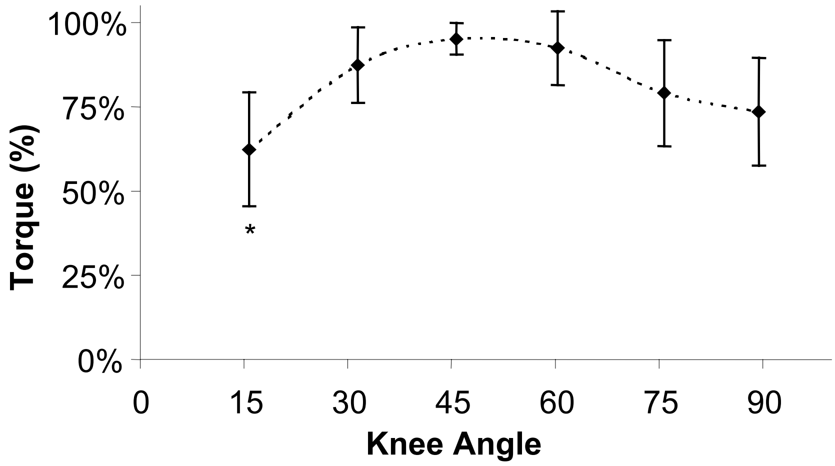

Figure 5.1.1.1 Effect of knee angle on NMES-induced isometric torque. Data are mean ± 1 Standard Deviation. * Only data from three subjects are included at 15 deg while all other data points represent the mean of seven subjects.

One subject was unable to extend the knee to an angle of 15 deg and current varied from the reference level for three of the remaining subjects; leaving only three data points for the 15 deg condition in Figures 5.1.1.1 and 5.1.1.2. The changes in stimulator current were unexpected, and only occurred at the 15 deg knee angle. This may possibly be explained by the stimulation electrodes moving closer together as the knee extended; thus reducing impedance between the electrodes and allowing a greater current for the same voltage.

Activation times and rates for both the rise and fall of torque were included in the analyses of regression. While the torque obviously took a greater time to reach 95% levels than 50%, both measures demonstrated the same changes with knee angle. The only difference was in fall times, and this minor effect will be discussed later. Because the 50% levels are more useful for modelling purposes (van Soest and Casius, 2001), further discussion will give results for 50% levels and it will be understood that 95% levels follow the same trends.

The time taken to generate active torque decreased as the knee became more flexed with Rise Delay decreasing on average from 67 ms at 15 deg to 31 ms at 90 deg (Figure 5.1.1.2). $RT_{50\%}$ produced a similar trend to Rise Delay (140 ms at 15 deg and 72 ms at 90 deg), however there was more variability in $RT_{50\%}$. Consequently, the regression equation was able to account for 63% of the variation in Rise Delay but only 35% of the variation in $RT_{50\%}$. Max Torque was not significantly correlated with either Rise Delay or $RT_{50\%}$. $RR_{50\%}$ however, varied both as a function of knee angle and of Max Torque (Table 5.1.1.1).

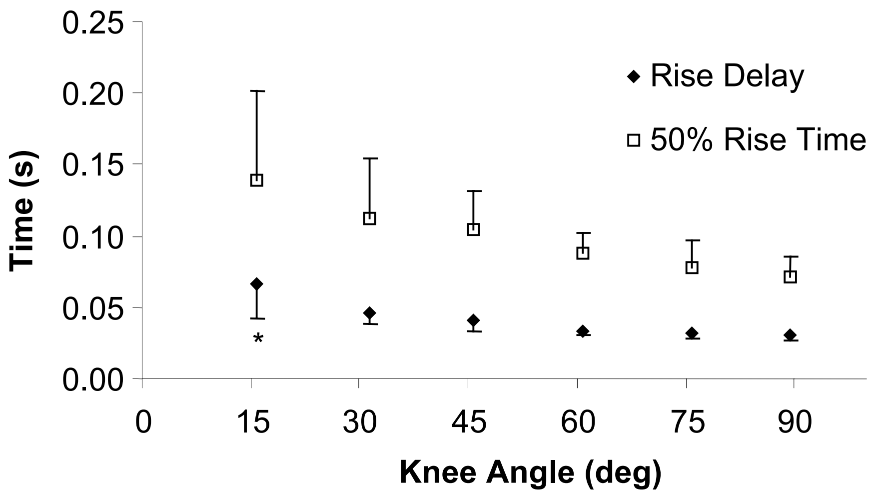

Figure 5.1.1.2 Effect of knee angle on Rise Delay and Rise Time 50%. Data are mean ± 1 Standard Deviation. * Only data from three subjects are included at 15 deg while all other data points represent the mean of seven subjects.

After the cessation of stimulation, Fall Delay and $FT_{50\%}$ remained constant with respective magnitudes of 58 ms and 60 ms (Figure 5.1.1.3) and had no significant correlation with any independent variable. $FR_{50\%}$ varied parabolically with knee angle and was also significantly associated with Max Torque (Table 5.1.1.1).

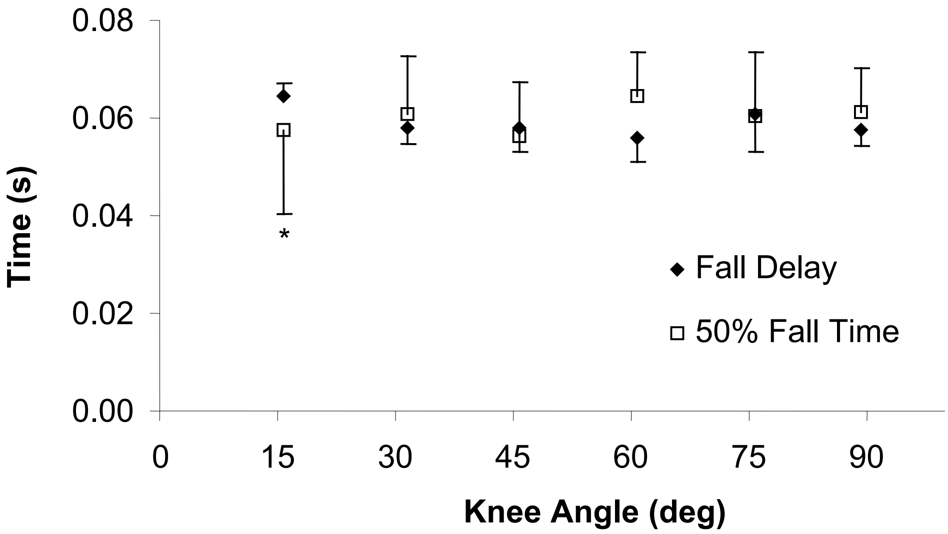

Figure 5.1.1.3 Effect of knee angle on Fall Delay and Fall Time 50% . Data are mean ± 1 Standard Deviation. * Only data from three subjects are included at 15 deg while all other data points represent the mean of seven subjects.

Table 5.1.1.1 Regression equations for each activation parameter.
Independent variables are listed in order of entry to the regression. The change in R^2 is given as each variable entered the regression analysis. Significance applies to the coefficient for each independent variable in the regression.

		Coefficient	R^2	Significance
Torque	Constant	-3.613		0.086
	Max Torque	0.75	0.67	0.000
	Angle^2	-2.39×10^{-3}	0.03	0.001
	Angle	0.243	0.07	0.003
	Total Regression		0.77	
Rise Delay	Constant	8.04×10^{-2}		0.000
	Angle	-1.30×10^{-3}	0.50	0.000
	Angle^2	8.40×10^{-6}	0.12	0.002
	Total Regression		0.63	
Rise Time 50% (RT_{50})	Constant	0.141		0.000
	Angle	-8.17×10^{-4}	0.35	0.000
	Total Regression		0.35	

		Coefficient	R^2	Significance
Rise Rate 50% (RR_{50})	Constant	5.161		0.650
	Max Torque	3.333	0.41	0.000
	Angle	0.263	0.07	0.037
	Total Regression		0.48	
Fall Delay *	Constant	0.058		
Fall Time 50% * (FT_{50})	Constant	0.060		
Fall Rate 50% (FR_{50})	Constant	-13.362		0.495
	Max Torque	5.522	0.55	0.000
	Angle²	1.937	0.06	0.003
	Angle	-1.99×10^{-2}	0.07	0.010
	Total Regression		0.68	

* No independent variables entered the regression equation for Fall Delay or Fall Time 50%. The mean of all results was therefore recorded as the constants.

5.1.2 Effect of fatigue on muscle performance.

Changes in muscle performance with fatigue were analysed using repeated measures ANOVA. Given the large number of variables and relatively few subjects, there were insufficient residual degrees of freedom to perform a multivariate ANOVA. Univariate results are therefore presented cautiously; mindful of the risk of type 1 errors being increased by the number of comparisons performed. Although 95% was chosen as the significance level for statistical comparisons, all variables deemed significant were found to have univariate p values less than or equal to 0.002. The risk of type 1 error therefore seems acceptably small, even considering the number of comparisons being performed.

There were variables missing for two subjects at the final time period; caused by stimulator current drifting away from initial levels. Two options were considered for replacing these missing values. The missing values could be replaced by an exponential fit from the individual's valid data. Alternatively, the missing values could have been replaced by the mean of all other subjects' data at that time. In practice, no differences were found in the statistical significance of whole group trends between the two methods. The exponential fit method was selected after visual inspection of the data because it produced a more consistent looking pattern for the individual subjects concerned.

Table 5.1.2.1 The effect of fatigue on knee extension torque parameters.

	Time[1]							
	Rested	0	1	2	3	4	5	Significance
Torque	14.9	10.0	8.0	5.5	4.6	4.4	3.6	0.000 #
(N m)	(3.5)	(3.0)	(2.2)	(1.4)	(1.1)	(1.2)	(0.8)	
Rise Delay	33	35	35	34	34	36	38	0.173
(ms)	(3)	(4)	(4)	(7)	(5)	(5)	(6)	
Rise Time 50%	86	87	85	95	87	92	83	0.749
(ms)	(16)	(20)	(28)	(59)	(51)	(72)	(55)	
Rise Rate 50%	87.7	58.4	47.7	33.4	30.2	29.8	26.7	0.000 #
(N m s^{-1})	(18.5)	(14.7)	(12.1)	(10.1)	(9.4)	(10.1)	(8.8)	
Fall Delay	58	60	62	65	63	66	69	0.520
(ms)	(5)	(7)	(13)	(9)	(7)	(9)	(19)	
Fall Time 50%	71	61	89	85	83	77	73	0.002 #
(ms)	(11)	(9)	(15)	(17)	(11)	(16)	(15)	
Fall Rate 50%	106.1	86.2	45.0	32.0	27.4	27.0	22.4	0.000 #
(N m s^{-1})	(29.5)	(29.2)	(7.4)	(11.1)	(8.9)	(9.1)	(7.7)	

Data represent mean and standard deviation

1 The rested values came from the first isometric trial from each session. The number of all other times refers to the time during the fatigue protocol.

\# ANOVA displayed significant effect for time ($P<0.05$).

Isometric torque declined significantly with fatigue with final values averaging less than 25% of rested levels (Figure 5.1.2.1). The rate of torque rise fell proportionally with peak torque so that the time taken to generate torque remained constant with fatigue. There was no significant change in either Rise Delay or Rise Time with fatigue. Similarly, Fall Delay did not change as the muscles fatigued; however, there was a significant change in Fall Time. Muscles took longer to relax as they fatigued, indicating that the change in relaxation rate was greater than the decline in torque levels produced.

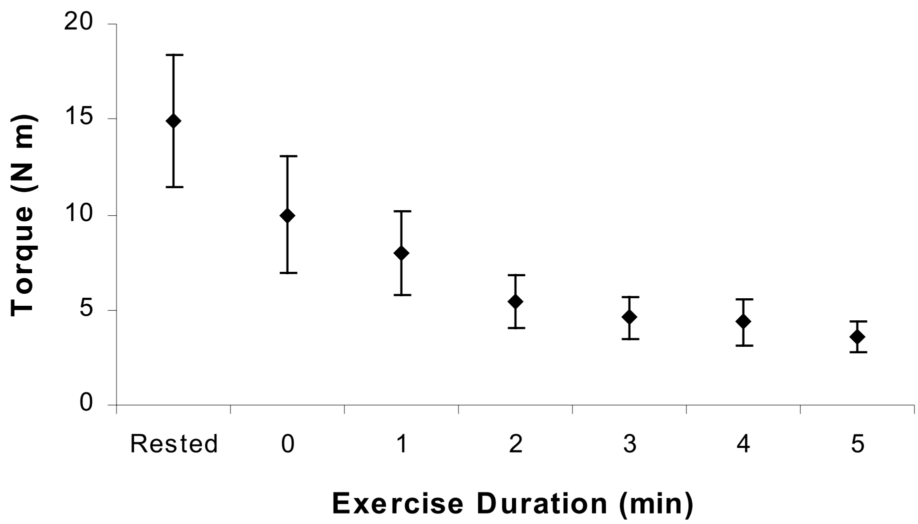

Figure 5.1.2.1 Change in Torque with fatigue.

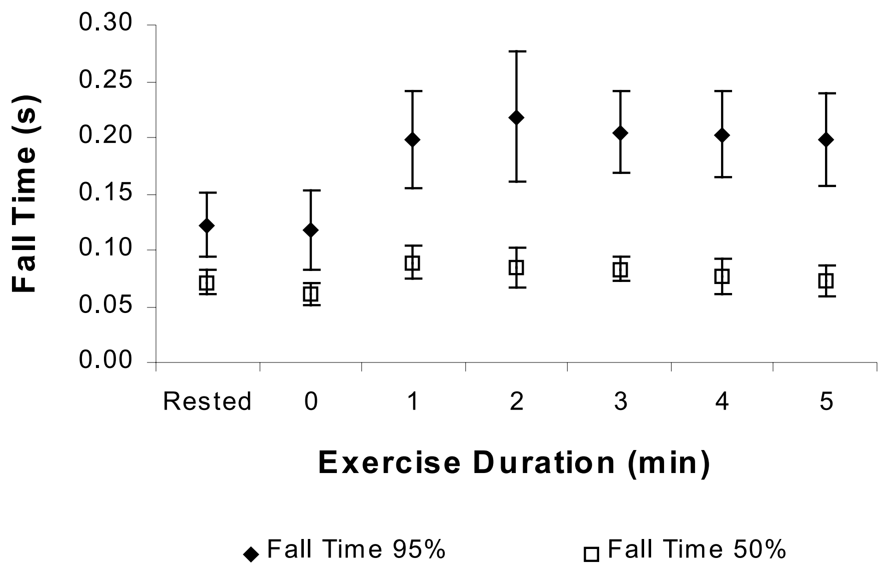

Figure 5.1.2.2 Change in Fall Time with fatigue.

5.1.3 Discussion

The primary aim of this study was to quantify activation rise and fall constants for use in forward dynamic simulations of electrically stimulated muscle contractions for SCI individuals. Rise and fall times were examined at a range of knee angles in order to investigate the effect of muscle length upon quadriceps recruitment time.

Individual responses to the stimulation varied dramatically with currents between 60 and 120 mA being required to elicit suitable contractions. Such wide variation in response to NMES stimulation meant that it was impossible to standardise the amount of current delivered to each subject whilst maintaining stable tetanic contractions. If currents were set too low, the muscle response was inconsistent; with various compartments of the muscle switching on and off. If the current was too high then subjects felt uncomfortable with the level of torque; perhaps fearing a risk of injury. Currents could not be standardised to generate a specific level of torque for the same reason. Torques low enough to suit some subjects did not result in stable contractions in the stronger subjects. Maximum torque could not be estimated owing to a potential risk of injury if large muscle forces were generated (Hartkopp et al., 1998); and consequently, the percentage of maximum was not an available option for standardisation.

Because the level of activation was not standardised between subjects, it was not possible to decide conclusively whether between subject differences were due to differing muscle performance, or whether stimulation current introduced variation. While this question cannot be resolved from the current data, Lieber and Kelly (1991) reported that subjects differed substantially in the amount of current required to activate their muscles; independent of electrode type, current and skin impedance. Increasing current certainly increases the torque output of an individual subject, but differences in current between subjects is less important to torque production than differences in the subjects' "intrinsic ability to be activated" (Lieber and Kelly, 1991, p715).

Max Torque was initially included as a predictor in the Linear Regression to normalise the torque curves and reduce the effect of large inter-subject differences that may have masked within-subject trends. Max Torque was kept as a predictor for all other variables to determine whether the strength of each subject affected those variables independently of knee angle. For example, it may have been anticipated that the time to develop torque would be greater in those subjects where the torque had a greater amount to rise. In practice, however, Max Torque correlated with the *rate* of torque rise and fall; hence the *time* to develop torque was unaffected by the level of torque produced by each subject.

When modelling muscle activations, there is often a point in time where muscle activation after that point would improve performance, while earlier activation would be detrimental (see Section 4.4.3 for explanation). It is impossible for a muscle to switch on instantaneously at the desired point; hence, the decision may be taken to reach 50% activation at this time. This balances the loss of performance from not being fully activated at this time against the loss from being partially active too early during the detrimental phase. For this reason, 50% rise and fall times have been utilised within this study. The 95% values showed similar trends but are not shown here for brevity.

All subjects demonstrated the trend of rise times varying with knee angle as illustrated by Figure 5.1.1.2, however, there was considerable variation between individuals in the time taken for torque to rise. There was no apparent pattern to explain this variation in either the maximum torque produced or amount of current required to induce contractions. High variability in SCI individuals may be expected to arise from individual differences in factors such as fibre type, degree of atrophy and morphology. Because of this individual variation, predictions of muscle activation patterns based on average values can be expected to be incorrect by up to 50%, particularly at more extended knee angles. If generic stimulation patterns are used across individuals without determination of individual muscle contractile properties, the patterns must be able to cope with substantial individual variations in the time to develop force. At present, however, there is no way to predict individual variation in time constants other than by direct measurement.

The time for torque to rise varied as a function of knee angle with more extended angles recording both greater Rise Delay and Rise Time. The increase in delay with knee extension can be explained by increased time to take up slack in the muscle's series elastic component. The change in Rise Time is most likely the result of differences in torque produced at each knee angle, although there was no tendency for Rise Time to decrease at the most extended knee angles. There may also have been some effect from changed compliance of the subject/dynamometer interface with knee angle as, at the more extended positions, the leg hung from the dynamometer strap with active torques being less than the weight of the leg and dynamometer bar. The compliance of the system was reduced by removing any padding between the subject and strap, however there may still have been some effect present in these results.

NMES has been used clinically with SCI individuals for lower limb tasks such as rising from a chair and standing with support (Hjeltnes and Lannem, 1990). Onset of muscle activation for these tasks occurs at very different knee angles, which will therefore result in changes in the activation timing constants. The regression equations presented in Table 5.1.1.1 predict a combined Rise Delay and $RT_{50\%\ to}$ be 221 ms at full knee extension but only 99 ms with the knee flexed to 90 deg. The increased delay time must be accounted for when predicting muscle response times during standing or else total collapse of the knee may result. Clearly, different activation constants should be used when modelling a wide variety of activities.

Combined Fall Delay and $FT_{50\%}$ averaged 118 ms with no variation between knee angles. As the muscles' series elastic elements are stretched at the time of stimulation cessation, the series elastic element would not be expected to affect Fall Delay at different knee angles. The same time constants could therefore be used to model muscle relaxation across a wide range of muscle lengths.

Comparisons between these findings and other reports are difficult to draw. The values found here cannot be compared with those of Gerrits et al. (1999) because that study reported only normalised maximum rate of rise, not the actual time. The present study found Rise Delays at 30 and 45 deg that were comparable to those measured by Krajl and Grobelnik (1973); with 15 deg producing a longer delay and the delay between angles 60 and 90 deg slightly shorter. Unfortunately, the knee angles used by Krajl and Grobelnik were not reported. Krajl and Grobelnik used a criterion of 10% peak torque to define their onset of torque development. This would have increased Rise Delay had the same criteria been used in the present study. Therefore, it could be speculated that these authors most likely used angles between 60 and 90 deg when measuring Rise Delay.

The current research was performed only with knee extensor muscles at different knee angles. While it can be hypothesised that the changes in activation timing were due to length changes in the muscles, further research is required before this can be confidently applied to other muscles. It is possible that mechanical changes within the knee joint (eg moment arm, patella position, etc) are partially responsible for the observed timing changes. Repeating this experiment with hamstring muscles, varying length by changing both the hip and knee angles, could partition muscle length from any joint angle effects.

The decline in torque levels with fatigue was similar in timing to values reported by Gerrits et al. (1999) for isometric quadriceps contractions by SCI subjects. While Gerrits' data demonstrated a similar plateau in fatigue occurring after approximately two minutes of exercise, their subjects stabilised at approximately 40% of maximum torque values, rather than the 25% found for the present study. A number of factors are known to affect amount of fatigue for SCI individuals exercising with NMES. Gerrits et al. used a 20 Hz stimulation frequency while the present study used 35 Hz. Lower frequency reduces the rate of fatigue (Katz et al., 1987) and hence is the most likely explanation for the higher plateau found by Gerrits et al. The current study performed the fatigue protocol after performance of the isometric tests at each joint angle (8 sets of three contractions with five minutes rest between sets). While this protocol was chosen to minimise fatigue, Figure 5.1.2.1 demonstrates that torque was already reduced at the beginning of the fatigue protocol. This may have affected the resulting plateau level. There were a number of other differences between stimulation protocols used in the two studies. Gerrits et al. used a shorter duty cycle (1 s on, 1 s rest), shorter pulse width (200 ms), and possibly a higher stimulation intensity. These factors do not explain the higher plateau level found by Gerrits et al. because all of these factors are more likely to increase the rate of fatigue (Mizrahi, 1997). Stimulation frequency therefore seems the most likely explanation for differences.

The time taken for torque to decline to 5% of the torque measured at stimulation cessation increased by approximately 60% over the course of the fatigue trials. An increase in relaxation time with fatigue was also found by Gerrits et al. (1999) for quadriceps contractions; and by Cameron and Calancie (1995) for isometric contractions of the wrist muscles of quadriplegic subjects. These two studies reported much greater changes in relaxation time, with fatigue inducing increases of more than 200% for both studies. It is not clear why the present study found less change, particularly given that the magnitude of fatigue induced torque changes were greater for the present study. Relaxation time shows a similar trend to torque in that the changes are most rapid at the beginning on an exercise period, with levels stabilising after approximately two minutes. Cameron and Calancie (1995) reported a similar finding, however Gerrits et al. (1999) only reported initial and final values.

Changes in the rate of force rise with fatigue were not reported for SCI subjects by Gerrits et al. (1999). Cameron and Calancie (1995) found that time to 50% maximum force rose from approximately 200 to 240 ms after four minutes of isometric exercise. Similar changes were found in the present study, however the trend was not statistically significant. Cameron and Calancie used only a simple T test between the first and last time period, perhaps making their analysis less conservative. Like the present study, Sahlin and Seger (1995) found no change in the rise time for able-bodied subjects performing isometric contractions in response to electrical stimulation. It therefore appears that, if rise time does increase with fatigue, the effects are relatively small compared with the time taken for force to decline after the cessation of stimulation.

5.1.4 Summary

The time for torque to decline after stimulation cessation was not affected by knee angle, however activation times for stimulation onset were greater at more extended knee angles. For this reason, care should be taken when modelling activities at different muscle lengths that activation time constants are modified appropriately. It would be inappropriate to use activation constants for activities like walking, where the knee is almost fully extended, if these constants have been measured for contractions performed with a more flexed knee. The regression equations provided by this study provide a means whereby activation times for the quadriceps muscles can be estimated for specific knee angles.

The cycling experiments described in Chapter 7 vary the crank angle where stimulation commences. The knee angles associated with these stimulation onsets, while varying, were always flexed more than 70 deg. Figure 5.1.1.2 suggests that the changes in activation times are not great for more flexed knee angles. For this reason, the effect of joint angle on activation will not be included within the present cycling model. This point will be discussed again in Section 7.2.3 to determine the effect of this simplification on the model's ability to predict responses to stimulation at different crank angles.

Progressive fatigue throughout the series of experiments at each knee angle, followed by a five minute period of repeated stimulation, caused isometric torque to decline to approximately 25% of its rested value. Despite such large changes in muscle force, there were only relatively small changes in the dynamics of each contraction. The time for torque to fall to 5% of active torque following the cessation of stimulation increased by 60% with fatigue. This change took place during the experiments at each angle and during the first minute of the fatigue trial. There was no subsequent change in relaxation time following this first minute. Time for torque to rise after stimulation onset, rise delay and fall delay were all unaffected by fatigue.

5.2 Isokinetic Contractions

5.2.1 Effect of velocity on passive torque

The Biodex dynamometer utilised a motor to perform passive extensions of the knee against gravity. Subtracting the effects of gravity enables the measurement of passive resistance about the knee joint at different velocities of movement. As can be seen in Figure 5.2.1.1, passive resistance was small and relatively constant for velocities at or below 120 deg s^{-1}. Repeated measures ANOVA found the effect of velocity to be significant when using the Greenhouse-Geisser adjustment for violation of the assumption of sphericity ($p=0.006$). Tests of Within-Subjects Contrasts found only the highest velocity (240 deg s^{-1}) to produce a passive torque different to that at 10 deg s^{-1}.

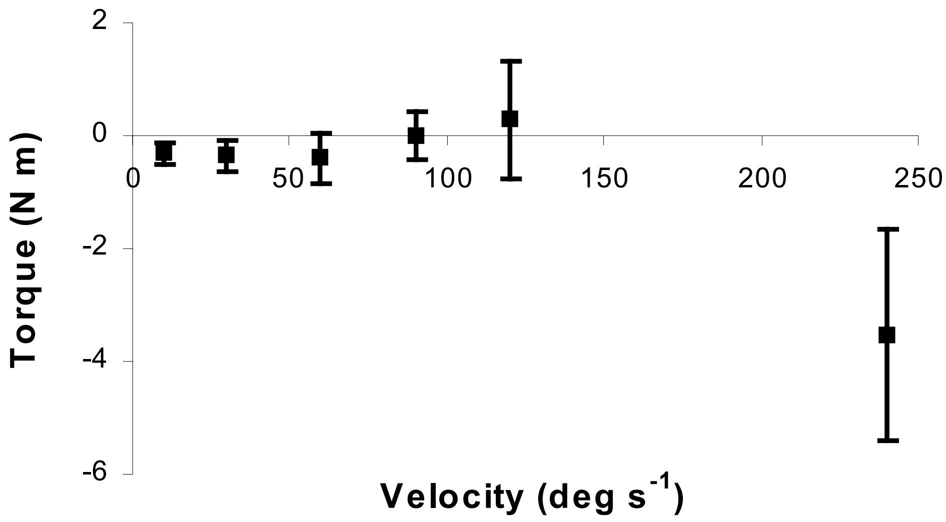

Figure 5.2.1.1 Effect of velocity on Passive Torque

The cycling experiments reported in Chapter 7 do not produce knee angular velocities in excess of 120 deg s^{-1}. The small passive torques recorded below this velocity supports the omission of passive viscoelastic joint torques from the current cycling model.

5.2.2 Effect of velocity on active torque

Active torque declined significantly with increased velocity (Figure 5.2.2.1, $p<0.001$) as would be expected from the force-velocity relationship of a muscle (Zajac, 1989). Data from these isokinetic tests will be used in Chapter 6 to fit constants to the model using Hill's hyperbolic force - velocity equation.

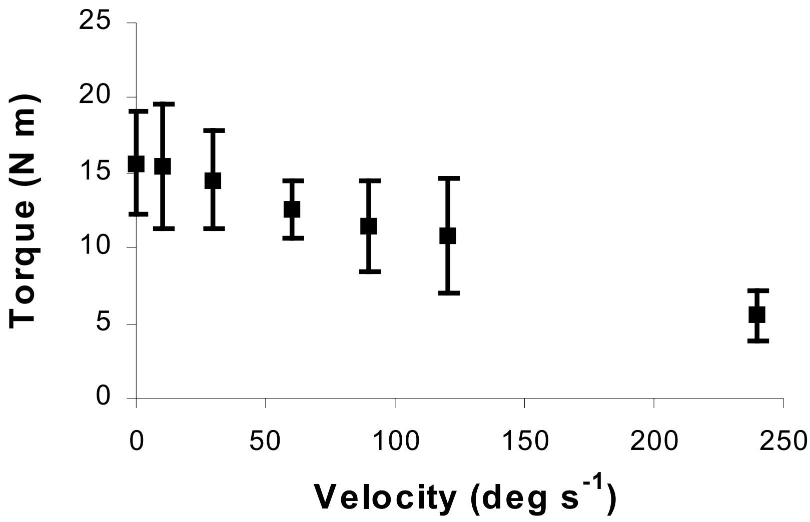

Figure 5.2.2.1 Effect of velocity on Nett Torque