University of Alberta

The Effects of Varying Fibre Compositions on Simulated SEMG Signals in the Time and Frequency Domains

by

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A thesis submitted to the Faculty of Graduate Studies and Research in partial fulfillment of the requirements for the degree of

Master of Science

Faculty of Physical Education and Recreation

Fall 2011
Edmonton, Alberta

ABSTRACT

Whether SEMG can be used as a tool to estimate muscle fibre type concentrations remains an interesting question in muscle physiology. It is speculated that fast twitch motor units may have increased conduction velocities and that this may lead to an increased SEMG mean power frequency when compared to their slow twitch counterparts. Unfortunately, the true relationship between conduction velocity and fibre type remains a mystery. This research makes use of a SEMG simulation model to help analyze how changes in contraction time, conduction velocity and twitch force assignment distributions within a muscle impact a simulated signal. Variations in contraction time and twitch force impacts SEMG signals in the time but not frequency domains. On the other hand, conduction velocity is proportional to SEMG frequency content but has no impact on force production. This work suggests that further extension of simulation models could include methods of varying fibre type compositions.

ACKNOWLEDGEMENTS

I would like to express my sincere gratitude to each of my respective committee members: Dr. Loren Chiu, Dr. Kelvin Jones and Dr. John Misiaszek. I am grateful for the patience you have all shown in this process. I would also like to thank my employers at the Canadian Athletics Coaching Centre, Dr. James Denison and Brian Kropman. The support from both of you has led me to complete this work in the face of my responsibilities at the Centre.

I would also like to thank my family, and in particular my mother for her support which has guided me through my education to this point. I would not be where I am today with you.

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CHAPTER 1:

REVIEW OF LITERATURE

INTRODUCTION

Electromyography (EMG) is a common tool for assessing the electrical dynamics occurring during various forms of muscle contractions (Basmajian, 1985). The most common technique is surface EMG (SEMG), which uses electrodes placed on the skin to observe electrical potentials due to any electrical activity in the underlying physiological structure. Action potentials are carried from the central nervous system through motor neurons towards the muscle fibres which these neurons innervate. The intersection between the motor neuron and the muscle fibres is known as the neuromuscular junction, and the action potential is transmitted across this junction into the respective contractile fibres. It is this activity that the SEMG will detect and is known as the motor unit action potential (MUAP). As a result of the simplicity, inexpensiveness and non-invasive nature of SEMG, it is used frequently in the analysis of many physiological phenomena including muscular fatigue, physical rehabilitation and biomechanical performance.

Although SEMG has the preceding advantages there are several complexities with collecting and processing data associated with this method. For one, the configuration of electrodes can dramatically influence the SEMG signal. For instance, if the electrode is placed directly over the innervation zone (IZ) it will produce a signal that is the result of two potentials travelling in opposite directions (away from the IZ) – resulting in some degree of signal cancellation. Similarly, an electrode placed over the muscle tendon will be influenced by the termination of action potentials at this point. Therefore the optimal location for electrodes is between the IZ and insertion point of the muscle. This suggests the length of the examined muscle is critical as there must be sufficient distance between the IZ and insertion point for electrode placement. For these reasons, standardized placements have been suggested (Rainoldi, 2004).

Also of importance is the use of electrode arrays, which are oftentimes linear with respect to the muscle fibre orientation. In research this often implies the use of two electrodes on either side of the IZ (bipolar configuration), but may sometimes refer to other types of arrays. Linear arrays afford researchers more accuracy for a number of physiological parameters including general muscle activation levels, muscle characterizations and muscle fibre conduction velocity estimations (Merletti, 2003; Rainoldi, 2001). The distance between electrodes in a linear array is also important as highlighted in the work of Beck (2005). Here, interelectrode distances (IED's) of 2, 4 and 6 cm were tested for their effect on signal amplitude and mean power frequency (MPF). While there appeared to be significant effects in absolute EMG amplitude and

MPF, normalized frequencies and amplitudes did not experience significant effects due to varying IED's. As EMG data is often presented in normalized values, this is positive for SEMG research efficacy.

As one might suspect, the distance from electrode to motor unit territory (i.e., the minimum distance between the muscle boundary and electrode) can significantly alter SEMG signals. Namely, as this distance increases MPF can drop by a factor of at least 300% (Fuglevand, 1992). Related to this is the finding that approximately 7% of the motor unit potentials registered by a SEMG electrode can be responsible for upwards of 50% of the signal amplitude (Keenan, 2006). Thus, the distance between the electrode and muscle must be known and accounted for.

The last potentially disruptive factor to account for when using SEMG for experimentation is the influence of crosstalk, which is the signal detected for a non-active muscle of interest in nearby contracting muscles. While crosstalk can be reduced with bipolar configurations, an improved method is to use a double differential spatial filter (a process in which aberrations in a signal can be filtered out) with a branched electrode (a special type of bipolar electrode) (van Vugt, 2000). This technique, which is easily achieved with standard EMG equipment, improves the accuracy and selectivity of SEMG signals by several magnitudes.

SIMULATION OF SEMG SIGNALS

Due to the difficulties in acquiring accurate and meaningful SEMG signals, simulation models can be useful in the characterization of muscular activity (Merletti, 1997) and in helping to understand experimental SEMG results (i.e., Farina, 2004b). Many such models have been developed and used for these purposes (Stegeman, 2000) and several use as a basis the model developed by Fuglevand (1992; 1993).

Fuglevand's original model, which is designed for isometric muscle actions, consists of three sub-models: 1) a motor neuron model, 2) a motor unit force model and 3) a SEMG simulation model. The latter two models are shown in the flowcharts of Figures 1-1 and 1-2 below, leaving the formulae of these models to Appendix D. The motor neuron component consists of the modeling of neural excitatory drive, the recruitment thresholds of the pool of motor units and the firing frequency. Excitatory drive for isometric muscle actions is usually a ramp function with a specified plateau that ensures the Size Principle is obeyed as smaller, lower threshold units (slow twitch) will then be recruited first. Firing frequency, not to be confused with contraction time, will govern the discharge rate of a motor unit once the excitatory drive surpasses that unit's recruitment threshold. It is important to note that a maximum voluntary contraction is defined as that produced when a motor unit has reached its maximum firing rate. Lastly, we note that the variation of contraction time, motor unit force and conduction velocity take place in the force component of the model (Figure 1-1; denoted ⬣), and leave the SEMG simulation component untouched. This fact is important to remember as it has a strong impact on the results of Chapter 2.

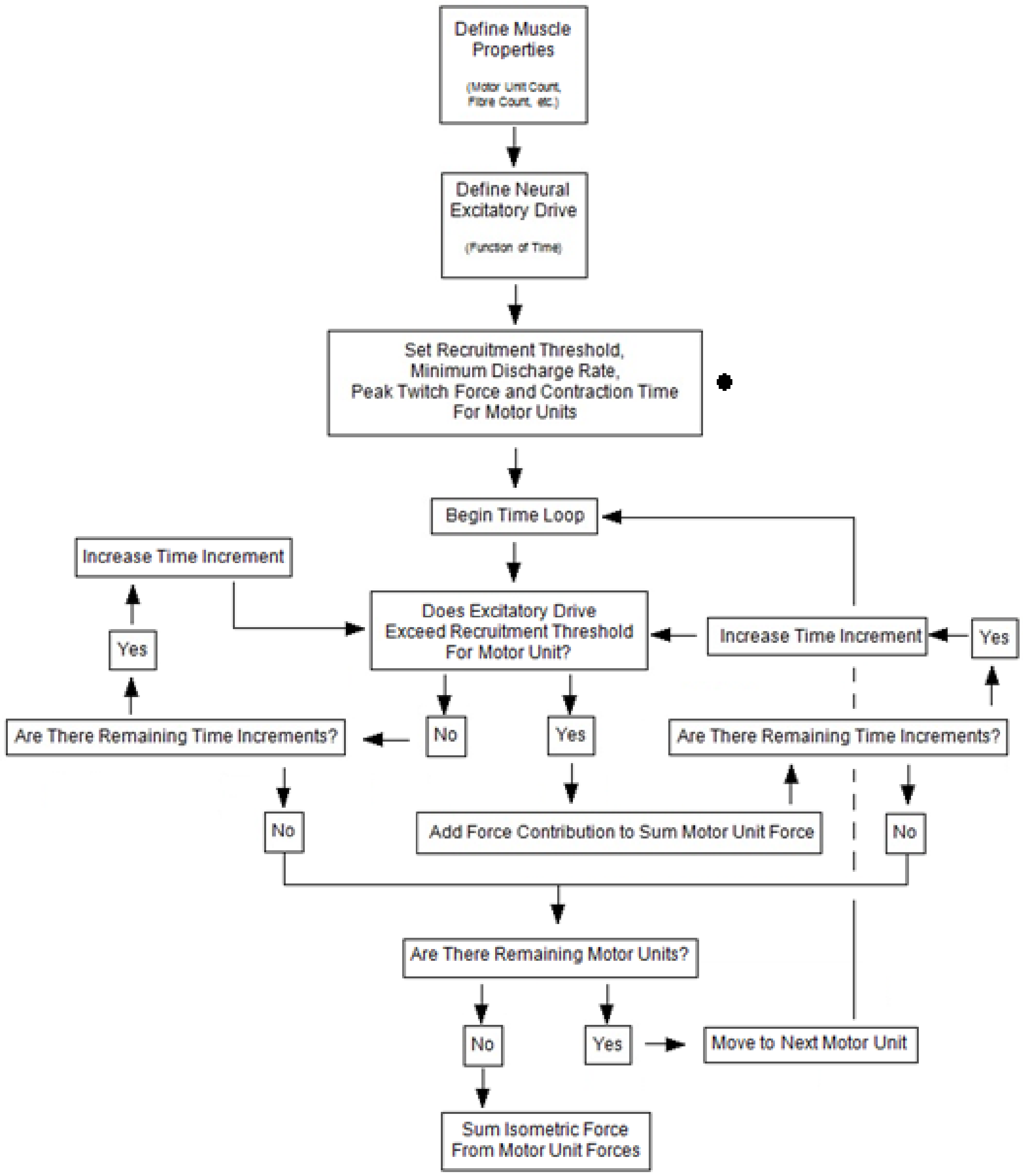

Figure 1-1. Flowchart of the Fuglevand motor unit force model. The ⬣ indicates the position in the flowchart where variations in this study take place. Here contraction velocity, twitch forces and conduction velocities were varied.

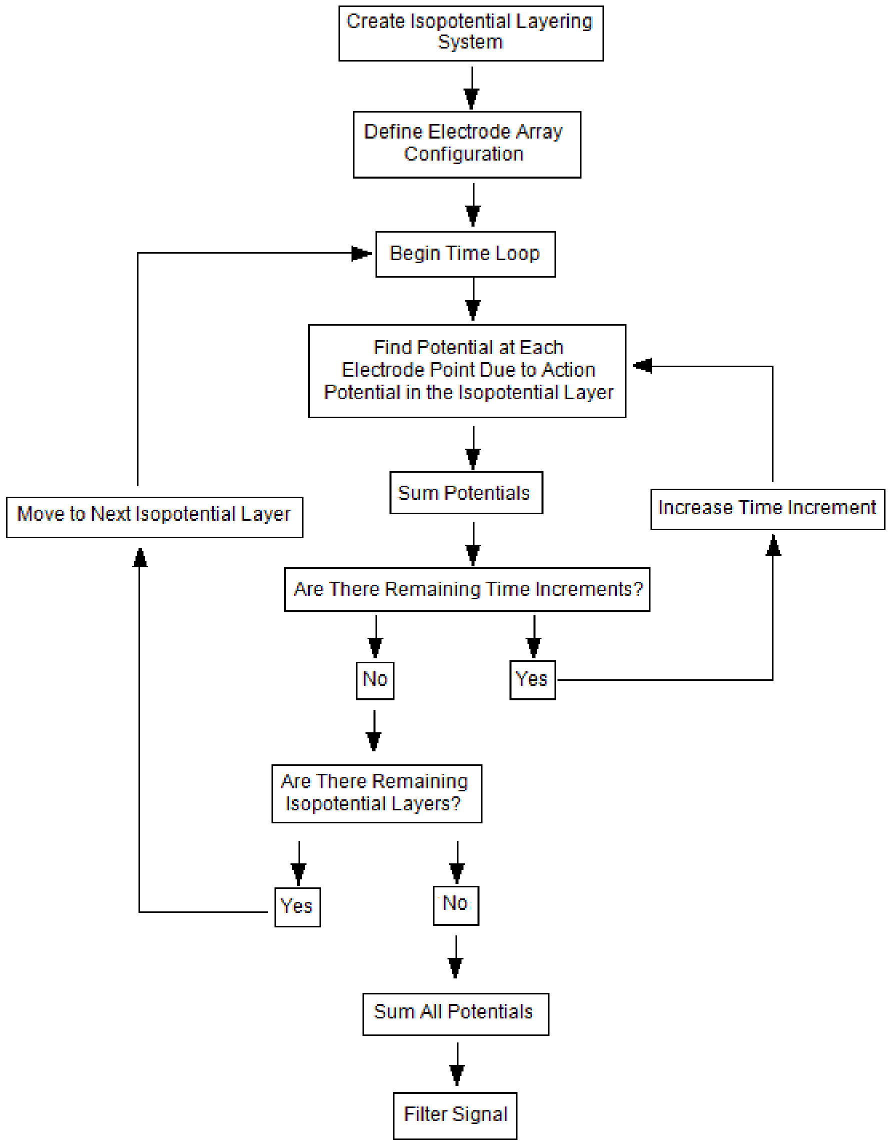

Figure 1-2. Flowchart of the Fuglevand SEMG simulation model.

This component of Fuglevand's model makes the assumptions that minimum and peak firing rates, together with the coefficient of variation (variability of discharge rate) are constant. These assumptions have, until recently, been cautiously implemented. These parameter settings are important; Enoka (2003) reports that the coefficient of variation has a significant influence on force fluctuations during steady state isometric contractions. In a paper from Moritz and colleagues (2005), these parameters were changed into a function of the motor unit recruitment threshold. By using the Fuglevand model in an attempt to match experimental results on force fluctuations of human hand muscles, these changes to the model were made and have resulted in better predictive accuracy of force fluctuations. It is important to note that while relating minimum firing rate to recruitment threshold is generally accepted, some caution has been raised for the use of linear relationships between both the peak firing rate and coefficient of variation with recruitment thresholds (Jones, 2005).

The isometric motor unit force model first utilizes an exponential assignment of peak twitch forces among the units. In turn, this twitch force variable is used in the mapping of contraction times. Thus, higher twitch force motor units have quicker contraction times (Figure 1-3).

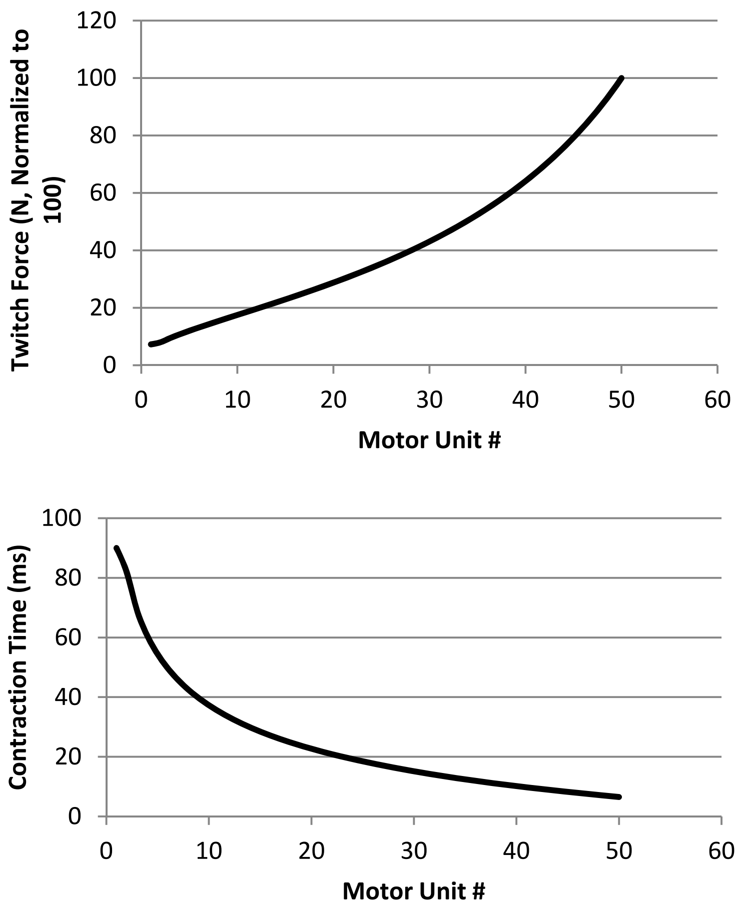

Figure 1-3. SEMG simulation model setting of motor unit twitch force and contraction time.

As contraction times are a primary mechanism of distinction between muscle fibre types (Burke, 1973), these assignments are crucial to the characterization of the muscle used for simulation. Using the firing rates from part one of this model together with the contraction times and twitch forces, the Fuglevand model creates a measure of isometric force production as a sum of individual motor unit contributions. The units of force in this model are mostly arbitrary, so the values are often given as a percentage of the force generated during a maximum voluntary contraction.

The third component to discuss is the SEMG simulation model. Here the depolarization of a muscle fibre acts to start a current running through the fibre volume. This current can reasonably be modelled as a dipole (Rosenflack, 1969). While the fibre has a set diameter, this current is assumed to run one dimensionally down the central portion of the fibre. In one of many extensions he has made to this model, Farina (2004c) extended this to allow the charge to run longitudinally or radially in the fibre. The electric potential is then evaluated at a point that simulates the SEMG electrode (Figure 1-4).

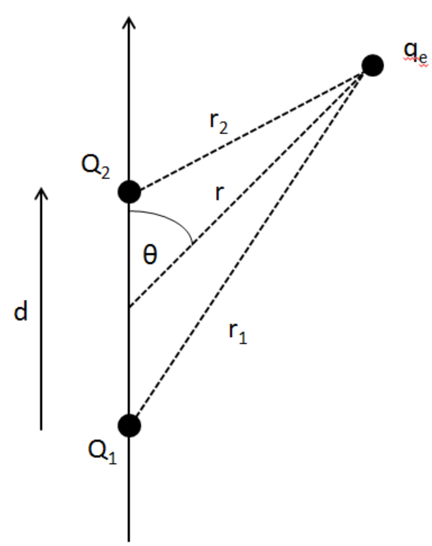

Figure 1-4. Electric potential measured at a point representing a SEMG electrode. Here the potential is measured at q_e at a distance r of a dipole with spacing d. (modified from Schumayer, 2010)

A muscle is broken down into several isopotential layers, each of which has a certain number of muscle fibres inside (from a varying number of motor units) that are roughly the same radial distance away from the electrode (Figure 1-5). Using these isopotential layers the electric potential is calculated for each discharge in each motor unit. Electrodes are modelled as an array of electrode points, and often a bipolar configuration is used to emulate common research methodologies.

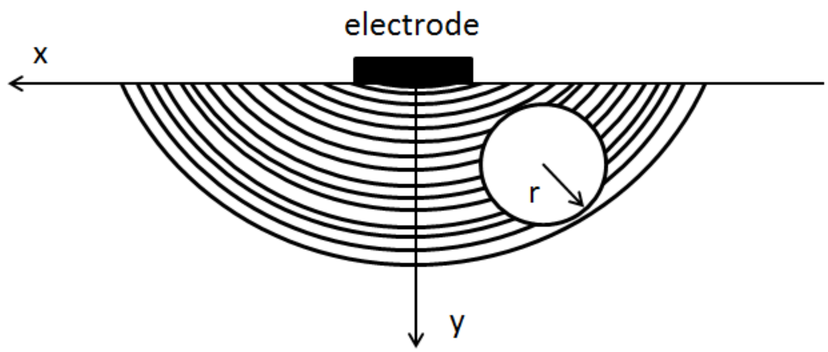

Figure 1-5. Isopotential layering system with a layer thickness of 0.5mm and a motor unit of radius R. Each motor unit of the simulated muscle will have a fibre count distributed throughout these layers that contribute to the potential measured at the electrode (modified from Fuglevand, 1992).

Extensions of the Fuglevand model, in addition to other SEMG simulation models, are often made and used for specific research purposes. For example, adaptations of this model and other SEMG modelling techniques are used in order to study fatigue (Farina, 2001; Fuglevand, 1999; Stegeman, 1992; Paiss, 1987). On the other hand the simulations themselves are becoming better understood in an effort to develop SEMG signal processing strategies, such as in the decomposition of signals to identify discharge times (Farina, 2010), how best to compensate for signal cancellation (Keenan, 2005) and to understand better how muscle motor units interact to produce SEMG signals (Keenan, 2006). These studies illustrate that these models can and are used effectively to support experimental SEMG findings and analyses.

FIBRE COMPOSITION EFFECTS ON EXPERIMENTAL SEMG FINDINGS

As mentioned, motor units can be classified as fast or slow twitch in accordance to their contraction durations. It is common to call the fibres innervated by such motor units as fast or slow twitch fibres, respectively, but the implication is on the type of motor unit. Although contraction time is one defining characteristic, researchers have often been curious as to whether or not other physiological parameters are correlated with motor unit type. Muscle fibre diameters, conduction velocities and twitch force values are all often studied as possible correlates to fibre type.

To discuss each of these physiological properties we begin with fibre diameters. Fibre diameter may change as a function of muscle fibre hypertrophy or atrophy. It is believed that the fibres in fast twitch muscle units have a greater propensity for hypertrophy. This is evidenced by strength training protocols looking at hypertrophy changes of both types (Hather, 1991; Hortobagyi, 1996), although these differences are not always substantial (MacDougall, 1979; Haikkinen, 1981). In these cases the assignment of exercise protocols has targeted hypertrophy of fast twitch fibres, yet in general, mean fibre diameters appears to be similar between the two fibre types (Mannion, 1997). This uncertainty over any correlation between fibre diameter and type is reflected in a fixed diameter assignment to all fibres in the Fuglevand model.

Conduction velocity is perhaps one of the most commonly studied parameters studied in conjunction with fast twitch fibres (Gerdle, 1988; Bilodeau, 2002). The main reason for these experiments is due to the fact that increased diameter size is proportional to higher average conduction velocity (Blijham, 2006). However, as discussed, there is insufficient evidence at this point to associate fibre types with fibre diameters and so doubts are raised about the relationship between fibre type and conduction velocity. Evidence supports this scepticism as conduction velocities appear to have a distribution that is independent of fibre type (Troni, 1983; Blijham, 2006).

Twitch force is another parameter commonly associated with muscle fibre type. Specifically, it is generally believed that the fibres of fast twitch units are capable of producing higher twitch force (Linssen, 1991; Gerdle, 2000; McArdle, 2007). While fast twitch units appear to innervate more fibres on average and for this reason produce more force (Milner-Brown, 1973), the twitch force capabilities of a single fibre is proportional to fibre diameter (Krivickas, 2011). Therefore, as the relation between fibre type and diameter is not definitive, the association between twitch force and fibre type is currently controversial. In the Fuglevand model, motor units having higher recruitment thresholds are assigned greater twitch force values. This assignment works as higher twitch forces are related to motor units with greater fibre counts by construction. Thus, motor units with greater twitch force innervate more fibres and thus produce more force, which avoids setting individual fibre twitch forces.

SEMG OUTPUT ANALYSIS

By design, SEMG is a technique used to evaluate muscle properties in a non-invasive way. Therefore, a wishful thought for physiologists is whether SEMG outputs can be used to assess muscle fibre composition. Signal amplitude has been successfully associated with increased force production in several studies (for example, Hagberg, 1989), while the signal mean power frequency (MPF) has been correlated to conduction velocity (i.e., fibre diameter) in others (Kupa, 1995; Wakeling, 2002). While we have discussed the difficulty in associating conduction velocity to fibre type, there is some controversial evidence that MPF could identify fibre types regardless (Gerdle, 2000).

In Gerdle’s study, muscle biopsies were taken from the vastus lateralis of 19 subjects. Each performed 100 maximal knee extensions (each taking one second each) with one second of passive rest. MPF and signal root mean square amplitude (RMS) were measured and correlated to the muscle composition found in the biopsy. The controversy surrounding this paper centres on the interesting results that both MPF and signal amplitude were not correlated to fibre diameter and that these measures were positively correlated to fibre type (MPF to type I and RMS to type II). That amplitude is linked to type II proportions is perhaps not surprising, but MPF has generally only been linked to other physiological parameters (i.e., conduction velocity) and not directly to fibre type. The more conventional results include the finding that as muscle force increases, which is typically where larger fast twitch motor units are recruited (Burton, 2004), MPF does not increase (Gabriel, 2009).

As we can see, there remains a significant debate as to whether variables such as fibre twitch forces, fibre diameters and conduction velocities, as well as how SEMG outputs such as the mean power frequency vary depending on fibre type. However, it is certainly evident that SEMG outputs are best studied in both the time and frequency domains. While SEMG amplitude is readily obtained (typically reported as an RMS) the MPF is found from a power spectral density function (PSD). This is generally a real-valued function with frequency domain that, as the name suggests, provides information on the power of the signal for certain frequency values. Methods of determining the PSD can be categorized as either parametric or non-parametric (Georgakis, 2003). A parameterized method, such as the autoregressive method, makes several assumptions based on the nature of the data and thus non-parameterized methods are generally a preferred method of PSD estimation (Zhang, 2010).

The most common non-parameterized method is the Fourier Transform (FT). The FT is a transformation from a function with time domain to one with frequency domain, and thus the result is generally time invariant:

$$\hat{f}(\xi) = \int f(t)e^{-2\pi it\xi}\, dt \tag{1.1}$$

for all real frequency values ξ. However, being time-invariant implies that this function will likely be insufficient for the non-stationary signals (signal changing over time) produced in typical SEMG studies. In order to provide information on what frequencies exist in the sample and when they are occurring, a different transformation method is needed.

The Short-Term FT (STFT) is an extension of the FT that uses the assumption that non-stationary signals may be locally represented as stationary ones. Mathematically, we can formulate this transform as

$$\hat{f}\left(\mathrm{t}',\mathrm{f}\right) = \int [\, f(t)\omega^{*}\left(t - t'\right)]e^{-2\pi itf}\, dt, \tag{1.2}$$

where ω^{*} is the (conjugated) window function that assesses the original function in the specified intervals. The product is a three dimensional viewing of frequency, time and amplitude. Indeed this method has often been used for spectral EMG analysis (Komi, 2000) but unfortunately some limitations still exist. Namely, the uncertainty principle dictates that we cannot know with precision the frequency of a signal at an exact moment in time (Zhang, 2010). This means that if we choose our sampling window function for the STFT to be too small, we will lose accuracy of the frequency data; but if the window is too large, we put at risk our assumption of the signal being stationary on this segment.

Fortunately a more accurate PSD estimation method exists: The Wavelet Transform (WT). In fact, as the WT is believed to increase analysis accuracy, several studies are emerging examining oft-studied questions using this new technique (for example, Karlsson, 2001). Depending on the research setting, perhaps the most popular method is the Continuous Wavelet Transformation (CWT) which has been used in many SEMG studies (Forrester, 2010; Hostens, 2004; Raymond, 2009), and can be expressed as

$$CWT(a,b) = \frac{1}{\sqrt{a}} \int x(t)\, \psi^{*}\left(\frac{\mathrm{t-b}}{\mathrm{a}}\right) \mathrm{dt}. \tag{1.3}$$

Here a and b are known as the translation and scale parameters and ψ is the (conjugated) “mother wavelet” function. The mother wavelet function is responsible for generating the window functions analogous to the STFT function. The choices of the windows (as well as scaling factors) still increase or decrease the accuracy of the result, but with more precision than that seen when using the STFT. For instance, one supposition is that the wavelet should be chosen to imitate the MUAP shape (Guglielminotti, 1992); however a more complicated but efficient choice of wavelets is the Morlet wavelet function:

$$\psi(t) = \frac{1}{\sqrt{\pi f_B}} e^{2i\pi f_C t} e^{\frac{-t^2}{f_B}} \quad . \tag{1.4}$$

Here, f_B and f_C represent bandwidth and frequency parameters. Recalling the time-frequency precision restrictions, the resolutions possible with this wavelet are given in (Yan, 2006):

$$\Delta t_i = \frac{f_C\sqrt{f_B}}{2f_i} \qquad\qquad \Delta f_i = \frac{f_i}{2\pi}\frac{1}{f_C\sqrt{f_B}}. \tag{1.5}$$

Once the CWT has been computed, the instantaneous mean power frequency can be found using:

$$IMPF(t) = \frac{\int_0^F fCWT(f,t)df}{\int_0^F CWT(f,t)df}, \tag{1.6}$$

which averaged out over time gives the MPF (integration limit of F is equal to one half the sampling frequency – the Nyquist frequency).

Theoretically the FT is deemed to be an acceptable method for the study of EMG spectral properties (Beck, 2005; Beck, 2006). However, there is some strong evidence that CWT provides for more accuracy than the STFT and other lesser known EMG methods of time-frequency analysis (Karlsson, 2000). This is the reason the WT is becoming more prominent in SEMG investigations.

SUMMARY

Muscle motor unit types, defined on the basis of contraction times, have often been examined in the effort to find other physiological parameters that are correlated to either fast or slow twitch units. Parameters classically examined for these correlations include the diameter, action potential conduction velocities and twitch forces of the innervated muscle fibres. Each of these variables is controversially correlated to fibre type with evidence supporting both sides of the debate. In turn these variables are believed to be associated with surface EMG outputs in both the time and frequency domain, which then gives researchers the hope of estimating fibre composition based on these signals even in the face of the uncertainty of their true correlations and the difficulty in using SEMG. To assist the analysis of SEMG data, the Wavelet Transform has started to appear more frequently in SEMG studies due to potential increases in computational accuracy. In addition, simulation models are often used in an effort to learn how muscular properties and dynamics influence SEMG signals. While certainly more research is needed to study the effects varying fibre type proportions have on SEMG signals, which can be supported through SEMG simulations, the question of what physiological properties are related to contraction time remains a looming mystery.

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CHAPTER 2:

The Effects of Varying Fibre Compositions on Simulated SEMG Signals in the Time and Frequency Domains

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This research was supported by the Natural Sciences and Engineering Research Council of Canada (NSERC) Postgraduate Scholarship Program

INTRODUCTION

Electromyography (EMG) is a common tool for assessing the electrical dynamics occurring during the various forms of muscle actions (Basmajian, 1985). Surface EMG (SEMG), which uses electrodes placed on the skin, is the most common technique used to measure electrical potentials due to any electrical activity of the underlying physiological structure. Action potentials are carried from the central nervous system through motor neurons towards the muscle fibres which these neurons innervate. The junction between the motor neuron and the muscle fibres is known as the neuromuscular junction, and the action potential is transmitted across this junction into the respective contractile fibres. It is this activity that the SEMG electrodes detect (known as the motor unit action potential (MUAP)). As a result of the simplicity, inexpensiveness and painless nature of SEMG's, they are used frequently in the analysis of many physiological points of interest including muscular fatigue, rehabilitation techniques and biomechanical performance, to name just a few.

Motor units are generally classified as fast or slow twitch depending on their contraction times (Burke, 1973). Several other physiological parameters have been studied to assess possible correlations with these motor unit types; such variables include muscle fibre diameters, conduction velocities and twitch force magnitudes. While it has been speculated that fibre diameter may be greater in fast twitch fibres, it has been shown that mean fibre diameter appears to be similar regardless of type (Mannion, 1996). Increased conduction velocities are found in fibres with greater diameter (Kupa, 1995; Blijham, 2006), thus any possible correlations between conduction velocity and fibre type is also in doubt (Troni, 1983). Lastly, twitch force magnitudes have been reported as higher for type II fibres (Linssen, 1991; Gerdle, 2000; McArdle, 2007). In most cases the justification for this association is again through fibre diameters (Krivickas, 2011), which thus brings forth uncertainty over any association between twitch force and fibre type. Resolving these correlations is an important question in physiology today.

It is a wishful idea that properties of muscle composition, such as fibre type proportions, could be estimated by SEMG outputs either in the time or frequency domains. While SEMG amplitude has been positively correlated with force production in many studies (for example Hagberg, 1989) this is clearly not enough to draw any intramuscular conclusions. More interestingly the signal mean power frequency (MPF) has in some cases been correlated to various muscular properties such as fibre cross-sectional area and conduction velocity (Kupa, 1995; Wakeling, 2002; Gerdle, 1988). Even more interestingly MPF has in some cases been linked directly to fibre type (Gerdle, 2000; Gerdle, 1988), furthering the call for more research on fibre type properties.

Traditionally SEMG frequency content has been studied with the help of the Fourier Transform (FT). However, it has been suggested that the

Wavelet Transform (WT), together with the Morlet wavelet, can provide more accuracy than the FT (Karlsson, 2000). The Continuous WT (CWT) allows for researchers to plot frequency content as a function of continuous time. With this method, instantaneous mean power frequency (IMPF) can be evaluated using

$$IMPF(t) = \frac{\int_0^F f CWT(f,t) df}{\int_0^F CWT(f,t) df}, \tag{2.1}$$

which can be time-averaged to provide a measure of the MPF.

As there are several difficulties to be accounted for in the experimental setup for SEMG studies (Rainoldi, 2004; Merletti, 2003; Beck, 2005) and in the processing of SEMG signals (van Vugt, 2000; Keenan, 2005), SEMG simulation models are a helpful method for both understanding how possible intramuscular workings contribute to SEMG signals and in comparisons to experimental findings. A popular isometric SEMG model by Fuglevand has been used in several such studies and is explained in detail in this original paper (Fuglevand, 1993). As expected, various extensions have also been developed (Moritz, 2005; Farina, 2001; Farina, 2004) since the publication of this original model.

Using this model, varying assignments of contraction times, conduction velocities and twitch forces can represent different fibre types to examine the effect that varying fibre type distributions may have on simulated SEMG signals in the time and frequency domain. This research studies these effects with the intention of supporting the analysis of experimental

SEMG data, particularly once the correlation between contraction time and other physiological properties is properly understood.

METHODS

This research uses the Fuglevand model with several adaptations, including those developed by Moritz and colleagues (2005) to account for a proportionality between motor unit recruitment thresholds and the minimum, peak and variation of firing rates (variables which are left constant in the original model). Simulations were run using custom written codes in MATLAB software (version R2009a; MathWorks, Massachusetts, USA). Parameters, unless specified in Table 2-1, are similar to those presented in the original paper (Fuglevand, 1993).

Model Parameters			
Number of Fibres	**Average Muscle Fibre Radius (μm)**	**Muscle Radius (cm)**	**MU Count**
47,259	23	0.50	50
MU Discharge Count	**Electrode to Muscle Territory Distance (cm)**	**Number of Electrodes**	**Interelectrode Distance (cm)**
100	1	2	2

Table 2-1. SEMG model parameters. These values were unchanged during this investigation.

Three conditions were tested to assess the influence that physiological parameters commonly associated with type I or type II fibres had on the simulated SEMG signal (Table 2-2). In each condition, contraction time was varied to account for fibre type distribution. In conditions 2 and 3, twitch force and conduction velocity were varied separately in accordance with their controversial connections to fibre type.

Condition Parameters			
Condition #	CT	TF	CV
1	Varying		
2	Varying	Varying	
3	Varying		Varying

Table 2-2. Condition parameters.

In all conditions, CT was assigned using scaled values of the original model formulation (Figure 2-1). For condition 2, the TF for each motor unit i (to a total of n) for the fast and slow distributions was assigned via the function

$$TF(i) = \frac{k(\frac{i}{n})}{k - (\frac{i}{n}) + 1}. \tag{2.2}$$

The values of k were chosen to best reflect approximately 80% type I (k=0.3) and 80% type II (k=-1.3) distributions with a linear function for the 50% type I/type II fibre composition (slow, fast and average, respectively; shown in Figure 2-2). In condition 3, CV was assigned to a value of 3, 4 or

5m/s to represent slow to fast proportions, respectively (taken from the ranges presented in Fuglevand, 1993).

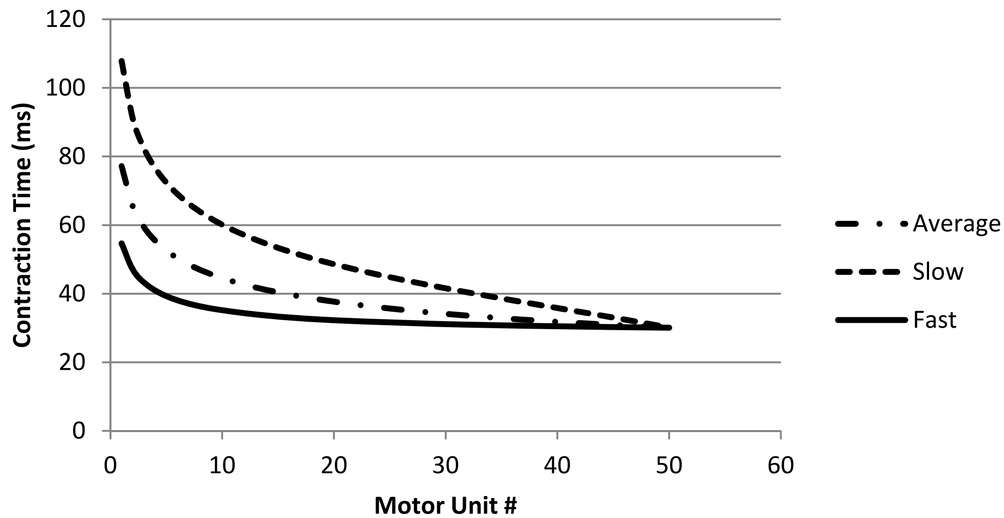

Figure 2-1. Motor unit contraction time assignment. The three figures represent the slow, average and fast fibre muscle compositions used in all conditions.

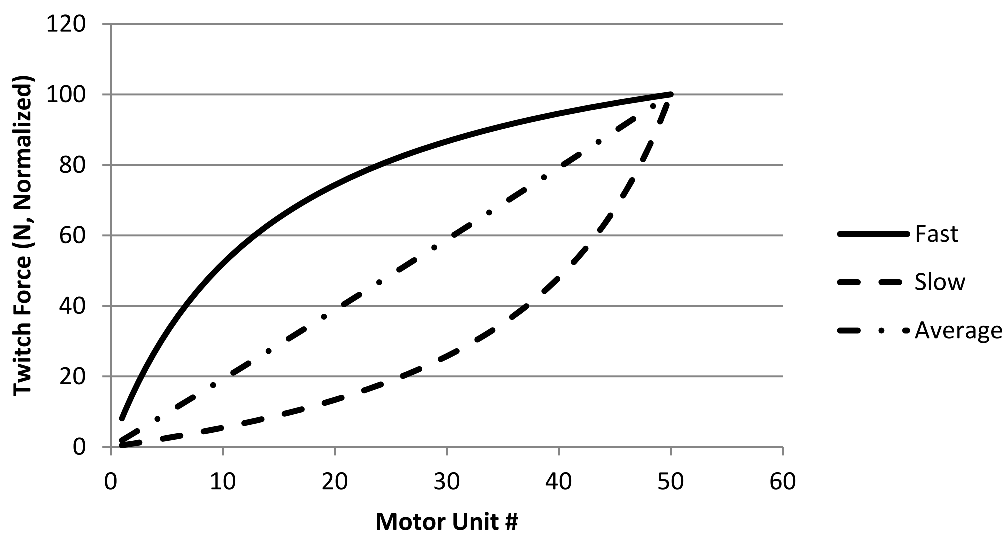

Figure 2-2. Motor unit twitch force assignment. The top curve represents the fast, the middle curve the average and the bottom curve the slow distribution. This setting was used in Condition 3.

Neural excitation, as in the original model, was modelled as a linear increase from 0 to 1 second followed by a constant excitatory drive that ran long enough to allow 100 motor unit discharges for each motor unit. Contractions were held at a sub-maximal level (60% MVC), and SEMG signals were high-pass filtered with a first order Butterworth digital filter (5Hz cut-off frequency) to simulate that performed in most laboratories (Fuglevand, 1992). At this point signal analysis took place including RMS, force and MPF calculations – the latter with use of the CWT with Morlet wavelet. As random elements were included in the model, particularly in the timing of motor unit contractions, ten trials of every condition were simulated with the mean and standard deviation values reported.

To test for differences between the three fibre-type distributions, one-way ANOVA was used in SPSS 11.0 (IBM Corporation, Armonk, NY). A separate ANOVA was conducted for each parameter (mean isometric force, SEMG amplitude and MPF) in each condition. If an ANOVA was significant, Tukey post-hoc tests were used to determine which means were different. For ANOVA and Tukey tests, alpha was set *a priori* ($\alpha=0.05$).

RESULTS

Condition 1: Varying of the CT

The first condition accounted for fibre composition differences by varying the CT with all other parameters fixed. For this condition we show the filtered SEMG output, the associated RMS of this time-based signal and the isometric force-time curves, and note that these curves are similar in shape for each condition and will thus not be depicted for subsequent conditions outside of the reporting of the variable calculations. Figures 2-3 and 2-4 show the SEMG signal and the RMS of this output.

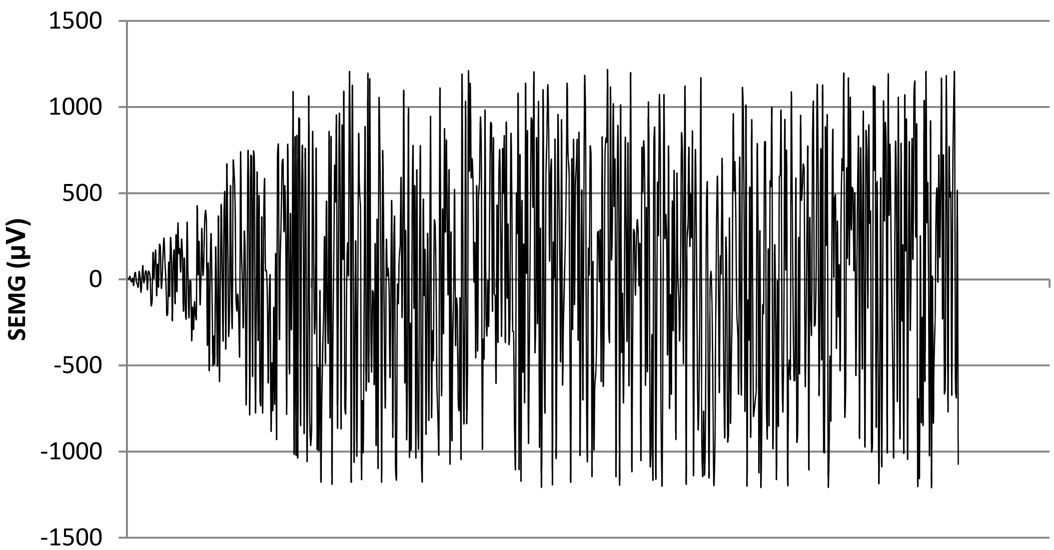

Figure 2-3. SEMG output as a function of time. Neural excitation grows linearly from 0 to 1 second, and then remains constant for the duration of the simulation.

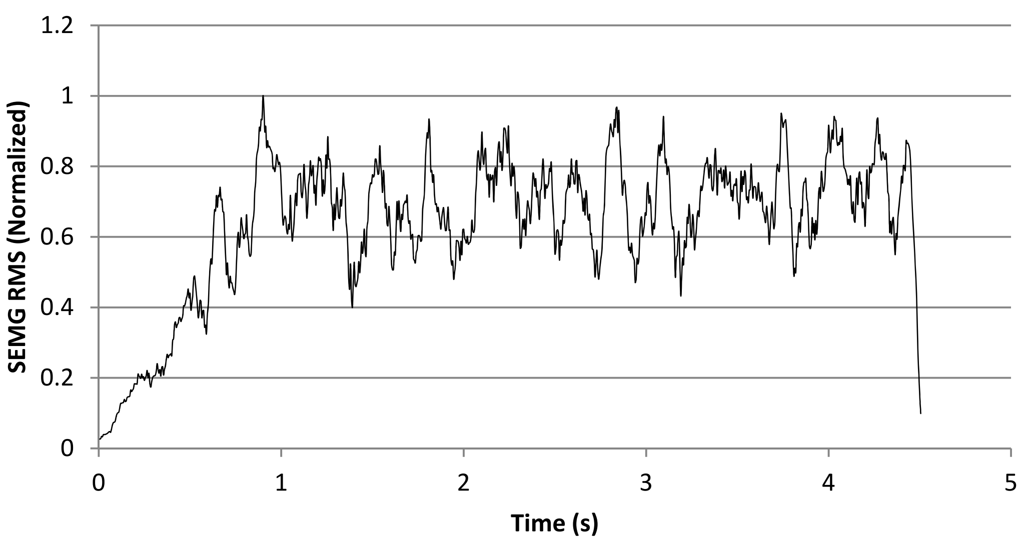

Figure 2-4. Normalized RMS of the time based signal (100ms window width).

The firing rates of the motor units are shown in Figure 2-5, noting that discharge rates are faster in the fast twitch units (as assigned). These firing rates were used for each condition.

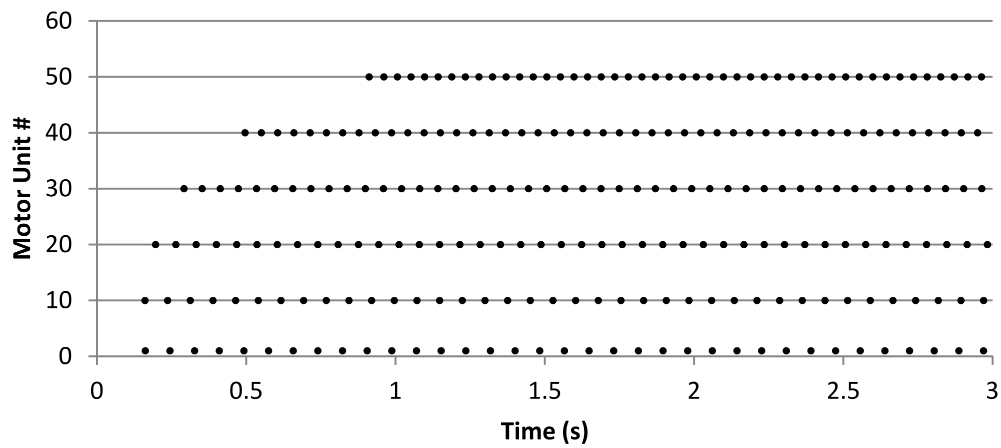

Figure 2-5. Motor unit firing rates. These rates, shown here for the first 3 seconds of excitation, were identical for each condition. Neural drive was linearly increased from 0 to 1 second.

In this condition, an ANOVA revealed a significant effect of fibre-type grouping on isometric force (P<0.05). Post hoc tests indicated significant differences existed between each of the fibre-type distributions (P<0.05). The reported force-time data was the average quantity found at and following the point at which the neural excitation hit a plateau and thereafter remained constant. Data are presented as values normalized to that generated by the fast fibre composition (Figure 2-6).

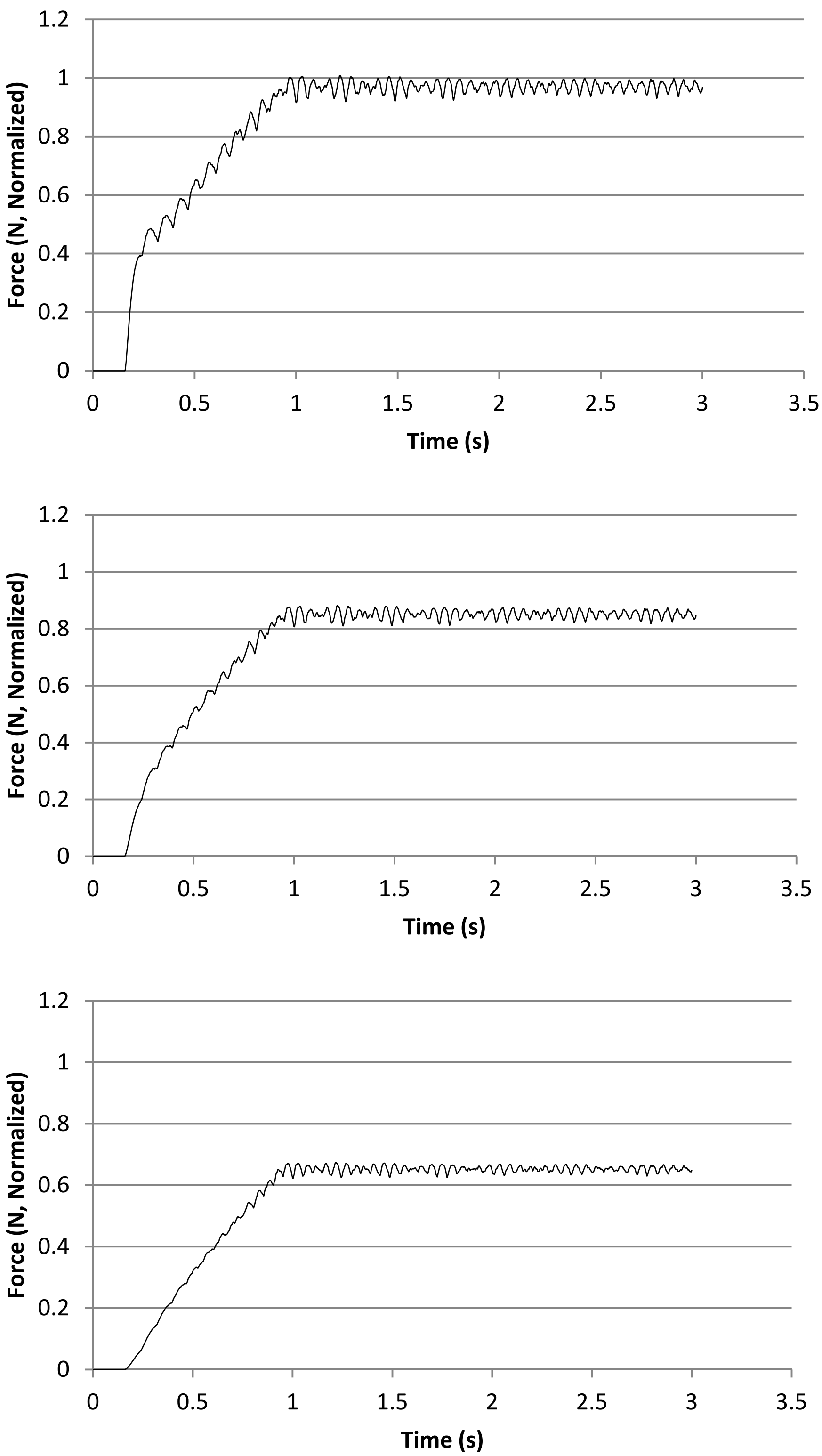

Figure 2-6. Isometric force production for the fast (upper graph), average (middle graph) and slow (lower graph) fibre compositions in Condition 1.

No significant differences were found for EMG amplitude or mean frequency in Condition 1. The results of this condition are presented in Table 2-3.

Condition 1 Results			
Fibre Comp.	**Mean Isometric Force at Maximum Excitation (Normalized to Fast Comp.)**	**SEMG Amplitude (Normalized RMS)**	**MPF (Hz)**
Slow	68.67% (±2.17)	100.12% (±4.27)	50.38 (±0.08)
Average	79.14% (±2.18)	100.16% (±2.69)	50.40 (±0.09)
Fast	100.00% (±1.67)	100.00% (±1.41)	50.41 (±0.08)

Table 2-3. Results of Condition 1 where fibre composition was varied through changes in the CT.

Condition 2: Varying of the CT and TF

Alongside contraction time, twitch forces were varied using the formulation depicted in the previous section. The ANOVAs for isometric force, MPF and signal RMS each revealed a significant effect of fibre-type grouping (P<0.05). The Tukey post-hoc analyses for isometric force, SEMG amplitude and MPF found significant differences between all fibre composition types and the measurement of each of these variables. The findings of this condition are presented in Table 2-4 below.

Condition 2 Results			
Fibre Comp.	**Mean Isometric Force at Maximum Excitation (Normalized to Fast Comp.)**	**SEMG Amplitude (Normalized RMS)**	**MPF (Hz)**
Slow	53.66% (±2.42)	70.37 (±2.36)	50.16 (±0.22)
Average	72.98% (±3.52)	81.79 (±1.51)	50.42 (±0.17)
Fast	100.00% (±2.01)	100.00 (±2.00)	50.93 (±0.21)

Table 2-4. Results of Condition 2 where fibre composition was varied through changes in the TF and CT.

Condition 3: Varying of the CT and CV

Conduction velocity was varied in condition 3 with the results reported in Table 2-5 below. The ANOVAs for isometric force, MPF and signal RMS each revealed a significant effect of fibre-type grouping ($P<0.05$). The Tukey post-hoc analyses showed a significant difference for the isometric force and MPF means between each fibre type composition ($P<0.05$). Both mean isometric force and MPF increased as fibre type composition became predominately fast.

In this condition, SEMG amplitude was found to be lower for average and fast than slow fibre type compositions. The Tukey post-hoc analysis for SEMG amplitude showed a significant difference only between the slow and average fibre type compositions ($P<0.05$). The SEMG amplitude was not significantly different between average and fast fibre type compositions($P=0.6939$).

Condition 3 Results			
Fibre Comp.	**Mean Isometric Force at Maximum Excitation (Normalized to Fast Comp.)**	**SEMG Amplitude (Normalized RMS to Slow Comp.)**	**MPF (Hz)**
Slow	68.67% (±2.17)	100.00 (±4.94)	49.95 (±0.26)
Average	79.14% (±2.18)	90.89 (±6.36)	50.23 (±0.18)
Fast	100.00% (±1.67)	87.60 (±4.56)	50.93 (±0.09)

Table 2-5. Results of Condition 3 where fibre composition was varied through changes in the CT and CV.

DISCUSSION

This research attempted to examine how one might expect a SEMG signal to change given changing physiological parameters and also the ability of the simulation model to correspond to the literature in this field. To vary conduction velocity and twitch force in conjunction with fibre type is a dangerous supposition given the lack of conclusive evidence in this area. Nonetheless, we can examine these signal changes to help further understand this model and to help in the event that these physiological properties become better understood.

In varying the contraction time and the possibly associated variables of twitch force and conduction velocity to account for different fibre type distributions, the simulated SEMG signals were found to vary in ways both expected and unexpected given a survey of the literature. We now discuss how the variation of these parameters affected the signal and how these changes compare to previous research.

Variation of the CT

Contraction time is the parameter responsible for the classification of motor units as fast or slow twitch. Thus, any variation of fibre distribution necessarily included a variation in contraction time assignments. In the Fuglevand model, motor unit twitch force F_i has the following proportion to contraction time T_i:

$$F_i \propto \frac{(1 - e^{-T_i^3})(e^{1-\frac{1}{T_i}})}{T_i^2} \qquad (2.3)$$

While Figure 2-7 does not illustrate a component of the model, it illustrates the implications of the relationship seen in equation 2.3. Namely, as the assignment of contraction time changes, so does the peak twitch force that unit produces. The motor units simulated in this study fall on this curve aside from some scaling differences. This relationship is similar to that reported by Fuglevand (1993).

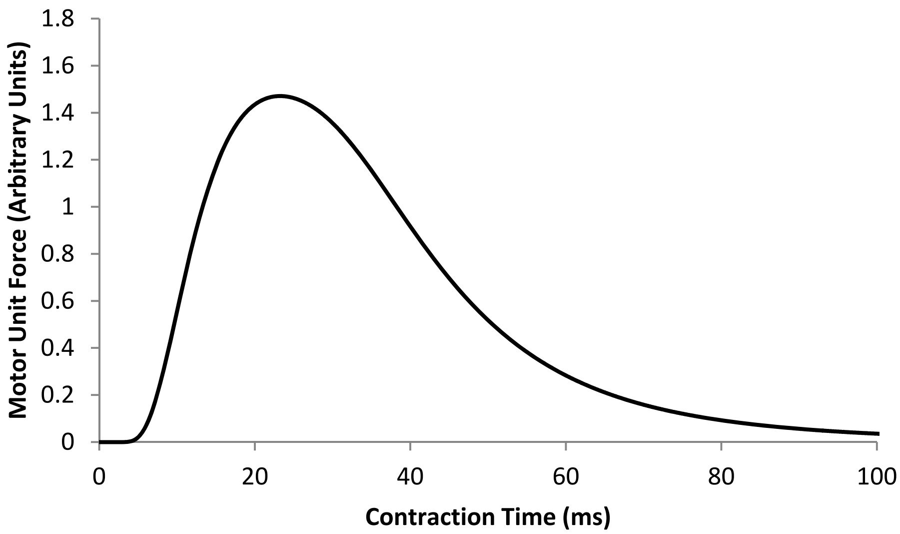

Figure 2-7. Contractile force versus contraction time. Fast twitch muscles with CT's of around 30ms tend to produce the most force in these simulations.

Recalling that motor units were not assigned contraction times any less than 30ms we can see that the simulated unit's lie on the decaying portion of this force-fibre type curve. This helps to explain the increases in force production seen in each trial for compositions tending towards predominantly type II fibres. However, this is the only variable measured which contraction time variations had any impact on, as we see no effect of differing fibre type proportions on the frequency spectrum of the signal or signal amplitude – an expected result as CT is independent of the SEMG component of this code. This does not match with experimental findings as signal amplitude and force production are related (Hagberg, 1989), thus raising serious concerns about the simulation model's ability to align with the literature if CT is varied alone.

Variation of the TF

Twitch force was varied alongside CT in Condition 2. It might seem intuitive that increased average TF should increase total simulated muscular force, and indeed this result is seen here. Similarly to experimental findings TF and signal amplitude are increasing commensurately, in contrast to Condition 1 (Wilmore, 2008). Furthermore, MPF increased with fast fibre proportions with no fibre diameter variations, which supports the seemingly controversial work of Gerdle (discussed in the preceding chapter, 2000; 1988).

That twitch forces influenced the SEMG signal requires an explanation as TF is a seemingly independent variable to the simulated SEMG signal within this model. To see why TF did influence RMS SEMG, we note first that twitch force assignment in this model plays a crucial role in how fibres are assigned to motor units:

$$nf_i = \left({}^{nf_{total}}/_{P_{total}}\right) \cdot P_i \tag{2.4}$$

Here, nf_i is the number of fibres innervated by a motor unit, nf_{total} represents the total muscle fibre count, P_i the twitch force of the motor unit i and P_{total} the sum of all motor unit twitch forces. This relationship did not automatically assign more fibres to motor units of higher twitch force as the ratio of P_i to P_{total} was also altered. The cumulative sum of muscle fibres innervated by each motor unit is depicted in Figure 2-8.

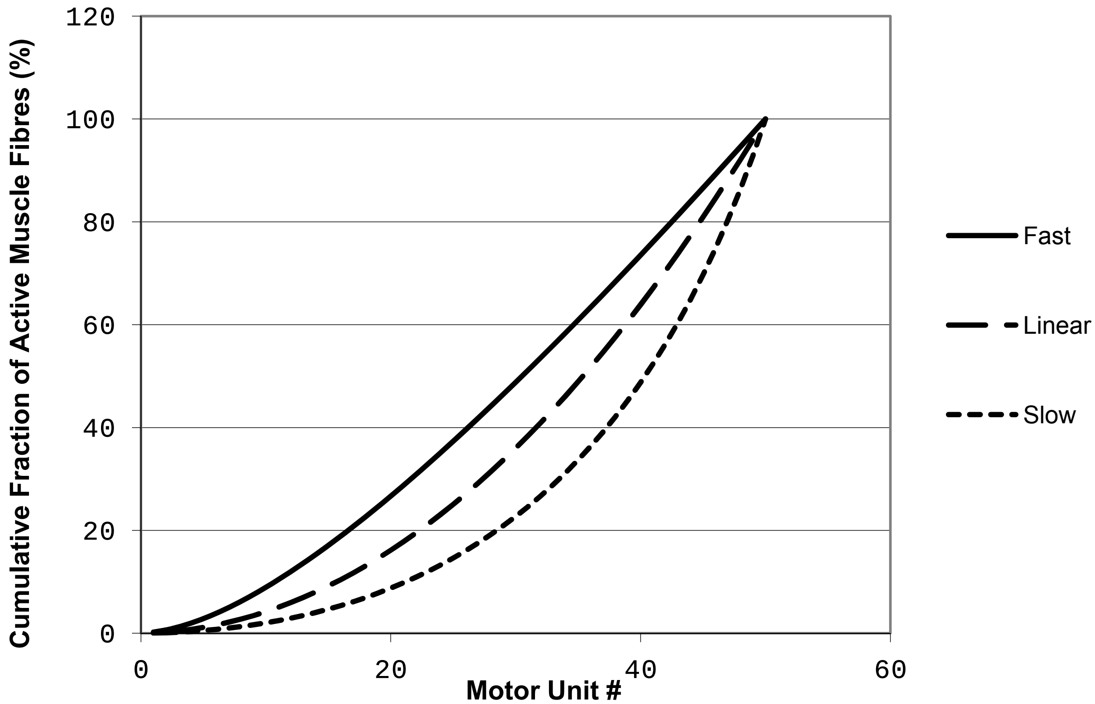

Figure 2-8. Cumulative sum of activated muscle fibres for each trial.

Discharge rates were left fixed across all conditions, yet during the ramp phase in neural excitation more fibres were innervated which contributed to a higher RMS SEMG. This corresponds with the literature in that force production differences between motor units is mostly due to an increased number of innervated fibres in a fast twitch motor unit (Milner-Brown, 1973). Thus the RMS SEMG results are slightly misleading as the signal strength was only different in the ramp phase.

That twitch force assignments acted to increase the signal frequency content is also explained by having more active fibres active during the first phase of excitation. Although the differences across the conditions was statistically significant, we can also note that these values may not be significant in a typical research setting which could expect changes across a few Hz (Arendt-Nielsen, 1989).

Of the trials performed, Condition 2 demonstrated the most accurate change in a SEMG signal in comparison to experimentally findings, although the results for RMS SEMG amplitude and MPF are slightly misleading. In addition to only minor changes in MPF, typically signal frequency content changes have been correlated to fibre cross-sectional area (Kupa, 1995). Since fibre diameter was fixed, the increase in MPF for fast twitch motor units seen here does not relate to the majority of research for frequency content changes (Gabriel, 2009).

Varying of the CV

Similarly to twitch force, conduction velocity is an often studied parameter when it comes to distinguishing muscle fibre types. CV has been found to be positively correlated to MPF, which may or may not be related to fibre type (Troni, 1983). We can also recall that the connection of CV to fibre type again hinges on the belief that fibre types have different diameters. Nonetheless, the link between CV, fibre type and MPF (the possibility of increased CV for fast twitch units implying increased SEMG MPF) is one of the most studied connections in the hope of allowing researching to examine muscle composition in vitro (Kupa, 1995). Indeed, the results of Trial 3 suggest that conduction velocity affects the simulated

MPF in the way that increasing type II fibre proportions (increased CV's) cause increases in the signal frequency content. Although CV acted to decrease amplitude as seen, increased CV acted to make the action potentials narrower. Therefore MPF increased for fast fibres for the same reasons as described elsewhere (Fuglevand, 1992). Across a broader range of CV`s, the relationship between MPF and CV can be seen in Figure 2-9.

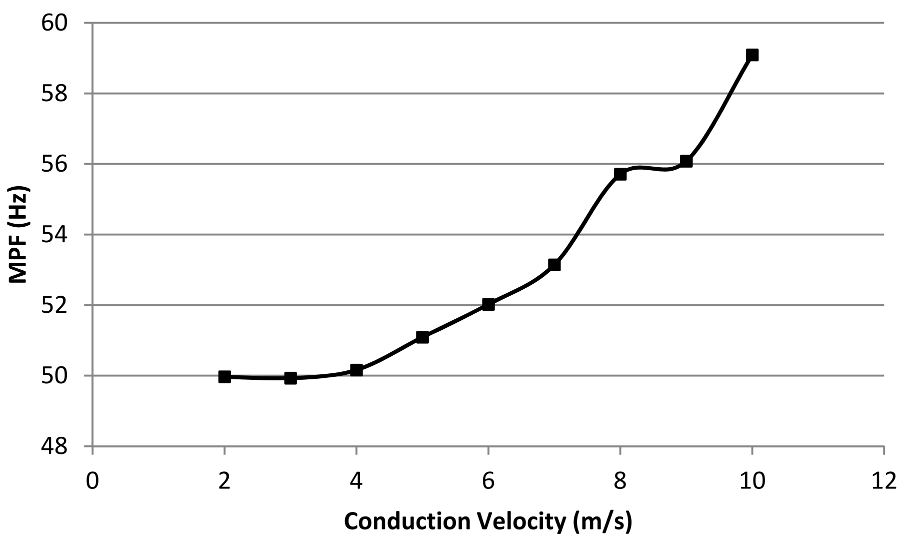

Figure 2-9. MPF versus conduction velocity. This figure shows this relationship on a larger scale than that simulated in trial 3.

CV is not proportional to the isometric force in the model code and so it was expected that the force results were the same as in Trial 1. However, CV curiously acted to make the signal amplitude inversely proportional to force production. The following equation helps to give an explanation for this finding:

$$\Phi(t) = \frac{I}{4\Pi\sigma_r}\left[\frac{1}{\sqrt{r_f^2 \cdot \sigma_z/\sigma_r + (CV \cdot t - z_e)^2}} - \frac{1}{\sqrt{r_f^2 \cdot \sigma_z/\sigma_r + (CV \cdot t - z_e + b)^2}}\right] \tag{2.5}$$

As we can see, increases in CV act to reduce the amplitude of a motor unit action potential, $\Phi(t)$. As the SEMG signal was a summation of these potentials, the incorrect inverse relationship between amplitude and force can be understood. It is worth mentioning that the post-hoc analysis found no significant difference between the SEMG amplitude mean values for the average and fast compositions. Increasing CV values act to dominate the denominators of (2.5), thus making the difference between the subtracted quantities smaller at sufficiently high CV values. Thus, as CV is increased we expect to see not only smaller amplitudes, but smaller differences to SEMG amplitude for these changes.

The results of this trial are interesting for a few reasons. One, CV increases with larger fibre diameters (Blijham, 2006), yet given the fixed diameters in this study, we still see increases in MPF. Perhaps this supports the conclusions of Gerdle that MPF could be linked directly to fibre type and not fibre diameter (2000). However, the fact that EMG and force were found to be inversely proportional does not correspond to experimental findings raising the same concerns as in Condition 1.

CONCLUSION

The Fuglevand model, which is the basis for many SEMG simulation strategies, allows researchers to study parameters commonly associated with muscle fibre type variations in more detail. Contraction time may be varied within the model with or without varying other parameters classically associated to fibre type such as conduction velocity and twitch force. Motor unit type distributions can thus be altered to represent a muscle with predominantly slow or fast twitch fibres. When contraction times and fibre twitch forces were varied together the simulated results are similar to those seen in some experiments. However, both twitch forces and conduction velocities are controversially linked to fibre type due to the unsubstantiated belief that fast twitch fibres have greater cross-sectional diameters. While this may indicate the need for model revisions, more research is certainly needed to assess the possible relationship of fibre type to the properties of twitch forces and conduction velocities before these and other simulation results can be used to more accurately compare to experimental research.

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CHAPTER 3:

SUMMARY, CONCLUSIONS, AND IMPLICATIONS

SUMMARY

The purpose of this study was an attempt in working towards understanding if surface electromyography (SEMG) can be used as a tool to evaluate muscle composition. It is speculated that the SEMG mean power frequency (MPF) could be increased for muscles with increased fast twitch motor unit compositions. However, there are several limitations to using SEMG in order to assess the composition of muscle fibres. For one there is uncertainty over how conduction velocity and twitch force correlate to motor unit fibre type, which is defined by contraction time (Farina, 2008). Secondly, SEMG is an impractical tool for making estimations on physiological properties due to complications in using the tool itself. Because of these difficulties, SEMG simulation models has at times been used to help in the analysis and prediction of SEMG data. In this research, a SEMG simulation model was used to run three trials in an attempt to study the effects motor unit contraction timing, twitch forces, and conduction velocities have on SEMG signals. Force production, signal amplitude and mean power frequency were examined for each trial.

The simulated SEMG signals varied under changes to each variable. The trial that seemingly produced the most similar results to experimental findings was when both contraction time and twitch force assignment were varied. In this trial, SEMG amplitude, isometric force and MPF increased when the muscle composition tended towards faster twitch motor units. Unfortunately, these results were a direct result of how the muscles were being excited neurologically in the ramping phase of the excitatory drive. Once excitation was at a maximum, the variations of this trial did not have an impact on the SEMG signal. This isn't useful with regards to our original question, as MPF does not vary. On the other hand, a positive result of this study was the illustration that MPF does change in proportion to conduction velocity.

RESEARCH CONCLUSIONS

This is the first work to examine the effect of these parameters on simulated SEMG signals. This was an important study as researchers have wishfully examined SEMG signals in the hope of finding characteristics of the signal that will give an estimation of muscle composition. Unfortunately, without firm experimental evidence on the true characteristics of motor units, the conclusions of this research are limited. However, these results indicate and support the idea that SEMG MPF could theoretically be used to estimate muscle fibre type concentrations.

To summarize, in this research we have found two major conclusions:

1) Changes in contraction time and motor unit peak twitch forces impacts SEMG in the time domain but not in the frequency domain.
2) This simulation shows that motor units that have a higher conduction velocity will also have a higher mean power frequency.

IMPLICATIONS

This research does not directly foster improvements in the understanding of muscle physiology nor does it directly lead to SEMG simulation model extensions; however, for these fields this work allows us to draw vital conclusions and ask important questions. It is these questions that could help to answer whether SEMG could be a tool to estimate muscle compositions which is the question overlaying the intent of this work.

For one this work highlights the necessity of resolving how fibre diameters vary in accordance with different fibre types. The literature indicates a number of studies that show that conduction velocities and twitch forces have been found to be positively correlated to fibre diameter (Blijham, 2006; Krivickas, 2011). Understanding these proportionalities could help to link these characteristics to fibre type. Unfortunately, some investigations have shown that, at least in the case of conduction velocity, fast twitch fibres may have higher conduction velocities independent of fibre diameter (Gerdle, 2000). Therefore, it might be necessary for research to circumvent study these two parameters' relationships to fibre diameter and instead evaluate them for fibre type itself.

Secondly, while strong connections between conduction velocity and SEMG spectral properties does exist – and is supported by this research – more precise functions mapping the two values could be useful. However, experimentally this could still be very difficult as SEMG certainly has several limitations both in experimental setups and in analyzing data. The biggest obstacle is highlighted in the work by Keenan which found that 7% of the motor unit potentials registered by a SEMG electrode can be responsible for upwards of 50% of the signal amplitude (2006).

Lastly, SEMG simulation models have been used in several studies to help in the analysis of data and to predict signal qualities for a variety of settings (Farina, 2004). This research indicates that an extension to incorporate different fibre type compositions is needed. The results of varying motor unit contraction times found in this investigation does not show similar trends to that seen in the literature as increases in force production are not met with changes in SEMG amplitude (Hagberg, 1989). Researchers interested in adapting the model for this purpose could use the methods and results of this thesis as a starting point.

It is the author's opinion that resolution of the first two aforementioned points would allow for muscle composition estimates for particular muscles through the use of SEMG and a subsequent frequency content analysis (namely, small muscles that lie directly beneath the skin) – a would-be marvellous achievement that is highlighted in and supported by the work of this thesis.

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APPENDIX A:

RESULTS OF CONDITION 1: FIBRE DISTRIBUTION ACCOUNTED FOR THROUGH VARYING MOTOR UNIT CONTRACTION TIMES

Below are the tables of results for condition 1. Data was normalized in the Tables A-1 and A-2 to the mean of the fast trial.

Isometric Force (Condition 1: Varying CT) (Normalized Units)			
Trial #	**Slow**	**Average**	**Fast**
1	67.63	77.78	99.71
2	66.81	80.26	99.21
3	65.76	76.91	99.78
4	69.64	84.45	97.57
5	67.23	82.18	100.63
6	68.93	79.62	100.67
7	67.48	80.40	100.35
8	69.38	78.32	97.87
9	70.13	80.61	101.12
10	73.40	79.92	103.44
Mean	68.67	79.14	100.00
St. Dev.	2.17	2.18	1.67
P Value		<0.001	

Table A-1. Results of Condition 1 fibre distribution versus isometric force production simulations.

RMS SEMG Amplitude (Condition 1: Varying CT) (Normalized Units)			
Trial #	**Slow**	**Average**	**Fast**
1	100.95	96.28	103.00
2	94.53	101.69	101.40
3	94.58	99.74	100.80
4	102.04	97.05	99.58
5	103.78	102.27	99.72
6	94.96	102.31	98.55
7	105.91	96.46	98.78
8	98.59	103.64	98.76
9	104.74	101.95	99.10
10	101.09	100.18	100.25
Mean	100.12	100.16	100.00
St. Dev.	4.27	2.69	1.41
P Value		0.992	

Table A-2. Results of Condition 1 fibre distribution versus signal amplitude simulations.

SEMG MPF (Condition 1: Varying CT)			
Trial #	**Slow**	**Average**	**Fast**
1	50.35	50.44	50.45
2	50.24	50.32	50.42
3	50.51	50.58	50.45
4	50.40	50.30	50.44
5	50.39	50.42	50.48
6	50.41	50.43	50.42
7	50.32	50.35	50.22
8	50.50	50.27	50.48
9	50.34	50.46	50.34
10	50.38	50.47	50.41
Mean	50.38	50.40	50.41
St. Dev.	0.08	0.09	0.08
P Value		0.718	

Table A-3. Results of Condition 1 fibre distribution versus mean power frequency simulations.

APPENDIX B:

RESULTS OF CONDITION 2: FIBRE DISTRIBUTION ACCOUNTED FOR THROUGH VARYING MOTOR UNIT CONTRACTION TIMES AND TWITCH FORCES

Below are the tables of results for condition 2. Data was normalized in the Tables A-1 and A-2 to the mean of the fast trial.

Isometric Force (Condition 2: Varying TF & CT) (Normalized Units)			
Trial #	**Slow**	**Average**	**Fast**
1	52.89	71.18	97.55
2	54.33	76.05	100.57
3	50.38	71.45	102.30
4	58.09	79.51	101.11
5	56.38	72.86	97.90
6	54.72	74.05	97.36
7	50.51	75.32	99.17
8	53.56	70.81	99.51
9	53.80	66.55	101.06
10	51.92	72.03	103.19
Mean	53.66	72.98	100.00
St. Dev.	2.42	3.52	2.01
P Value		<0.001	

Table B-1. Results of Condition 2 fibre distribution versus isometric force production simulations.

RMS SEMG Amplitude (Condition 2: Varying TF & CT) (Normalized Units)			
Trial #	**Slow**	**Average**	**Fast**
1	66.57	82.87	99.78
2	71.02	82.59	99.44
3	69.15	82.33	102.38
4	72.58	79.42	101.90
5	71.96	81.68	100.37
6	72.42	83.69	97.69
7	71.97	82.42	100.84
8	67.21	83.02	96.11
9	68.18	79.58	99.93
10	72.61	80.29	102.27
Mean	70.37	81.79	100.00
St. Dev.	2.36	1.51	2.00
P Value		<0.001	

Table B-2. Results of Condition 2 fibre distribution versus signal amplitude simulations.

SEMG MPF (Condition 2: Varying TF)			
Trial #	**Slow**	**Average**	**Fast**
1	49.80	50.77	50.80
2	50.19	50.35	50.79
3	50.24	50.34	51.07
4	50.09	50.57	50.48
5	49.92	50.46	51.25
6	50.12	50.34	50.94
7	50.47	50.24	51.05
8	50.31	50.43	50.88
9	49.99	50.49	51.01
10	50.44	50.21	51.07
Mean	50.16	50.42	50.93
St. Dev.	0.22	0.17	0.21
P Value		<0.001	

Table B-3. Results of Condition 2 fibre distribution versus mean power frequency simulations.

APPENDIX C:

RESULTS OF CONDITION 3: FIBRE DISTRIBUTION ACCOUNTED FOR THROUGH VARYING MOTOR UNIT CONTRACTION TIMES AND CONDUCTION VELOCTIES

Below are the tables of results for condition 3. Data was normalized in the Tables A-1 and A-2 to the mean of the fast trial.

Isometric Force (Condition 3: Varying CV & CT) (Normalized Units)			
Trial #	**Slow**	**Average**	**Fast**
1	67.63	77.78	99.71
2	66.81	80.26	99.21
3	65.76	76.91	99.78
4	69.64	84.45	97.57
5	67.23	82.18	100.63
6	68.93	79.62	100.67
7	67.48	80.40	100.35
8	69.38	78.32	97.87
9	70.13	80.61	101.12
10	73.40	79.92	103.44
Mean	68.67	79.14	100.00
St. Dev.	2.17	2.18	1.67
P Value		<0.001	

Table A-1. Results of Condition 3 fibre distribution versus isometric force production simulations.

RMS SEMG Amplitude (Condition 3: Varying CV & CT) (Normalized Units)			
Trial #	**Slow**	**Average**	**Fast**
1	98.88	88.95	85.86
2	103.56	92.97	89.00
3	90.94	100.15	92.27
4	101.71	102.22	96.50
5	99.55	88.78	80.98
6	102.01	89.85	92.93
7	107.87	93.13	87.02
8	101.73	84.71	91.04
9	101.03	84.20	89.15
10	98.19	83.90	85.27
Mean	100.00	90.89	87.60
St. Dev.	4.94	6.36	4.56
P Value		<0.001	

Table A-2. Results of Condition 3 fibre distribution versus signal amplitude simulations.

SEMG MPF (Condition 3: Varying CV & CT)			
Trial #	**Slow**	**Average**	**Fast**
1	49.71	50.40	51.11
2	50.30	50.33	50.97
3	49.83	50.31	50.89
4	49.74	50.19	50.96
5	49.49	49.99	50.93
6	50.06	50.18	50.85
7	50.22	50.33	50.81
8	49.96	50.54	50.92
9	49.98	50.13	51.01
10	50.18	49.95	50.89
Mean	49.95	50.23	50.93
St. Dev.	0.26	0.18	0.09
P Value		<0.001	

Table A-3. Results of Condition 3 fibre distribution versus mean power frequency simulations.

APPENDIX D:

EQUATIONS OF MOTOR UNIT AND SEMG MODEL

Using the following system of equations, inputted into Matlab, the simulations of this paper were carried out. This represents a more extensive outline than the flowchart shown in Chapter 1.

Motor Unit Force Model	
Motor Unit Recruitment Threshold (for motor unit i with constant a)	$RTE(i) = e^{a \cdot i}$
Motor Unit Discharge Rate (for motor unit i, time t, excitatory drive $E(t)$ and minimum firing rate $MFR(i)$)	$FR(i,t) = [E(t) - RTE(i)] + MFR(i)$
Time of j'th Discharge of i'th Motor Unit	$ISI\ (i,j) = t_{i,j} - t_{i,j-1} = \frac{1}{FR(i,t)}$
Peak Twitch Force (with constant b)	$P(i) = e^{b \cdot i}$
Contraction Time	$T(i) = 90 \cdot \left(\frac{1}{P(i)}\right)^{1/4.2}$
Gain Factor	$g(i,j) = \frac{1 - e^{-2(T(i)/ISI(j))^3}}{T(i)/ISI(j)}$
Motor Unit Twitch Force	$f(i,j) = g(i,j) \cdot \frac{P(i) \cdot t}{T(i)} \cdot e^{1-(t/(T(i))}$

Sum Motor Unit Twitch Force (k Discharges)	$F(i,t) = \sum_{j=1}^{k} f(i,j)(t - t_{i,j})$
Sum Muscle Force	$F_M(t) = \sum_{i=1}^{n} F(i,t)$
SEMG Simulation Model	
Measurement of Action (Electric) Potential	$\varphi(t) = \frac{I}{4\Pi\sigma}\left[\frac{1}{r_1} - \frac{1}{r_2} - \frac{1}{r_3} + \frac{1}{r_4}\right]$
Sum Motor Unit Action Potential for n_e Isopotential Layers of n_f Total Muscle Fibres	$muap(t) = \sum_{j=1}^{n_e} \varphi_j(t)$

Table D-1. Motor unit and SEMG model equations.