An Accurate Biceps Muscle Model with sEMG and Muscle Force Outputs

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Abstract

A differential, time-invariant, surface electromyogram (sEMG) model has been implemented. The model uses realistic physiological parameter values to simulate both electrical sEMG and muscle force output signals. The combination of these signals is used to validate the accuracy of the model with respect to experimental results. The novelty of this sEMG model implementation is that it assigns more realistic distributions of variables to create life-like motor unit (MU) characteristics and defines individual parameter values to type I and type II muscle fiber types. Variables such as muscle fiber conduction velocity, jitter (the change in the inter-pulse interval between subsequent action potential firings) and motor unit size have been considered to follow normal distributions about mean values reported by experts in the field. In addition, motor unit firing frequencies have been considered to have non-linear and type-based distribution that is in accordance with reported experimental observations. Motor unit recruitment is also related to the motor unit type. The model has been simulated to predict single-channel differential sEMG signals and force outputs from voluntary, isometric contractions of the biceps brachii muscle. This model has been experimentally verified by conducting experiments on ten participants who performed isometric contractions of the biceps brachii at three different force levels. Signal features of the simulated signals were compared with experimental results. The simulated sEMG signals show similar values and rates of change of RMS to the experimental signals. To verify the accuracy of the model, the force output was simulated at varying contraction levels. This simulated force increased linearly at a comparable rate to the experimental force exerted.

Keywords: Electromyography (EMG), Model, Simulation, Time-invariant

1. Introduction

The electromyogram (EMG) is the recording of muscle electrical activity. Surface EMG (sEMG) is measured by electrodes affixed to the surface of the skin, above the muscle of interest. This measurement technique yields a composite signal influenced by all active muscle fibers in the electrode recording field. sEMG is used in neuromuscular diagnostics, ergonomics, and for prosthesis control. Currently, sEMG is largely used to determine binary states, such as high/low contraction strength or fatigued/not fatigued muscles. Attempts are being made to improve signal processing to obtain more information about the underlying muscles from sEMG such as time-variant fatigue indices and classification of subtle muscle actions. However, the relationship between muscle contraction under different conditions and associated changes in the sEMG signal is not completely understood, and hence the practical applications of sEMG are limited.

To increase the understanding of this relationship, and to predict sEMG corresponding to voluntary muscle contraction, sEMG models have been developed by a number of researchers. An important early step in the field of sEMG modelling was by Rosenfalck [1], who defined the equation for the action potential shape. This work has been utilized by the majority of subsequent modelling studies. Merletti extended this model by defining the current distribution equations for simulation of a single active motor unit, the location of each action potential in relation to the recording electrode and the influence of the cutaneous tissue on the signal [2,3].

More recently, a number of models have been presented in the literature, including the work of Farina et al., who presented a whole-muscle sEMG model with a normal distribution of conduction velocities about a realistic mean value [4,5]. Prior to this, sEMG models reported in the literature assumed single, average values rather than considering the statistical distribution of parameter values.
Each of these models has contributed to increase the understanding of the neuromuscular system by allowing simulation of experiments where variables can be altered in ways that are not possible or reliable in vivo. Existing models have been used to study surface electromyography as an experimental technique, including sEMG electrode placement [6] and the effect of subcutaneous fat on sEMG recording [7].

However, these models are based on limited assumptions regarding motor unit type, size, recruitment and distribution. In particular, to the knowledge of the authors, no model has been presented which defines individual conduction velocity ranges, firing frequencies and recruitment thresholds for different muscle fiber types. The objective of this work is to implement a sEMG model that simulates multiple muscle fiber types, is populated by physiologically accurate motor unit properties and is validated experimentally by both electrical and mechanical output signals.

To this end, a model that utilises lifelike values and distributions for muscle fiber conduction velocities, motor unit firing frequencies and motor unit recruitment patterns is reported here. To simulate a whole-muscle sEMG, the interplay between multiple motor units with distributed values for muscle fiber properties has been considered. In addition to the simulated electrical sEMG signal, the model outputs a force value, determined by the contributions of twitch forces from active motor units. Both the mechanical and electrical outputs of the simulation have been compared with experimental results.

2. Materials and methods

The differential sEMG model described above has been implemented. The precision of the model has been experimentally verified. Experiments were conducted on the biceps brachii muscles of ten healthy human participants, executing voluntary isometric contractions at varying force levels. The theory and procedures followed to implement the sEMG model and conduct human experiments are described here.

2.1 sEMG model theory

A single motor neuron and its associated multiple muscle fibers are referred to as a motor unit (MU). The motor unit action potential (MUAP) is the summation of all the muscle fiber action potentials (APs) from one MU.

The action potential of a single muscle fiber (measured by a differential bipolar electrode injected into the fiber) is approximately triphasic in shape, with a mean of zero. The MUAP has a similar shape, as it is generated by the superposition of the APs from each muscle fiber. The muscle fibers in a motor unit are dispersed throughout the muscle volume so a practical measurement of a MUAP with needles electrodes is not possible. Numerous techniques have been applied to extract the MUAP from composite sEMG, including signal decomposition [8] and the recording of sEMG with electrode arrays [9]. When recorded with a biphasic bipolar surface electrode, the sEMG is an interferential summation of many MUAPs, firing non-synchronously and attenuated by the muscular and cutaneous tissue that have low-pass filter characteristics to the electrical signals.

The frequency of a MUAP is dependent on the firing frequency of the neuron, and the amplitude is dependent on the number of active fibers in the MU and the type and size of these fibers. As multiple APs can superimpose upon one another, the amplitude of a MUAP is also dependent on the frequency of neural stimulation.

The principle factors that influence sEMG are the frequency of each MUAP (and the variation of this frequency), the amplitude and shape of each MUAP, the superposition of the various signals, the electrical properties of the cutaneous tissue and experimental noise. In addition, the electrode location in relation to the muscle fibers must be considered.

The model reported here considers the entire muscle with multiple MUs. The MU size, initial temporal offset and the muscle fiber conduction velocity have been considered to have normal distribution with mean and standard deviation values based on experimental results. The MUAP rates are based on experiments of Gydikov [10], and the differences in type I and type II fibers have also been considered. The MU recruitment order is based on the size principle, where smaller (slow-twitch) fibers are recruited initially, and larger (fast-twitch) fibers are recruited at higher contraction levels. This model is a more accurate representation of a real muscle.

The model covers four stages of EMG generation: neuronal stimulating pulse, muscle fiber action potential, motor unit action potential and surface EMG simulation.

2.2 sEMG model equations

The motor neuron communicates with the muscle fibers at the neuromuscular junction (NMJ). When a neural impulse is received, APs travel along the muscle fiber, from the NMJ towards each end of the muscle. The frequency of the stimulating pulse (signal from the motor neuron) determines the firing frequency of the motor unit and is given by an impulse function ($\delta$) in the time domain ($t$), where $n$ is the pulse number and $r$ is the firing rate:

$$\text{stimulus} = \delta \left( t - \frac{n}{r} \right) \quad (1)$$

In a voluntary muscle contraction, the interval between subsequent firings is variable, even when the average firing frequency is constant. This is modelled by variation in the firing rate, $r$ such that $r = r + dr$ where $dr$ is the variation of this rate. An initial temporal offset, $\tau$, models the time before the start of MU firing.

The action potential of a muscle fiber, $V_{m}$, generated by the stimulating pulse, is represented by the expression

$$V_m(z) = A(\lambda z)^3 e^{-iz} - B \quad (2)$$

where $A$ is the amplitude of the action potential, $B$ is the resting membrane potential, $\lambda$ is the scale factor in mm$^{-1}$ (influencing the AP pulse width) and $z$ is the distance (along the muscle fiber length) between the NMJ and the recording site. Previous studies have used values of 96 mV and -90 mV for $A$ and $B$. 


respectively to give a realistic size and shape of the signal typically recorded from a human muscle fiber [2]. This signal is then multiplied by scaling factors dependent on the MU size.

For each muscle fiber, the current distribution between the inside and outside of the muscle cell can be modelled as the second spatial derivative of the AP voltage, \( V_m \) [11]. The current distribution, \( I_m \), of the fiber action potential is therefore given by

\[
I_m(z) = C \frac{d^2 V_m(z)}{dz^2} = CA\hat{I}(\lambda z) \left( 6 - 6\lambda z + \lambda^2 z^2 \right) e^{-\lambda z} \tag{3}
\]

\( C \) is a constant that takes into account the conductivity of the axoplasm \( \sigma_r \), the muscle fiber conduction velocity, \( v \), and the diameter of the muscle fiber, \( d \) [1,12].

\[
C = \frac{d^2 \sigma_r \pi}{4v^2} \tag{4}
\]

In the work reported here, the conduction velocity of the muscle fibers is time-invariant.

\[
z = v \cdot t \tag{5}
\]

where \( v \) is the conduction velocity of the muscle fiber and \( t \) is time.

The current distribution can therefore be expressed in the time domain as;

\[
I_m(t) = CA(\lambda v)^2 (\lambda vt + \lambda^2 t^2) e^{-\lambda vt} \tag{6}
\]

The current distribution of a single fiber action potential over time, as modelled by Eq. (6), is shown in Figure 1.

![Figure 1. The current distribution of a single fiber action potential in the time domain.](image)

The attenuation of an MUAP is a function of the distance between the MU and the electrodes and tissue conductivity. A function \( f(t) \) is defined, which is influenced by the distance along the fiber \( z \), the radius of the fiber \( r \) and the depth of the fiber \( y \) from the skin’s surface [13] (Figure 2).

\[
f(t) = \frac{1}{4\pi\sigma_i} \frac{1}{\sqrt{(z-z')^2 + \sigma_r^2(z-r)^2 + (y-y')^2}} \tag{7}
\]

\( \sigma \) is the conductivity of the external medium, \( \sigma_i \) is the ratio of the internal muscle fiber conductivity and the external conductivity, such that:

\[
sEMG = \sum_{n=0}^{\infty} K_m \cdot f(t) \cdot I_m(t) \cdot \delta \left( t - \frac{n}{F} - \tau_m \right) + noise \tag{9}
\]

A single simulated motor unit is a superposition of the contributing action potentials from all the fibers in the MU. Each fiber fires almost simultaneously, meaning that an MUAP is of similar shape to the fiber action potential, with an amplitude that is dependent on the size of the motor unit (number of muscle fibers). In this model, any number of muscle fiber types can be defined, and each is assigned a corresponding average conduction velocity and distribution spread. This allows both slow and fast fibers to be modeled within a single muscle. The variable \( K_m \) is denoted to represent the size of the motor unit.

Figure 2 shows the components of the model. These components determine the final model equation used, given below as Eq. (9).

![Figure 2. Representation of sEMG model implemented. The neuron-stimulating pulse generates an AP in the muscle fiber, which propagates along the length of the fiber 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 fiber, the distance (along the fiber) between the AP and the sensor, and the conductivity of the tissue. The convolution of these parameters gives the surface potential of a single muscle fiber. When this potential is multiplied by \( K_m \) (MU size), it gives the potential of a motor unit.](image)
2.3 Model implementation

To simulate the model, values for each parameter were chosen from published data. The model parameters used for the biceps muscle are listed in Table 1.

Table 1. Parameters used for sEMG model simulation of the biceps brachii. For each parameter, the ± value represents the standard deviation of the distribution.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value for biceps brachii 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 [18]</td>
<td>4.3 ± 0.29</td>
</tr>
<tr>
<td>Conduction velocity (fast fibres) [19]</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 (%) [17]</td>
<td>42 – 50</td>
</tr>
<tr>
<td>Depth of MU from surface [20]</td>
<td>35 mm ± 2 mm</td>
</tr>
<tr>
<td>Duration of AP along fiber [21]</td>
<td>16 mm</td>
</tr>
<tr>
<td>Cutaneous tissue [20]</td>
<td>Single, isotropic, 3-mm layer</td>
</tr>
<tr>
<td>Muscle half-fiber length [19]</td>
<td>65 mm</td>
</tr>
<tr>
<td>Simulation sampling frequency</td>
<td>10000 Hz</td>
</tr>
</tbody>
</table>

Each motor unit was assigned a recruitment threshold and firing frequency, based on the fiber type. A Gaussian distribution was generated for the recruitment threshold and firing frequency, based on values reported by Gydikov from needle electrode studies [10]. Figure 3 shows a sample distribution of these values.

![Motor unit firing frequencies (pulses per second) with recruitment threshold (% MVC).](image)

2.4 Force output

The force output by a muscle is dependent on the number of active motor units and the size and firing rate of each of these MUs. The implementation of the force output here is based on the work by Fuglevand [14], who described a model with both sEMG and force outputs. As the muscle fibers in the biceps brachii run approximately parallel to the direction of muscle pull, the force can be modeled as a summation of the twitch force of each active MU in the MU pool.

The peak twitch amplitude, P, of each MU is related to its recruitment threshold and distributed such that the last recruited MU has a peak amplitude 100 times that of the first recruited MU. The contraction time of the twitch force is related to the peak amplitude, as described in [14]. The total force exerted during a contraction is the summation of the contributing force trains from each active MU.

2.5 Experimental procedures

Ten healthy male participants, with no history of neuromuscular disease or injury, completed the following experimental protocol, approved by the RMIT Human Research Ethics Committee. The experimental procedures conformed with the Helsinki Declaration of 1975, as revised in 2004.

The skin surface above the biceps brachii muscle was abraded and cleaned with alcohol. Delsys single-channel differential surface electrodes were used. These electrodes have two bipolar silver (99.9%) contacts (dimensions 10 mm × 1 mm) at a fixed inter-electrode distance of 10 mm [15].

The electrodes were placed on biceps, on the line between the antecubital fossa and the acromion process, at 1/3rd distance from the antecubital fossa [16]. The participant was 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 (S type force sensor -INTERFACE SM25). The force sensor was attached to a flexible steel wire, terminating in a ring that was placed around the participant’s hand. The subject was asked to pull their arm away from the force sensor, resulting in an isometric muscle flexion, resisted by the wrist strap. To determine each participant’s maximum voluntary contraction (MVC), three maximal contractions of 5 seconds were performed with 120-second rest time between contractions. The force transducer measured the force exerted. The average force exerted in each of the three contractions was averaged to give the MVC.

A graphical display gave feedback to the participant on the force being exerted. The participant was asked to contract to 30%, 50% and 80% MVC and hold each of these for 15-second intervals. Each contraction type was repeated three times, with enough rest time between contractions to ensure the muscles were not fatigued.

2.6 Data analysis

For each data set, the root mean square (RMS) of the sEMG was calculated. The RMS of an sEMG signal represents the magnitude and in the non-fatigued state, is associated with the force exerted by the muscle.

\[
RMS = \sqrt{\frac{\sum x^2}{N}}
\]  

(11)

The RMS of the simulated force output was also calculated.

3. Results

For both experimental and simulated data, three recordings were obtained at each contraction level. The RMS of sEMG and the force output were averaged across the three trials in each case. The magnitude of the simulated data was normalized to match the experimental data at 80% MVC, where all motor units are active.
For each subject, the RMS was plotted at each contraction level, and linear trend lines fitted. For clarity, the plot for a single subject (subject 1) is shown in Figure 4. The gradient of the sEMG RMS/% MVC trend lines for each of the ten subjects are shown in Table 2(a). Table 2(b) shows the accuracy of the linear fits.

Figure 4. RMS of sEMG against % MVC for a single subject.

<table>
<thead>
<tr>
<th></th>
<th>Experimental</th>
<th>Simulated</th>
</tr>
</thead>
<tbody>
<tr>
<td>Subject 1</td>
<td>2.3520</td>
<td>2.6210</td>
</tr>
<tr>
<td>Subject 2</td>
<td>2.1063</td>
<td>1.7660</td>
</tr>
<tr>
<td>Subject 3</td>
<td>2.0894</td>
<td>1.5992</td>
</tr>
<tr>
<td>Subject 4</td>
<td>2.2177</td>
<td>2.1698</td>
</tr>
<tr>
<td>Subject 5</td>
<td>2.0173</td>
<td>2.4225</td>
</tr>
<tr>
<td>Subject 6</td>
<td>2.1411</td>
<td>1.3087</td>
</tr>
<tr>
<td>Subject 7</td>
<td>2.0157</td>
<td>1.7111</td>
</tr>
<tr>
<td>Subject 8</td>
<td>1.8024</td>
<td>1.2667</td>
</tr>
<tr>
<td>Subject 9</td>
<td>2.2344</td>
<td>2.1117</td>
</tr>
<tr>
<td>Subject 10</td>
<td>1.8387</td>
<td>1.9923</td>
</tr>
</tbody>
</table>

Table 2. (a) Rate of change of RMS with force (MVC). (b) Accuracy of linear fit for rate of change of RMS ($R^2$ value).

<table>
<thead>
<tr>
<th></th>
<th>Experimental</th>
<th>Simulated</th>
</tr>
</thead>
<tbody>
<tr>
<td>Subject 1</td>
<td>0.9985</td>
<td>0.9966</td>
</tr>
<tr>
<td>Subject 2</td>
<td>0.9999</td>
<td>0.9574</td>
</tr>
<tr>
<td>Subject 3</td>
<td>0.9994</td>
<td>0.8516</td>
</tr>
<tr>
<td>Subject 4</td>
<td>0.9987</td>
<td>0.9999</td>
</tr>
<tr>
<td>Subject 5</td>
<td>0.9954</td>
<td>0.9719</td>
</tr>
<tr>
<td>Subject 6</td>
<td>0.9998</td>
<td>0.9993</td>
</tr>
<tr>
<td>Subject 7</td>
<td>0.9801</td>
<td>0.9997</td>
</tr>
<tr>
<td>Subject 8</td>
<td>0.9789</td>
<td>0.9999</td>
</tr>
<tr>
<td>Subject 9</td>
<td>0.9999</td>
<td>0.9921</td>
</tr>
<tr>
<td>Subject 10</td>
<td>0.9992</td>
<td>0.9233</td>
</tr>
</tbody>
</table>

Figure 5 shows the average RMS of sEMG across all 10 subjects for each contraction level. The error bars show the standard deviation of the data from this mean value. The RMS of the experimental and simulated force outputs were also compared. The averages for all ten participants are plotted in Figure 6.

4. Discussion

A model for simulating sEMG was implemented. One novelty of this model was the consideration of physiological parameters such as MU size, conduction velocity and firing rate to have a statistical randomness. Another novelty of this implementation was the non-linear recruitment strategy based on Gaussian distributions of the experimental results reported by Gydikov and Kosarov [10], where at low contraction levels, only slow MU types are active, with firing frequencies in the range of 13-15 Hz. At approximately 40% MVC, fast (or type II) fibers begin to be recruited. At around 60% MVC, all slow fibers are active, while at 80% MVC, all MUs of both fiber types are active. This is in line with experimental results reported by Kukulka and Clamann [17]. Fast MUs, recruited between 40-80%, have firing frequencies that cover a large range of values between 13-24 Hz. The implementation of these firing frequency and recruitment patterns in combination with the described distributed variables, means that this model is closer to accurately modelling physiological parameters. To the knowledge of the authors, this is the first study where such a model has been implemented.

Muscle contraction provides two sets of measurements; sEMG and force of contraction. To validate this model, the model was simulated to generate these two output parameters, and features of these were compared with experimental results. For this purpose, sEMG signals were recorded from the biceps brachii of ten healthy participants.

The relationship of RMS of sEMG with % MVC of the simulated and experimental results was shown to be approximately linear (Figure 5). From Tables 2(a) and 2(b), it
is observed that with the exception of the simulated signals for subject 1, the linear trendlines fit the data well (Table 2(b)). The observed rates of change of RMS of sEMG with % MVC (Table 2(a)) show that with the exception of subjects 6 and 8, the variation in gradient values is similar between the experimental and simulated results. If the sEMG model is to be considered accurate, then the relationship of the RMS of the simulated sEMG with % MVC should be similar to the experimental result. From Figure 4, it is observed that this is correct and that the gradients are quite similar. When considering the average of all participants together (Figure 5), the results indicate a close relationship between the simulated and experimental data. A linear relationship is observed in both cases.

The other measurable output of the muscle contraction is the force output. From Figure 6, it is observed that both the simulated and experimental force measurements are increasing at similar rates. As the force output of the model is calculated by the summation of twitch forces from motor units with varying amplitude and time characteristics, the agreement of the simulated force with experimental data serves to validate the model using an alternative method to the comparison between the electrical sEMG signals.

This is a more accurate representation of the real world as it has inbuilt randomness with simulation parameters following a statistical distribution. When simulated with these distributed variables and non-linear recruitment patterns, the observed linearity of RMS of sEMG with % MVC and of simulated force, match the patterns observed experimentally in this work. The model described here is highly adaptable, and could be used to simulate fatiguing, injured or diseased muscles in the future [22,23].

5. Conclusion

An sEMG model has been presented that demonstrates accuracy when compared with experimental results. The model replicates real-world simulation parameters to generate both sEMG and force signals from the biceps brachii performing isometric voluntary muscle contractions at varying contraction levels. The model is realistic and naturally incorporates features such as jitter and non-linear recruitment patterns. The model has been experimentally validated by conducting experiments on ten volunteers. These experiments have shown a strong correlation between the linear sEMG and force relationships of simulated and experimental data.

References