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Musculo-skeletal modelling of NMES-evoked knee extension in spinal cord injury

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Abstract

The objective of this study was to explore relationships among constants used in musculo-skeletal models predicting torque generated about the knee by the quadriceps muscles. A model was developed and matched to data collected from individuals with spinal cord injury performing quadriceps contractions evoked using neuromuscular electrical stimulation. After fitting tendon slack lengths to the quadriceps muscles, the model was able to accurately match experimentally measured knee extension torques using previously reported values for the moment arm–knee angle and force–velocity relationships. Fitting new constants to these relationships did not improve the match between measured and modelled knee extension torques. There was significant interaction between variables used within the model. Using a narrower active force–length relationship for the muscles required the model to have smaller moment arms about the knee to accurately match measured torque across the full range of motion. Reduced moment arms, however, lowered the model's linear velocity of muscle shortening for each angular velocity of the knee, requiring different constants within the force–velocity relationship to predict the appropriate amount of torque decline. The present study demonstrates that, when a model does not fit the observed data, it is difficult to determine exactly which components are responsible because of the interdependent nature of parameters.

Keywords: Muscle modelling; Hill's model; Spinal cord injury; Neuromuscular electrical stimulation; Knee

1. Introduction

Muscle actions in multi-segmental movements, particularly of those spanning more than a single joint, are complex and not intuitive. In recent years, musculoskeletal models have been employed with increasing frequency to enhance our understanding of the intersegmental dynamics inherent in motor control (Zajac, 2002). Neuromuscular skeletal models have been able to predict in vitro forces generated by electrically stimulated animal muscles through a range of length and velocity conditions (Krylow and Sandercock, 1997). Such models have also been applied to human movement, particularly through the use of EMG to determine muscle activation and predict load sharing arrangements between muscles (e.g. Lloyd and Besier, 2003; McGill and Norman, 1986). Modelling voluntary contractions always has the requirement that activation patterns between synergistic muscles be predicted from some optimising routine (e.g. Chadwick and Nicol, 2000) or from EMG measurements (e.g. Lloyd and Besier, 2003). These studies have emphasised the importance of appropriate physiological variables with the models such as tendon slack length, muscle moment arms and fibre lengths. The present study explores the relationship between these variables when used to predict joint moments using musculo-skeletal models.

Complete spinal cord injury (SCI) prevents the transmission of motor drive from the brain to the motor nerves. For individuals with a damaged spinal

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cord, but intact motor nerves, muscle contractions can be evoked through the use of neuromuscular electrical stimulation (NMES). NMES-evoked exercise of the leg muscles has been proposed as a partial solution to many of the secondary complications arising from SCI (Davis, 1993). Modelling has been employed to develop effective muscle control strategies for NMES-evoked exercise (Gföhler and Lugner, 2000; Yamaguchi and Zajac, 1990) and offers a number of particular advantages over direct experimentation. Firstly, there are often relatively few SCI individuals available for experimental procedures requiring trained subjects. Furthermore, the subjects available will vary in their level of lesion and completeness of injury, further exacerbating the problem of finding a relatively homogeneous sample. Computer simulations enable researchers to advance their understanding of NMES control systems, while minimising the demands made on SCI subjects.

As well as providing potential benefits for individuals with SCI, NMES provides an excellent form of exercise for validating the predictions made by musculo-skeletal models. NMES-evoked muscle contractions are under direct control of the researcher because there are no neural pathways descending from the brain after complete SCI. Therefore, muscle activation patterns can be kept simple, without the complex interactions between synergistic muscles that may be present during voluntary contractions.

Musculo-skeletal models predicting performance of an entire muscle may be based on simplified equations that accurately predict performance characteristics, or on more complex series of equations derived from an understanding of the underlying anatomical structure of muscles. For example, Perumal et al. (2002) used a single quadratic equation to characterise the change in quadriceps torque with knee angle. By contrast, Meijer et al. (1998) used a quadratic function to characterise the variation in single fibre force with fibre length, a helical model for fibre length as a function of muscle length, another quadratic function to give muscle length changes with joint angle and repeated these equations for each of the four quadricep muscles. The latter method, although more numerically complex, has the potential to provide a greater understanding of the underlying mechanisms determining muscle performance.

An anatomically based muscle model requires a large number of constants to be assumed for each component equation within the model (e.g. fibre lengths, muscle moment arms, Hill constants, etc.). While many of these constants have been examined under a variety of conditions, they have not been validated for NMES-evoked muscle contractions in SCI. Other constants such as tendon slack length are very dependent upon other aspects of the model and have not been extensively studied to date (Out et al., 1996).

The purpose of this study was to develop suitable constants for use in a model of the quadriceps muscles from individuals with SCI. A primary focus of the study was to investigate interactions between parameters within the model. For example, changing the moment arm of a muscle model affects both the linear velocity of shortening and the range of length changes for that muscle. Selection of moment arms for a model will therefore affect predictions of both the decline in muscle torque with increased joint angular velocity and the change in torque with joint angle. The present study quantifies the magnitude of some of these interactions and implications for the choice of modelled parameter constants will be discussed.

2. Model details

A “Hill”-type musculo-skeletal model was developed to predict performance of muscles contracting under NMES-evoked recruitment. Fig. 1 shows that muscle–tendon kinematics and muscle activation are independent inputs to the model that determine the force of contraction. For the present study, activation was assumed to be constant, leaving only inputs from muscle–tendon length and velocity. There is a feedback loop whereby fibre length is determined by tendon length as well as whole muscle length; however, feedback from tendon force to whole muscle kinematics was not used in the present study. All modelling was

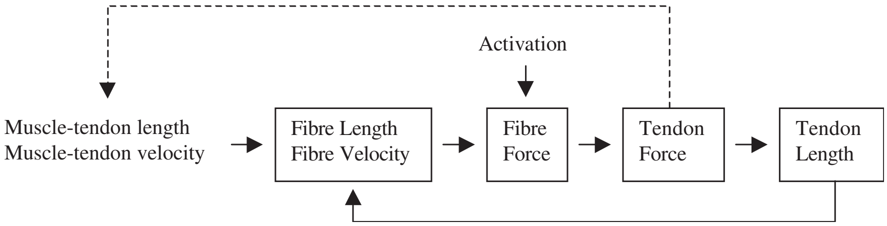

Fig. 1. The conceptual muscle model. This illustration shows the two independent inputs to the model: muscle activation and muscle–tendon length changes (determined by segmental kinematics). Feedback from the tendon force to the length of the muscle and tendon is shown as a dotted line because, although this would be present in unconstrained movements, the present study uses only isokinetic contractions that prevent muscle forces from affecting joint kinematics.

performed using Mathematica 3.0 (Wolfram Research, Champaign, IL). Full details of the model equations are provided in Appendix A.

3. Methods

Seven SCI subjects (one female, six males) with lesions between vertebrae T4 and T9 were recruited for this study (six classified ASIA-A, one ASIA-B; Ditunno et al., 1994). All participants underwent informed consent procedures according to the guidelines of the University of Sydney Human Ethics Committee and received medical screening for pre-existing musculoskeletal or cardiorespiratory disorders before participating in the study. The subjects had previously trained using NMES, however not all were in regular training at the time of the study.

Experimental measurements were taken using a Biodex Isokinetic Dynamometer (Biodex Medical Systems, New York, USA). Subjects were strapped firmly into the Biodex chair with the seat back positioned at 90° to the seat cushion. The usual padding between the shank and the dynamometer arm was discarded in order to reduce compliance between the leg and the dynamometer. Muscle contractions were evoked using a laboratory-designed stimulator delivering monophasic square wave pulses at 35 Hz with a pulse width of 250 μs. Each subject received enough current (between 60 and 120 mA) to produce strong stable tetanic contractions via 8 × 10 cm self-adhesive electrodes (Medtronic, Cat No. 86906350) placed over the belly of the quadriceps muscles. Voltages proportional to the torque and angle of the dynamometer and stimulation current were collected at a frequency of 1000 Hz using an IBM compatible personal computer.

Subjects presented for two experimental sessions for either isokinetic or isometric exercise. Each session was preceded by 5 min “loosen-up” consisting of passive leg extension and flexion at a rate of $60^\circ\,\mathrm{s}^{-1}$. The amplitude of NMES current was initially set at zero, then increased gradually until strong, stable, tetanic contractions of the quadriceps were produced. Current amplitude could not be equalised between subjects because of the great difference in stimulation sensitivity between subjects.

Isometric contractions were performed at randomly ordered knee angles between 15° and 90° of flexion using a duty cycle of 2 s stimulation followed by 2 s recovery. Three contractions were recorded from each trial with knee angle, stimulation current and torque averaged across the active periods of each contraction. Both active (stimulated) and passive trials were collected at each angle in order to calculate nett active torque by subtracting passive torque from the active trials. The present study did not consider any parallel elastic component within the muscles that could generate moments about the knee at long muscle lengths. It is likely that there was some torque exerted by passive stretching of the hamstrings at the 15° angle for at least some subjects. This would have no influence on the active torque measurements as only the nett active torque is reported for isometric trials. For dynamic trials, a knee angle of 60° was chosen for all measurements because the contribution of elastic components would be minimal at this angle.

Isokinetic contractions were conducted on a different day to the isometric session. Three contractions were performed at six different velocities between 10 and $240^\circ\,\mathrm{s}^{-1}$, while stimulation was triggered manually by the experimenter to produce concentric extension contractions. Non-stimulated trials were also conducted at each velocity to assess passive resistance to movement.

A reference angle of 60° was chosen for analysis of each trial. At each velocity, passive joint torque at the reference angle was calculated by subtracting weight torque at that angle from the torque measured during passive trials. Nett active torque was calculated by subtracting passive torque from the measured active torque with contractions being eliminated if stimulation did not commence at least 300 ms prior to the joint passing through 60° (Carroll et al., 1989). The stimulator was designed to produce a constant voltage output, and therefore variations in current were possible if impedance between the stimulation electrodes changed during the experiment. Trials with currents later found to deviate more than 7% from reference values were discarded from the analysis. The average absolute deviation in current of all trials from the reference values was 1.4%.

Changes in active and passive torques with joint angular velocity were assessed from the dynamic trials using repeated measures ANOVA with velocity as the independent variable. SPSS (10.0) was used for all statistical comparisons and the 0.05 level adopted for statistical significance. For those cases violating the assumption of equal variance between cells, significance was established using the Greenhouse–Geisser adjustment (Vincent, 1995).

Tendon slack lengths for the model were adjusted to ensure that maximum knee extension torque occurred at an appropriate knee angle. A numerical function was generated within Mathematica to evaluate the RMS error between experimentally measured and modelled isometric knee extension torques at each joint angle, as functions of the vastus and rectus femoris slack lengths. The FindMinimum function within Mathematica was used to identify slack lengths that minimised the RMS error value using adaptations of Brent’s method (Wolfram, 1996).

The fitting process was subsequently used to determine whether adjustments to the quadriceps moment arm–knee angle relationship could further improve the fit between measured and modelled knee extension torques. To assess whether model changes provided a better match to experimental data, paired *t*-tests were performed using the squared differences between all measured and modelled isometric torque values. This process was repeated using isokinetic contractions to find values for constants *a* and *b* in Hill's force–velocity relationship (Stern, 1974). Finally, the relationship between the chosen model constants was explored, with examples used to explore the interaction between the quadriceps muscles' force–length relationship, moment arms about the knee and force–velocity relationship.

4. Experimental results

Knee extension torque varied as a near parabolic function of knee angle (Fig. 3). 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. All subjects produced their maximum torque at either 45° or 60° of knee flexion.

The Biodex dynamometer utilised a motor to perform passive extensions of the knee against gravity. Subtracting the effects of gravity enabled the measurement of passive resistance about the knee joint at different velocities of movement. As can be seen in Fig. 2, passive resistance was small and relatively constant for velocities up to $120^\circ\,\mathrm{s}^{-1}$. Repeated measures ANOVA found the effect of velocity to be significant ($p = 0.006$); however, tests of within-subjects contrasts found only the highest velocity ($240^\circ\,\mathrm{s}^{-1}$) to produce a passive torque different to that at $10^\circ\,\mathrm{s}^{-1}$. This suggests that the effects of viscous resistance may be ignored when modelling knee velocities up to $120^\circ\,\mathrm{s}^{-1}$, but not for $240^\circ\,\mathrm{s}^{-1}$.

The dynamic joint torques used for further modelling were calculated by subtracting the passive torque measures from the active measures at each velocity. Therefore, the effect of viscous resistance has been eliminated from the results, even at the velocity of $240^\circ\,\mathrm{s}^{-1}$. Active knee extension torque declined significantly with increased velocity ($p<0.001$), with average torque at $240^\circ\,\mathrm{s}^{-1}$ being only 35% of the isometric value (Fig. 5).

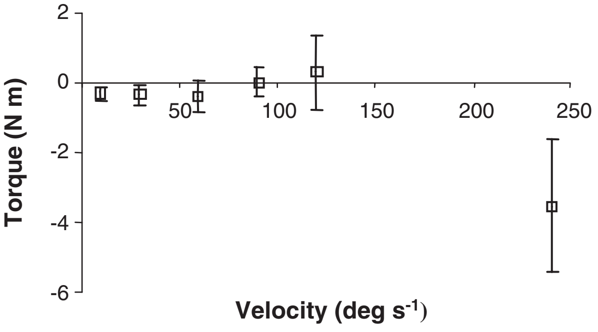

Fig. 2. Effect of knee extension velocity on passive torque.

5. Modelling processes

5.1. Tendon slack length

The purpose of this first section was to determine appropriate tendon slack lengths for the musculoskeletal model. Slack length is a critical parameter for determining the length–tension relationship of an entire muscle because it determines the length of a fibre for a particular muscle–tendon length. Slack length cannot be measured directly owing to the difficulty in establishing a length without tension on the series elastic component (Hoy et al., 1990). Hoy et al. (1990) estimated slack length by choosing lengths that would produce joint torque curves approximating those measured in vivo. Any change in a model's fibre length, muscle moment arms or whole muscle–tendon length would alter the model's joint torque–angle curve requiring new slack lengths to be fitted. The procedure of Hoy et al. (1990) was therefore used to produce appropriate slack lengths for the present model.

Fig. 3 illustrates modelled torques compared to the experimental data. Modelled data were generally well aligned with measured results owing to the fit of tendon length, but tended to underestimate measured values at each end of the range of movement. This result should have been anticipated because, in reality, not all quadriceps muscles would be expected to reach optimum length at exactly the same joint angle. The three

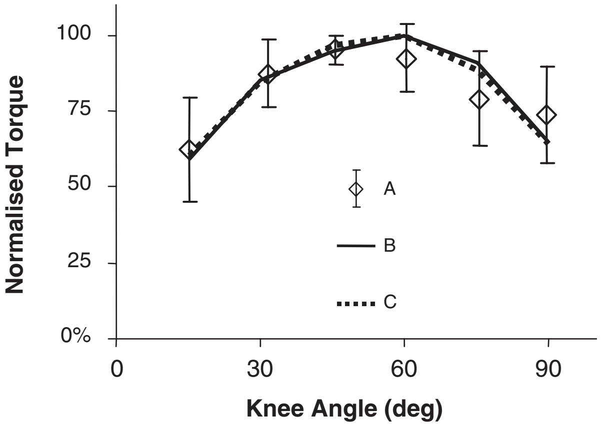

Fig. 3. Torque generated by NMES-evoked quadriceps contractions at different knee angles. (A) Mean ± standard deviation of measured data for all subjects. (B) Torques modelled after fitting tendon length and using moment arms from Kellis and Baltzopoulos (1999). RMS error = 13.8%. (C) Torques modelled after fitting moment arms as well as tendon length to the measured data. RMS error = 13.5%.

vastii muscles have been modelled as a single muscle by the present study, in common with the methods of Hoy et al. (1990) and Meijer et al. (1998), because of the similarity of anatomical properties within the group. The present study and that of Hoy et al. (1990) adjusted slack lengths for vastus and rectus femoris in order to have both muscles generating peak torque at the knee angle coincident with that generating maximum torque in vivo. Meijer et al. (1998) allowed rectus femoris and vastus to achieve peak torques at different knee angles, thus widening the active torque–angle curve. Forcing three vastii to behave as a single group, and allowing only rectus femoris to vary seems, however, to be an artificial solution. With rectus femoris having the smallest cross-sectional area, it is possible that quite a large error in estimated rectus femoris slack length could arise in order to compensate for the assumption of fixed angle of peak torque for all three vastii. While all four muscles could be modelled separately, this would increase uncertainty in the fitting process. Furthermore, the same criticism could be applied to modelling the quadriceps as four single fibres, rather than using a range of fibre and tendon lengths within each muscle (Ettema and Huijing, 1994). The present authors therefore prefer the option of forcing all quadriceps to peak at the same angle, rather than an arbitrary grouping of three vastii and allowing only rectus femoris to vary. It should be noted, however, that the method of Meijer et al. (1998) was trailed for the present study, and resulted in changes to rectus femoris and vastus tendon lengths of only 3% and 1.5%, respectively.

The tendon slack lengths resulting from the current study are shown in Table 1, together with those previously reported by Hoy et al. (1990) and Meijer et al. (1998). While there is consistency between the three studies in reported rectus femoris lengths, there is considerable variation in vastus length. While reasons for the differences are unclear, this does illustrate the importance of fitting tendon lengths within the specific model being developed, as only very small changes in tendon length are required to completely alter the modelled pattern of torque–angle curves (Sinclair, 2001).

5.2. Quadriceps moment arms at the knee

One option for improving the quality of fit between measured and modelled isometric torque was to modify the muscles' moment arm–joint angle relationship. While there were other options available (e.g. to modify the model's fibre lengths or the sarcomere force–length relationship), moment arms were a likely candidate because of the wide variety of values available from the literature. The purpose of this section was therefore to determine whether modifying the moment arms would improve the degree of fit.

Within the muscle model, the moment arm equation from Kellis and Baltzopoulos (1999) was replaced by a quadratic equation of the form

$$d_{\mathrm{k}} = a + b\theta_{\mathrm{k}} + c\theta_{\mathrm{k}}^2, \tag{1}$$

where a, b and c were constants. Altering a muscle's moment arm will affect its tendon slack length; hence, five independent variables were fitted simultaneously to minimise the squared differences between modelled and experimentally measured torques for vastus and rectus femoris.

Fitting moment arm as well as slack length made little difference to the overall quality of fit, with RMS residual error decreasing from 13.8% to 13.5% ($p = 0.31$) (Fig. 3). Fitted moment arms were reduced slightly, but remained similar to those of Kellis and Baltzopoulos (1999) (Fig. 4). The tendon slack lengths

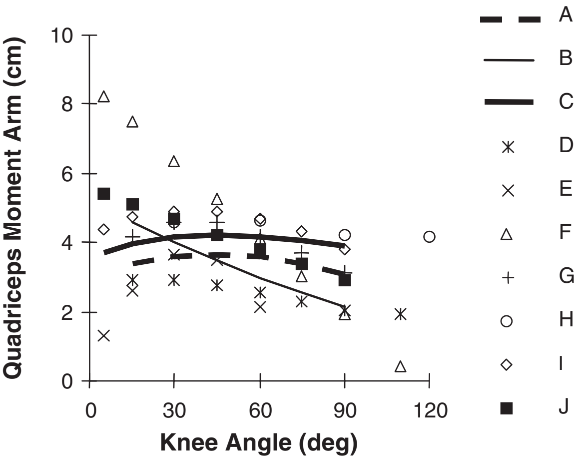

Fig. 4. Comparison between moment arms derived from a different source. (A) Moment arms fitted to experimental data using force–length equation from Delp et al. (1990); (B) moment arms fitted to experimental data using force–length equation derived from Stephenson et al. (1989); (C) Kellis and Baltzopoulos (1999); (D) Buford et al. (1997); (E) Grood et al. (1984); (F) Hawkins and Hull (1990); (G) Marshall et al. (1990); (H) Nisell (1985); (I) Smidt (1973); (J) Visser et al. (1990).

Table 1
Comparison between slack lengths fitted by Hoy et al. (1990) and those fitted within the present section

	Hoy et al. (1990)	Meijer et al. (1998)	Present study
Rectus femoris (m)	0.373	0.333	0.351
Vastus (m)	0.228	0.267	0.156

fitted with quadriceps moment arm varied by less than 3% from those reported in Table 1.

5.3. *Force–velocity relationship*

Constants for Hill's force–velocity equation (Eq. (6), Appendix A) vary with fibre type. Furthermore, there is substantial variation in the range of Hill constants that have been measured experimentally (Baratta et al., 1995). Muscles of SCI individuals are relatively homogeneous, comprising mostly Type II fibres (Crameri et al., 2000; Round et al., 1993). Therefore, vales of 0.35 × isometric force and 2.25 × fibre length have been chosen as initial values for Hill constants *a* and *b*, respectively. It could be expected, however, that SCI may alter the dynamic properties of skeletal muscle in ways not fully understood. Computer simulations were repeated, using Hill constants *a* and *b* as independent variables for fitting. Similar to previous sections, modelled knee extension torques were compared to experimental data and the Hill constants adjusted to minimise the squared residual error between measured and modelled data. Fig. 5 illustrates the comparison between measured results and those modelled using Hill constants derived from Pierrynowski and Morrison (1985).

Fitting new Hill constants produced almost the same values used by Pierrynowski and Morrison (1985) (Table 2) and did not reduce the RMS error between measured and modelled values ($p = 0.45$). The error was therefore due to experimental variation between subjects, rather than a systematic lack of fit to the model. These results give no justification for changing the model's constants away from those originally selected from Pierrynowski and Morrison (1985). It appears that NMES-evoked muscle contractions have the force–velocity relationship that would be anticipated from a muscle primarily composed of fast twitch fibres.

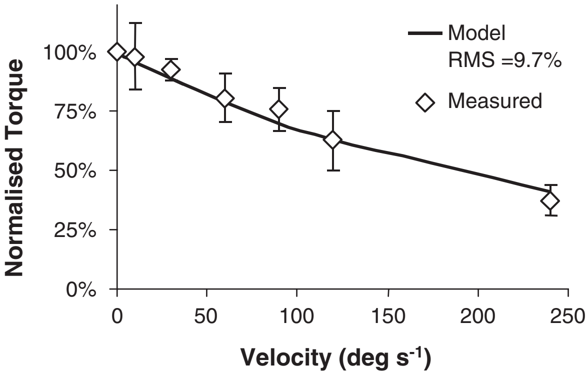

Fig. 5. Comparison between torques measured at 60° knee flexion for all subjects at each velocity of knee extension and those modelled using Hill constants from Pierrynowski and Morrison (1985).

5.4. *Interaction between parameters*

The present model has thus far used the dimensionless fibre force–length equation described by Delp et al. (1990). An alternative model for the force–length relationship was based on formulae presented by Stephenson et al. (1989), using human filament lengths reported by Walker and Schrodt (1974) (full details may be found in Sinclair, 2001). The resulting force–length relationship generated a curve of similar shape to that of Delp et al. (1990), but produced less active force at shorter fibre lengths (Fig. 6).

When the narrower force–length pattern was used to estimate isometric torque at each knee angle, the model found that torque would be further reduced at extreme joint angles (Fig. 7C). Decreasing the model's quadriceps moment arm, however, could widen the torque–angle curve by reducing the amount of muscle length change for a given range of joint angles. Using the methods described above, a new moment arm–angle equation was generated, improving the quality of fit to the experimental data (Fig. 7D).

The curve illustrated by Fig. 7D initially looked like a satisfactory solution. The fit between measured and modelled data was much improved and the new moment

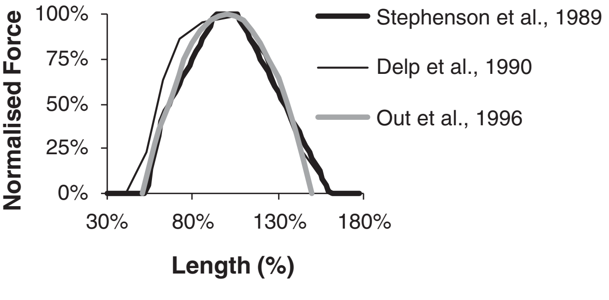

Fig. 6. Comparison between sources of the force–length relationship for a single muscle fibre.

Table 2
Changes in Hill constants with fitting procedure

	Constant *a*	Constant *b*
Constants from Pierrynowski and Morrison (1985)	0.35 × isometric force	2.25 × fibre length
Constants fitted using force–length relationship from Delp et al. (1990)	0.351 × isometric force	2.254 × fibre length
Constants fitted using force–length relationship from Stephenson et al. (1989)	0.550 × isometric force	2.196 × fibre length

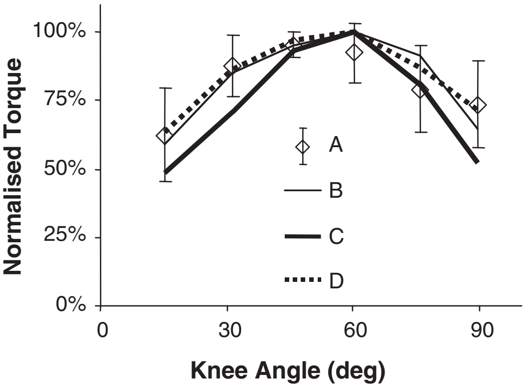

Fig. 7. The effect of force–length relationship on torque generated by NMES-evoked quadriceps contractions at different knee angles. (A) Mean ± standard deviation of measured data. (B) Torques modelled using force–length from Delp et al. (1990) and moment arms from Kellis and Baltzopoulos (1999). RMS error = 13.8%. (C) Force–length equation replaced by that derived from Stephenson et al. (1989). RMS error = 16.7%. (D) Force–length equation of Stephenson et al. (1989), but with moment arms fitted to the experimental data. RMS error = 13.5%.

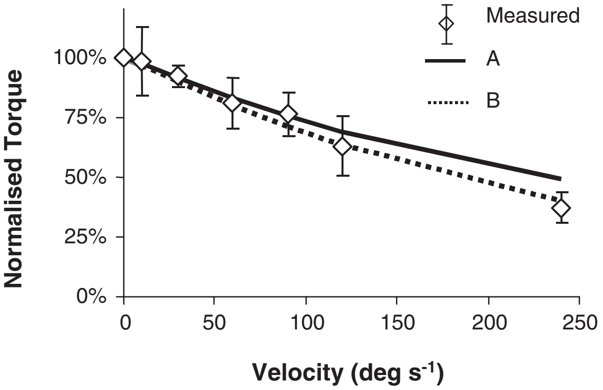

Fig. 8. The effect of fitting Hill constants on modelled torque–velocity relationship. (A) Model results when using force–length equation from Stephenson et al. (1989), moment arms from Fig. 4B and Hill constants from Pierrynowski and Morrison (1985). RMS error = 10.5%. (B) Hill constants from (A) replaced by those fitted to experimental data. RMS error = 9.4%.

arm–angle relationship was within the bounds of values available from the literature. Indeed, the linear shape of the relationship was quite similar to that found by Visser et al. (1990) (Fig. 4B). Problems with this fitting procedure later became apparent when investigating Hill constants during dynamic contractions. At 60°, the fitting routine produced moment arms that were smaller than those found by all but two of the studies included in Fig. 4. A smaller moment arm requires that, for a given angular velocity of the knee, the linear velocity of shortening for the muscle would be reduced. Using smaller moment arms therefore meant that the model gave a smaller decline in torque with increasing angular velocity of the knee (Fig. 8A). To correct this problem and accurately model dynamic contractions meant that different Hill constants would be required. In fact, to correct the model as shown in Fig. 8B, Hill's constant *a* had to be increased from 0.35 up to 0.55 (Table 2). This was well outside the ranges of values commonly reported in the literature (Baratta et al., 1995; Fitts et al., 1991), and therefore was perceived not to be a valid solution from the fitting routine. Thus, using a narrower force–length relationship forced us to use an inappropriate force–velocity relationship to preserve the model's fit to experimental data.

The above discussion demonstrates that fitting constants to one part of a model will have consequences on other parts of the model. For example, the two force–length functions used above are both commonly used in the modelling literature, yet the relationship from Delp et al. (1990) gave a better fit to measured data. Changing the moment arms was successful in the immediate task of fitting isometric torques at different knee angles and produced moment arms that were quite reasonable when compared to the range of published values available (Fig. 4). The fitted moment arms, however, had deleterious effects on the model's performance during dynamic contractions and required unreasonable estimates of Hill constant *b* to remedy this. As a consequence, prior literature reporting suitable values for Hill constants based on whole muscle models (e.g. Baratta et al., 1995) must be interpreted in consideration with the force–length relationship assumed within the model.

Besides adjusting moment arms, there were alternative solutions to fit the model to measured isometric torques. For example, optimum fibre length could have been increased as longer fibres generate force over a greater length (Lieber, 2002). Alternatively, the optimum length of individual quadriceps muscles could have been set to occur at different knee angles, or the fibre force–length relationship altered. Just because changing moment arms resulted in an excellent fit does not mean that this produced the most valid model.

Meijer et al. (1998) found that a simple model using parameters that were all measured from the same four subjects gave the best fit to voluntary isometric knee extension torque–angle curves. They concluded that this was due to using parameters sourced from a single population so that co-variance of parameters was taken into account. Unfortunately, not all components of the model could be measured from cadavers, and this may have affected the conclusions of Meijer et al. The parameter showing the greatest difference between values estimated from the literature and measured by Meijer et al. was vastus fibre length. Their vastus fibre length was 11.3 cm; significantly longer than has been commonly reported in the literature (Wickiewicz et al., 1983; Herzog et al., 1990). Meijer et al. used the force–length equation of Out et al. (1996) (Fig. 6), which was not measured directly within their study. Had they used a wider force–length relationship (e.g. Delp et al., 1990), they might have concluded that a shorter fibre length from the literature provided a better fit. Alternately, had they used three separate vastii and allowed them to reach maximum torque at different knee angles, this would also have improved the fit of their literature value parameter model to be indistinguishable from their measured parameter models.

When a model does not fit the observed data, it is difficult to determine exactly which parameters are responsible because of the interdependent nature of the parameters. While Meijer et al. (1998) demonstrate that measuring all parameters from the one source improved the fit for their data, they could not be sure that this was the best solution. When developing a model to apply to a range of individuals, it is difficult to justify the use of parameters measured from only a small number of subjects. This is particularly so when some of the parameters (e.g. vastus fibre length from Meijer et al., 1998) are considerably different to those commonly reported in the other literature.

5.5. *Activation level*

A caveat implicit to our research was the assumption that muscle activation was constant across all conditions. It is unlikely that activation was maximal during this experiment; however, a large percentage of the muscle was likely to be active (Hillegas and Dudley, 1999). Furthermore, Gorman and Mortimer (1983) report that the fibre recruitment curve for monophasic surface stimulation is relatively steep, implying that those fibres that are active are likely to be tetanically fused. The most likely scenario for activation is therefore that most of the quadriceps fibres were fully active, a few were partially active and a few were not activated.

If activation were to change with knee angle, most likely due to movement of muscle fibres with respect to the electrodes, then this would generate a systematic change in muscle torque with joint angle. The close match between measured and modelled joint torque–angle relationships suggests there is not a strong effect, although it is possible that the fitting of slack lengths may have masked such a change.

Even if activation were constant, an activation less than 100% would affect the selected Hill constants, owing to the presence of an activation term in the force–velocity relationship (Eq. (6)). Re-fitting the constants assuming an activation of 80% had no effect on the value for Hill constant a, however constant b rose slightly from 2.25 to 2.52 × lfo. This effect was relatively small, however, compared to the changes discussed when the alternate force–length relationship of Stephenson et al. (1989) was used. It therefore seems unlikely that the activation level will have affected the outcomes of this research.

6. Conclusions

Musculo-skeletal modelling was able to match knee extension torques for NMES-evoked knee extension contractions by individuals with SCI. The torque–angle and torque–angular velocity relationships from isometric and isokinetic contractions were well matched to experimental data using parameters derived from prior literature.

Model parameters for tendon slack length were fitted to isometric data collected at different knee angles between 15° and 90° of flexion. The best fit for rectus femoris and vastus came from tendon slack lengths of 0.351 and 0.156 m, respectively. Fitting the model's muscle moment arms to isometric data at each knee angle was unable to provide a better match than those made using moment arms derived from Kellis and Baltzopoulos (1999).

Passive, isokinetic knee extensions were conducted at velocities between 10 and $240^\circ\,\mathrm{s}^{-1}$, with no significant passive resistance up to $120^\circ\,\mathrm{s}^{-1}$. Only at $240^\circ\,\mathrm{s}^{-1}$ was resistance significantly greater than at $10^\circ\,\mathrm{s}^{-1}$, suggesting that velocity dependent resistance may be excluded from models of isokinetic contractions for extension velocities up to $120^\circ\,\mathrm{s}^{-1}$. During active, NMES-evoked contractions, knee extension torque declined in accordance with Hill's force–velocity relationship. Hill constants reported by Pierrynowski and Morrison (1985) for fast twitch muscle fibres provided an accurate match to the decline in muscle force with velocity. This finding is consistent with previous reports that leg muscles of SCI individuals are comprised primarily of fast twitch fibres.

Muscle parameters such as moment arms, tendon lengths and Hill constants interact in complex ways. For example, the choice between two commonly used force–length relationships has been shown to affect the choice of constants within the force–velocity relationship. Therefore, great care must be taken when selecting values to ensure that parameters selected for one variable do not negatively impact on other components of the model. This is a major limitation of obtaining parameters by fitting curves to experimental data, rather than measuring the parameters directly.

Appendix A. Model details

The "Hill-type" muscle model consists of a contractile element in series with an elastic element (Fig. 9).

The total length of the muscle–tendon unit (l_{mt}) in the anatomical position was determined from equations

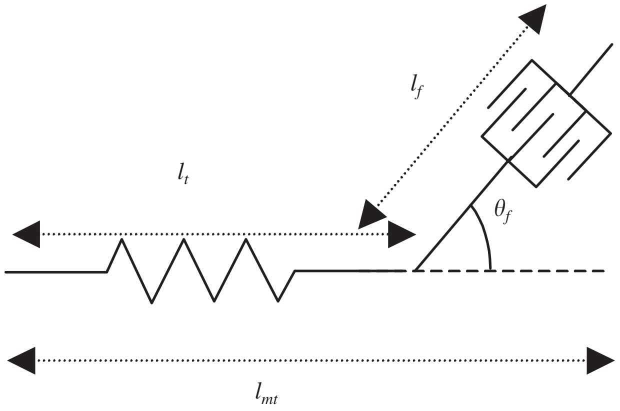

Fig. 9. Illustration of the muscle model showing whole muscle–tendon length (l_{mt}), tendon length (l_{t}), fibre length (l_{f}) and pennation angle (θ_{f}).

presented by Hawkins and Hull (1990), scaled to a thigh length of 0.39 m. Length changes away from the anatomical position were made by integrating the muscle's moment arm with respect to joint angle. Quadriceps moment arm, as a function of joint angle (d_{k}), was derived from data reported by Kellis and Baltzopoulos (1999):

$$d_{\mathrm{k}} = 0.0347 - 0.01469\,\theta_{\mathrm{k}} - 0.0084857\,\theta_{\mathrm{k}}^2\,m. \tag{2}$$

Pennation angle (θ_{f}) varied as a function of fibre length (l_{f}) according to the following equation, derived from Pierrynowski and Morrison (1985):

$$\theta_{\mathrm{f}} = \mathrm{ArcCsc}\left[\frac{l_{\mathrm{f}}\,\mathrm{Csc}[\theta_{\mathrm{fo}}]}{l_{\mathrm{fo}}}\right]. \tag{3}$$

Tendon length (l_{t}) was calculated from total muscle–tendon length, fibre length and pennation angle according to

$$l_{\mathrm{t}} = l_{\mathrm{mt}} - l_{\mathrm{f}}\,\mathrm{Cos}[\theta_{\mathrm{f}}]. \tag{4}$$

Tendon force (F_{t}) was calculated from length changes, assuming a linear tendon stiffness to allow the tendon to strain 3.3% at maximal force production (Zajac, 1989). While the assumption of linear tendon stiffness is an oversimplification, there is little data to suggest what shaped non-linear curve to use for paralysed muscle. The tendon applied force in the direction of the fibre (F_{f}) in proportion to the angle of pennation of the fibre (Fig. 9):

$$F_{\mathrm{f}} = \frac{F_{\mathrm{t}}}{\mathrm{Cos}[\theta_{\mathrm{f}}]}. \tag{5}$$

The maximum isometric fibre force varied as a function of fibre length according to a relationship published by Delp et al. (1990). The model worked by finding a numerical solution to the equation of fibre velocity (l'_{f}) as a function of force (6, Stern, 1974), using the NDSolve function within Mathematica. The NDSolve function uses an iterative procedure to solving differential equations and automatically adjusts solution step size to changes in the stiffness of the equation. Activation level (A) was set at 1 for all simulations reported in the present study:

$$l'_{\mathrm{f}} = \frac{b(F_{\mathrm{f}} - F_{\mathrm{fo}})}{aA + F_{\mathrm{f}}}. \tag{6}$$

A number of constants were derived from the literature for use in these simulations. These were:

Hill's constant a	$a = 0.35\,F_{\mathrm{fo}}$	Pierrynowski and Morrison (1985)
Hill's constant b	$b = 2.25\,l_{\mathrm{fo}}$	Pierrynowski and Morrison (1985)
Optimum fibre length for vastus	0.085161 m	Wickiewicz et al. (1983)[a]
Optimum fibre length for rectus femoris	0.07452 m	Wickiewicz et al. (1983)[a]
Resting angle of pennation for vastus	0.157 radians	Pierrynowski (1995)
Resting angle of pennation for rectus femoris	0.175 radians	Pierrynowski (1995)
Relative force of vastus and rectus femoris	$F_{\mathrm{fo\,v}} = 5.04\,F_{\mathrm{fo\,rf}}$	Based on physiological cross-sectional areas reported by Pierrynowski (1995)

[a] Fibre lengths from Wickiewicz et al. (1983) were scaled up to use an optimum sarcomere length of 2.64 μm. Lengths were scaled because Wickiewicz et al. counted the number of sarcomeres in series, then estimated optimal fibre length assuming a sarcomere length of 2.2 μm. For human sarcomeres, however, a more appropriate optimal length, based on the length of human actin filaments, is 2.64 μm (Burkholder and Lieber, 2001).

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