Changing Motor Unit Firing Frequencies With Force In The Human Biceps Brachii

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Abstract. An sEMG model of the human biceps brachii muscle has been implemented and populated with experimentally obtained motor unit parameters. Of particular interest are the differences between type I and type II muscle fibre characteristics. Needle electrode studies reveal that in addition to different action potential conduction velocities, the different fibre types have specific ranges of recruitment thresholds and firing-frequencies. In order to assess the effect of these parameters with force, simulations were conducted over a range of force levels from 30 – 80% of the maximum voluntary contraction (MVC). In addition, experiments were conducted on human participants to verify the accuracy of the implemented model to experimental results.

1 Introduction

The surface electromyogram is a composite signal resulting from the summation of the induced electrical activity of all active motor units in the recording field. These motor units can be type I (slow) muscle fibres or type II (fast) muscle fibres. The two types of muscle fibre have varying characteristics including conduction velocity, size, fatigability and action potential firing rates. Despite these established facts [1], many sEMG studies, particularly modelling studies, assume average values without considering the motor unit types.

Needle electrode studies have determined that firing frequencies of type I and type II fibres are significantly different, and that the muscle force at which the motor units are recruited varies. In low-level isometric contractions of the biceps brachii, Gydikov found that only type I fibres were activated, and that they were firing at an average rate of 8-14 Hz. At higher force levels (greater than 40% of the maximum voluntary contraction), the type II fibres were also active, at firing rates of 12.5 to 24.5 Hz, increasing with force [2].

The study presented here investigates the amplitude and spectral features of simulated sEMG signals from a model populated with non-linear firing frequency and recruitment threshold parameters, based on Gaussian distributions of the values reported by Gydikov. The objective is to assess whether such an implementation improves the accuracy of the sEMG model as compared to experimental results. Such an improvement in model accuracy could enhance the understanding of the mechanics of sEMG signal generation and the usefulness of sEMG models for research and practical applications.

2 Methodology

2.1 SEMG Model

The sEMG model implemented in this work models the surface EMG signal using motor unit (MU) equations developed by Rosenfalck [3] and Merletti [4]. The novelty of this implementation is two fold. The first is the simulation of a whole muscle signal, created by the interaction of multiple MUs of varying types. The second novelty of this implementation is that it has been populated based on experimentally obtained physiological parameters.

Fig 1 shows the various parameters that have been considered in this sEMG model and the influence of these parameters on the simulated sEMG signal. The neural stimulating pulse generates an action potential (AP) in the muscle fibre, which propagates along the length of
the fibre with a current distribution defined by $I_m(t)$. When recorded at the surface of the skin, this current distribution is influenced by $f(t)$, a function which incorporates the depth of the muscle fibre, the distance (along the length of the fibre) between the AP and the sensor, and the conductivity of the tissue. The convolution of these parameters gives the surface potential resulting from a single muscle fibre. When this potential is multiplied by $K_m$ (MU size), it gives the surface potential due to a MU.

The sEMG signal generated by this model can be described by;

$$sEMG = \sum_{m=1}^{m} K_m \cdot f(t) \cdot I_m(t) \cdot \delta\left(t - \frac{r}{r_m} - \tau_m\right) + \text{noise}$$

(1)

where sEMG is the signal recorded at the skin’s surface, m is the number of active MUs in the muscle, $K_m$ is the size of the MU, $I_m$ is the current distribution of a muscle fibre action potential [3] and the delta function simulates the neural input signal. An initial temporal offset, $\tau$, models the time before the start of MU firing. In a voluntary muscle contraction, the interval between subsequent firings is variable, even when the average firing frequency is constant. This is modelled in Eq. 1 by variation in the firing rate, $r$, such that $r = r + \delta r$ where $\delta r$ is the variation of this rate.

The function $f(t)$ in Eq. 1 incorporates the attenuation of the MU signal by the tissue through which it passes. $f(t)$ is influenced by the distance along the fibre $z$, the radius of the fibre $x$ and the depth of the fibre $y$ from the skin’s surface [5].

$$f(t) = \frac{1}{4\pi\sigma_e} \sqrt{\frac{1}{(z-z')^2 + \sigma\left[(x-x')^2 + (y-y')^2\right]}}$$

(2)

$\sigma_e$ is the conductivity of the external medium, $\sigma$ is the ratio of the internal muscle fibre
conductivity and the external conductivity, such that:

\[
\sigma = \left( \frac{\sigma_r}{\sigma_e} \right)^2
\]  

(3)

The sEMG model was populated with the muscle parameters outlined in [6], and shown in Table 1. Of particular interest are the different MU conduction velocities for fast and slow fibre types.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value for biceps simulation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Number of motor units (MU)</td>
<td>100</td>
</tr>
<tr>
<td>Average conduction velocity m.s(^{-1}) [13]</td>
<td>4.3 ± 0.29</td>
</tr>
<tr>
<td>Conduction velocity (fast fibres) [14]</td>
<td>4.9 ± 0.3</td>
</tr>
<tr>
<td>Conduction velocity (slow fibres)</td>
<td>3.9 ± 0.3</td>
</tr>
<tr>
<td>Percentage of type 1 fibres (%) [12]</td>
<td>42 - 50</td>
</tr>
<tr>
<td>Muscle fibre diameter [5]</td>
<td>25 µm</td>
</tr>
<tr>
<td>Depth of MU from surface [15]</td>
<td>35 mm ± 2 mm</td>
</tr>
<tr>
<td>Duration of AP along fibre [16]</td>
<td>16 mm</td>
</tr>
<tr>
<td>Cutaneous tissue [15]</td>
<td>Single, 3mm isotropic layer</td>
</tr>
<tr>
<td>Muscle half-fibre length [14]</td>
<td>65 mm</td>
</tr>
<tr>
<td>Simulation sampling frequency</td>
<td>10000 Hz</td>
</tr>
</tbody>
</table>

Tab 1. Biceps sEMG simulation parameters

In previous models, the AP firing frequencies of the motor units have been distributed about a common mean value. Thus, the slow and the fast fibre types have been assigned the same range of firing frequencies values. However, experiments have shown that the firing frequencies of type I muscle fibres are lower than those of type II. In addition, the force at which the MU is recruited is also dependent on the MU type.

The implementation reported in this paper has incorporated recruitment thresholds and variable MU firing frequencies developed from Gaussian distributions and based on the results described by Gydikov in [2]. The firing frequencies of type I and type II fibres are significantly different, and the muscle force at which the motor units are recruited is dependent on fibre type. At low force levels only type I fibres are active, with firing rates distributed about averages of 8-14 Hz. At higher force levels, type II fibres are also active, at firing rates of 12.5 to 24.5 Hz, increasing with force. A sample of the firing rates and recruitment thresholds of a set of MUs is shown in Fig 2.

In this model implementation, the ratio of type II fibres was set to 0.4 at 100% MVC [7]. The model was simulated at 3 different force levels from 30 to 80% MVC. At each force level, 10 seconds of data was simulated 3 times. For each data set, the root mean square (RMS), mean power frequency (MPF) and zero crossings per second (ZC) were calculated. These signal features were then averaged across the three trials for each force level.

2.2 SEMG Experiments

The model was experimentally verified. Three healthy male subjects, with no history of neuromuscular disease or injury, completed the following experimental protocol, approved by the RMIT Human Research Ethics Committee.

Prior to affixing the electrodes, the skin above the biceps brachii muscle was prepared by abrading and cleaning with alcohol. The sEMG was recorded using single channel differential surface electrodes fixed 10mm apart, on the biceps brachii muscle. The electrodes were placed on the line between the antecubital fossa and the acromion process, at 1/3 from

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During the experiment, the participants were seated in a sturdy chair with their feet flat on the floor. The upper arm was rested on the surface of a desk, in a horizontal position with the palm facing upward. The elbow was fixed at 90 degrees, with the fingers in line with a wall mounted force sensor attached to a wrist strap held in the hand. The subject was asked to pull their fingers back towards the forearm (resisted by the wrist strap), resulting in isometric muscle flexion. The maximum voluntary contraction (MVC) was determined by the average of three trials of maximum exertion, measured by a force sensor.

Each subject was asked to perform isometric contractions for 15 seconds. This was repeated three times for each of the three contraction levels. Of the 15 s of recorded data, the 10 s with the most stable force level was extracted for analysis.

3 Results

Signal features that are frequently used to characterise the sEMG signal were calculated from both the experimental and the simulated data. The root mean square (RMS) is a measure of the amplitude of the signal and is known to increase with force during non-fatiguing contractions. From the results, it is observed that in both the simulated and experimental data sets obtained here, the RMS increased linearly with force in the range of 30% to 80% MVC. For the experimental results, the gradient of the %MVC/RMS trend lines for each of the three subjects are 0.0018, 0.0019 and 0.0024, all with $r^2$ values above 0.97. For the associated simulated results, the gradient of the %MVC/RMS trend lines for each of the three subjects is 0.18, with $r^2$ values above 0.93. Fig 3 and Fig 4 show the average RMS of the experimental and simulated data at each force level for subjects 1 and 2.

The spectral characteristics of the sEMG are represented by the signal features MPF and ZC. The MPF of both the experimental and simulated data show a slight increase with force, with a range of 6.9 Hz for the experimental data and 2.6 Hz for the simulated data. The MPF values of the experimental data were also slightly higher than of the simulated data.

Similarly, the ZC signal feature shows little variance with force, in both the simulated and experimental data sets.
Fig 3. Subject 1: Average RMS values at each force level for experimental and simulated data

Fig 4. Subject 3: Average RMS values at each force level for experimental and simulated data

4 Discussion

The RMS values for both the simulated and experimental data sets increase linearly with force between 30-80% MVC. The rate of change of RMS is similar in both cases. The difference between experimental and simulated data are most likely due to the varying size of type I and type II motor unit types, which have not been implemented in this sEMG model. However, the results are similar enough to suggest that this recruitment threshold model is
accurate to experimentally acquired data. In particular, it is noted that a non-linear recruitment pattern and firing frequency variation produced a linear force/amplitude relationship in both cases.

The lack of variance of MPF and ZC for both the simulated and experimental data sets suggests that changes in firing frequency alone do not have a large impact on these signal features. Rather, it is likely that changes in the conduction velocity of the muscle fibres will alter the sEMG spectrum, a proposal supported by other researchers [9]. As this study is conducted under non-fatigue conditions, the conduction velocity is constant and the model is time-invariant. The ZC rate of the experimental data set is slightly higher than the simulated data, suggesting that the experimental data is noisier. In future work, system noise will be incorporated into the model to test this assumption.

To fully investigate the impact of the relationship between motor unit firing frequency and force on the sEMG spectrum, signal features other than the MPF could be considered. Most importantly, studies of the time-variant case (i.e. during muscle fatigue) may allow spectral changes in the sEMG signal to be studied.

5 Conclusion

The implementation of recruitment thresholds and firing frequency variation with force from needle EMG studies has been the basis for the development of a new sEMG model that avoids a number of assumptions made by earlier models. The results of experimental verification of the model reveals that amplitude characteristics predicted by this model are verified by the experimental data. The spectral properties of simulated signals have been shown to have only a small variation with force, which is similar to experimental results.

This sEMG model displays accuracy which may improve the usefulness of sEMG models for research and practical applications. A model that can accurately predict the signal features is useful for developing targeted signal processing techniques as well as identifying signal variations due to disease or disorder.

References