A model for generating Surface EMG signal of *m. Tibialis Anterior*

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***Abstract*—A model that simulates surface electromyogram (sEMG) signal of *m.Tibialis Anterior* has been developed and tested. This has a firing rate equation that is based on experimental findings. It also has a recruitment threshold that is based on observed statistical distribution. Importantly, it has considered both, slow and fast type which has been distinguished based on their conduction velocity. This model has assumed that the deeper unipennate half of the muscle does not contribute significantly to the potential induced on the surface of the muscle and has approximated the muscle to have parallel structure. The model was validated by comparing the simulated and the experimental sEMG signal recordings. Experiments were conducted on eight subjects who performed isometric dorsiflexion at 10, 20, 30, 50, 75, and 100% maximal voluntary contraction. Normalized root mean square and median frequency of the experimental and simulated EMG signal were computed and the slopes of the linearity with the force were statistically analyzed. The gradients were found to be similar (p>0.05) for both experimental and simulated sEMG signal, validating the proposed model.**

I. Introduction

When a muscle contracts, the electrical activity that is associated with the contraction induces a potential and this can be recorded at the surface using surface electrodes and is the surface electromyogram (sEMG). It is the interferential summation of the motor unit action potentials of all active motor units. SEMG has been used extensively to determine the muscle properties and to estimate strength of muscle contraction. However, due to the complex and gross nature of the signal, the applications are limited, and efforts have been made to model sEMG signal to improve the understanding of the signal. However, sEMG signal models have required the shape of the muscle to be simple and thus the models have only been developed for few muscles such as the biceps brachii because it can be modelled as a homogeneous, linear space invariant [1, 2]. In such fusiform muscles, the resulting volume conductor computation is obtained by the convolution of the transmembrane current with the impulse response of the system to compute the sEMG [3].

Tibialis Anterior (TA) is structurally a bipennate muscle, consisting of a superficial and deeper unipennate halves. Due to the change in direction of the muscle fibres between the two halves, the Tibialis Anterior cannot be modelled as a homogeneous and linear space invariant model [4]. However, modeling TA as a unipennate muscle as a first approximation will allow the linear space invariant volume conductor equation to be used. This will not be an accurate representation; however this will provide a simpler option. This paper has tested if this approximation simulation is similar to the experimental observations.

It has been shown that in a pennate muscle, the electrode samples surface potentials exclusively from closest sources and contribution by deeper sources are negligible [5]. Using this finding, we modelled the TA on the assumption that the deeper unipennate half would not contribute significantly to the EMG. In this study we have investigated the suitability of modelling the TA as a unipennate muscle. The model was populated using parameters that have been reported in literature.

II. Materials and Methods

A. SEMG Model Theory

A motor unit is comprised of an alpha motor neuron and all the muscle fibres it innervates. When a motor neuron is recruited, it sends action potentials to the neuromuscular junction that eventually traverses to ends of the muscle fibres. Collectively, these action potentials that originate from all muscle fibres that are innervated by one motor neuron, or belonging to one motor unit, is known as the motor unit action potential (MUAP).The MUAP from all the active motor units in the proximity of the electrodes undergo attenuation due to the subcutaneous tissue, and the summation of all of these result in the total potential generated by the contracting muscle. This potential induced at the surface electrodes is the recorded sEMG.

The proposed model for the sEMG of the Tibialis Anterior is derived from experimentally determined parameters in literature (refer to Table I). The model uses the firing rate equation developed by De Luca & Hostage [6] with a firing rate variability following an exponential decrease with contraction intensity after recruitment. A custom designed recruitment threshold distribution has been developed based on Klass et al.[7]. This distribution allows the flexibility to manipulate the distribution. The difference between fast and slow motor units was made on the basis of conduction velocity, and fibre diameter.

SEMG model that has been developed and reported in this study was based on model developed by Wheeler et. al [1]. However, the volume conductor which represents the attenuation of the action potential due to the surrounding tissue was modified to take into consideration the inclination of the fibres (m.Tibialis Anterior) with respect to the detection system, as seen in Figure 1.

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$$f(t)=\frac{1}{4\pi\sigma_e}\frac{1}{\sqrt{(Z'_{NMJ}-Z'\pm Z\cdot\cos\theta)^2}} \qquad (1)$$
$$+\sigma\left[(X'-X)^2+(Y'_{NMJ}+Y\pm Z\cdot\sin\theta)^2\right]$$

where $(0, Y'_{NMJ}, Z'_{NMJ})$ is the location of the neuromuscular junction, (X',Y',Z') is the coordinate axis of the electrode, and (X, Y, Z) is the coordinate axis for the muscle fibres. The σ is the square of the ratio between the muscle fibre conductivity σ_i and the external medium conductivity σ_e.

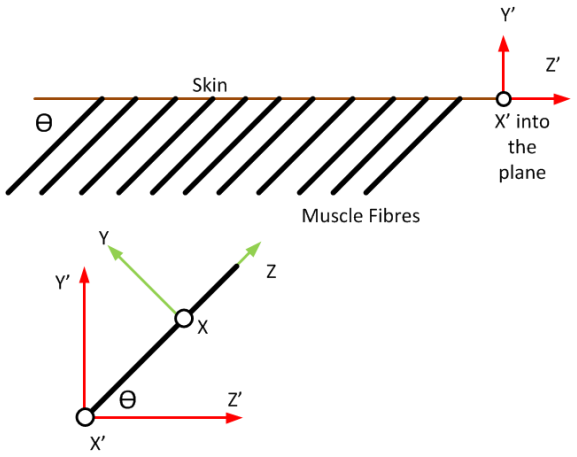

Figure 1. The coordinate axes of the electrode (X', Y' ,Z') and the inclined muscle fibres (X, Y ,Z).

B. Model Implementation

SEMG model was populated using the values of the parameters reported in experimental studies in literature. The parameters for which no values were reported, suitable assumptions were made. These values have been listed in Table I.

The location of the innervation zone was assumed to run the length of the muscle. The innervation ratio was determined by dividing the total number of muscle fibres in the TA by the number of simulated motor units [8]. A randomised spatial location of the two different types of motor units was used.

The firing rate equation implemented in this model has been adapted from the works of De Luca & Hostage [6] for the Tibialis Anterior. The firing rate equation is a function of the maximum voluntary contraction and the recruitment threshold of the motor unit [6]:

$$\lambda(\varphi,\tau)=D\cdot\varphi+(C-A\cdot e^{\frac{-\varphi}{B}})\cdot\tau+E \qquad (2)$$

where the constants are A= 65, B= 0.3, C=-15, D=.54, and E= 20.2, and ϕ is the excitation level (MVC) and τ is the recruitment threshold.

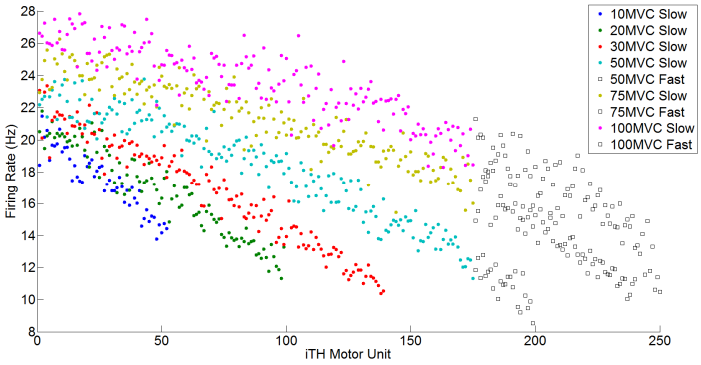

Figure 2. Firing rate spectrum with variance for the modelled 250 motor units at 10, 20, 30, 50, 75, and 100% MVC. The black squares represent the fast type motor units.

TABLE I. MODEL PARAMETERS USED FOR SIMULATING EMG SIGNAL FROM THE TIBIALIS ANTERIOR.

Parameter	Tibialis Anterior Simulation Value
Number of motor units (MU) [9]	250
Average conduction velocity m/s [10]	4.3+/- 0.29
Conduction velocity (fast fibres) [11]	4.9 +/- 0.3
Conduction velocity (slow fibres)	3.9+/- 0.3
Percentage of type 1 fibres (%) [12]	70
Muscle fibre diameter Type 1 (μm) [13]	35.46
Muscle fibre diameter Type 2 (μm) [13]	50.68
Motor unit innervation zone location z-axis (mm)[14]	29.8, uniform distribution
Cutaneous tissue (mm) [15]	Single, 3mm isotropic layer
Duration of AP along fibre (mm) [16]	16
Muscle fibre length (mm) [17]	45
Pennation angle (degrees) [17]	20
Innervation ratio [8]	Poisson distribution with λ = 524 muscle fibres.
Simulation sampling frequency (Hz)	10000

Figure 2 shows the firing rate spectrum for the 250 motor units at 10, 30, 50, 70, 90 and 100% MVC. It is noted that the earlier recruited motor units always have a firing rate higher than later recruited motor units. However, there is a linear increase in firing rate of a recruited unit with increasing MVC. The firing rate ranges from 8 – 30Hz.

The recruitment threshold distribution for the Tibialis Anterior motor units was developed using the values reported by Klass et. al. [7]. The recruitment range was between 1% - 90.2% MVC, with a median of 26.3% MVC, skewness of 0.641 and kurtosis -0.491. This custom designed recruitment threshold distribution allows the flexibility to study the effect of different types of distribution on EMG for future work.

Due to lack of literature on the different recruitment ranges of the fast and slow motors in *m. Tibialis Anterior*, the slow motor units were assumed to be recruited first, followed by the fast motor units. Using the current recruitment threshold distribution, this range is 1 -40% MVC for the slow motors and 40-90 MVC for the fast.

C. Experimental and Simulation Protocol

Eight healthy subjects with no prior neuromuscular injury or disease performed the experiments. Experiments were conducted in accordance with the Helsinki Declaration of 1975, as revised in 2004, and approved by the RMIT Human Research Ethics Committee.

SEMG signal was recorded using Delsys myomonitor 4 (DELSYS, Boston) EMG recording system. The electrode was placed on the Tibialis Anterior according to the SENIAM guidelines, 1/3rd the distance between the tip of the fibula and the tip of the medial malleolus [18]. The skin was abraded and cleaned with an alcohol swap before fixing the electrode.

During the experiments, the subjects sat in a study chair with the hip, knee and ankle flexed at 90, 140, and 90 degrees respectively. The foot was secured to a footplate with Velcro straps to ensure no foot movement during dorsiflexion. A force transducer was attached to the footplate (S type force sensor- INTERFACE SM100) to measure the force exerted.

Prior to the experiments, all subjects were guided to elicit maximal voluntary contraction (MVC). During the experiment, the output of the force sensor was displayed on the screen to provide the subjects with feedback. Once the MVC was determined, the subjects were instructed to perform two isometric dorsiflexions at 10, 20,30,50,75, and 100% MVC, for 10s, with a 2 minute break between each trial. SEMG signal was simulated using the proposed model ten times at 10, 20, 30, 50, 75, and 100% MVC.

D. Data Analysis

Root Mean Square (RMS) and Median Frequency (MDF) of the experimental were calculated at each MVC level, where the force was steady. Similarly, the RMS and Median frequency were calculated for the simulated signal and averaged for each MVC level. The experimental and simulated RMS were normalised to their respective value at 100% MVC.

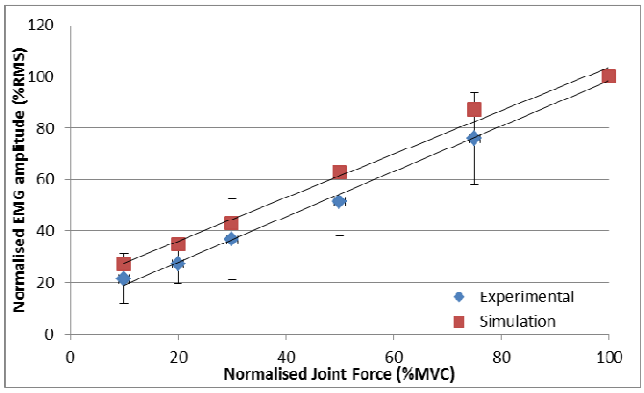

Figure 3. Normalised EMG RMS with respect to the normalised joint force (%MVC) of the experimental and simulated EMG signal.

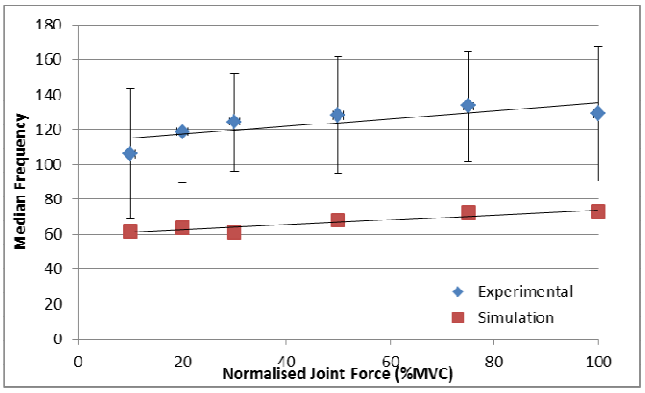

Figure 4. Median frequency with respect to normalised joint force(%MVC) of the experimental and simulated EMG signals.

E. Statistical Analysis

A one factor Analysis of Covariance (ANOCOVA) was performed for comparison of the linear regression slopes of the normalised EMG RMS and median frequency between experimental and simulated signals at a significance level $\alpha = 0.05$. The factor investigated represents the source of the EMG data used, either being experimental or simulated. This statistical test has been used for comparing slopes amongst groups [19].

III. Results

Figure 3 shows the change in normalised EMG RMS of the eight test subjects and for the simulated EMG signal measured as % RMS with respect to normalised joint force measured as % MVC. The rate of change (Srms) in the experimental normalised sEMG RMS is 0.8805 and 0.8468 for the simulated EMG with the force.

Figure 4 shows the change in the median frequency with respect to the normalised joint force measured as % MVC for the eight test subjects and the simulated EMG signal. The gradient of the experimental median frequency (Smf) was found to be 0.2247 and 0.1424 for the simulated EMG.

The coefficient of determination for the experimental linear regression was computed to be 0.048; however, this could be due to the high inter-subject variability.

ANOCOVA statistical results show that there is no significance ($F(1,50)=0.0472; p>0.05$)) between the slopes (Srms) of normalized RMS of the experimental and simulated EMG. Similarly the statistical results for slopes of median frequency (Smf) show that there is no significance ($F(1,50)= 0.0376; p>0.05$)) between the experimental and simulated EMG signal.

IV. Discussion and Conclusion

This paper reports a model for simulating SEMG signal of the Tibialis Anterior muscle during isometric contraction. This model has considered a simplified muscle anatomy and is based on the assumptions that the deeper unipennate half of the Tibialis Anterior muscle does not contribute significantly to the sEMG [5]. This simplified model allows the use of volume conduction equation,

making it easier for simulation analysis.

Tibialis Anterior muscle is structurally a bipennate muscle, and is therefore inhomogeneous, due to the change in conductivities of the muscle fibres between the two halves [4]. Based on this, the linear space invariant filter equation would not strictly apply. However, modeling the bipennate nature of the muscle is computationally challenging and may be the reason for such studies not having been performed earlier.

Mesin et al [5] found that in a pennate muscle, such as the medial gastrocnemius, the electrode only samples action potentials from sources closer to the electrode, with deeper sources providing negligible contribution to the surface potential [5]. This study has used this characteristic of a pennate muscle, and modelled the *Tibialis Anterior* as unipennate, because the deeper half would not contribute as much to sEMG content compared with the superficial half.

The statistical results show the gradients of the normalised EMG RMS and the median frequency between the experimental and simulated EMG signals are not significantly different. This demonstrates that a unipennate model can be used to approximately model the Tibialis Anterior muscle. This eliminates the need for a computationally intensive volume conductor equation to represent the Tibialis Anterior.

The uniqueness of this model is that it incorporates a novel firing rate equation with a customised recruitment threshold distribution. It also has the strength because the parameters are based on values that have been reported in literature by experimental studies. The results support the assumptions of the model and we report that this model can be used for investigating the neuromuscular properties of the Tibialis Anterior. We also propose that this model can be considered to study the effect of neuromuscular changes due to ageing.

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