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Modeling of Single Muscle Fiber Action Potential With Varying Depth

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Modeling of Single Muscle Fiber Action Potential with Varying Depth

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Abstract

Electrophysiological Imaging has improved the clinical uses of Electroencephalography (EEG), Electrocardiography (EKG), and Electromyography (EMG). Despite this, electromyography remains too difficult to implement due to the difficulty of interpreting muscle pathological states. The most accurate forms of electromyographic imaging currently available require intrusive measurement instruments or generate inconsistent results. We aim to explore the utilization of ultrasound in performing accurate electromyography through nonintrusive methods. To do this, we are utilizing an action potential simulation of a single muscle fiber under ultrasonic oscillation to emulate the

Key Words: EMG, Ultrasound, Single Muscle Fiber Action Potentials, Volume Conductor

I. Introduction

Electromyographic (EMG) signals represent the electrical activity of a muscle during contraction. A skeletal muscle consists of many fibers each fiber is innervated through a synaptic nerve termination of a motor neuron located in the anterior horns of the spinal cord. Fibers innervated by the same motor neuron constitute a motor unit. The motor unit potential is the summation of individual fiber action potentials with the uptake area if the recording electrode. Thus, electromyography (EMG) provides convenient means to study muscular functions during biofeedback training, activities of sports or daily living[2,9,19,21]. It is also useful in interpreting pathologic states of musculoskeletal or neuromuscular systems[23,25,27]. In particular, EMG offers valuable information concerning the timing of muscular activity and its relative intensity[34].

Traditionally, surface EMG signals are recorded over muscle sites that are usually determined by surface landmarks and anatomic guides. Yet, no two EMG recordings produce the same data quantifications[14]. Changes in the positions of the electrodes can cause significant difference in the derived electrical muscle activity and can produce substantial alterations in the relationship between mechanical and electrical activity[30]. The electrode position has a definite effect on the EMG-tension relationship, primarily because of its marked effect on the amplitude and phase of the EMG signal at the same tension[14,32]. Both the timing and relative effort intensity may be altered. Because myoelectric signals propagate through muscles and layers of tissue, EMG signals detected by the recording electrodes may be erroneously interpreted as generated by muscle fibers within the sampling field of the electrode[4,23]. The volume conduction often blurs the onset and cessation of muscular activity and alters the relative effort intensity leading to erroneous interpretations[3,4,24].

While other biomedical signals such as EEG and EKG use multi-channels for topography, EMG recordings have been limited only to few channels[7,21,26]. There are three types of electrodes that currently are used to detect EMG signals. Needle electrodes are used mainly for motor unit EMG and nerve conduction studies. Intramuscular fine wire electrodes have been considered for their field selectivity and their reduced range of pick up[23]. Surface electrodes are noninvasive small metallic discs, which can be tapped to the skin. Surface electrodes have wider range of pick up and different shape and surface area have been reported in literature[20,28,29]. Comfort and ease of use are the major advantages of surface electrodes in recording EMG during function and motion analysis. The spectral content of signals detected by intramuscular fine wire electrodes allows filtering of some of the lower frequency volume conducted signals and thus reduces its[10,27]. Surface electrodes, on the other hand, are most affected by the volume conductor[30]. Because of its noninvasive nature, surface electromyography is the object of continuous development. However, surface electromyography continues to suffer from its poor specificity. Similarly, but for different reasons, optical tomography has a poor image resolution when imaging thick tissues over several millimeters[11,12,15]. Ultrasound imaging techniques have been extensively explored in many engineering applications[16,31,33]. In this paper, we propose a novel technique, which combines ultrasound techniques with surface electromyography. Ultrasound modulated surface electromyography is a hybrid modality, which provides localization of muscle fibers and thus enhance electromyographic diagnostics. In ultrasound modulated surface electromyography, the potential distribution at the surface of a muscle is modulated by ultrasound waves propagating through the subcutaneous tissues and volume conductor.

The objective of this study is to investigate the theoretical mechanisms involved in acousto-electric interactions in a muscle model and investigate the potential application of ultrasound modulation in electromyography. We propose to design and implement novel processing techniques, which are capable of reducing the effects of volume conductor and limited sampling of electromyographic signals (EMG), using ultrasound modulation and non-invasive surface electrodes. The designed concepts will be demonstrated using simulated single muscle fiber action potentials and a continuous ultrasound wave. A muscle model will be used to study the electrophysiological field distribution and the effects of the volume conductor on surface single muscle fiber action potentials.

The proposed method would, if functional, provide an unintrusive method of sampling electromyographic signals (EMG) that provides more a higher resolution with greater control of the region detected. There are no additional risks when compared to standard surface electromyography, and the equipment required only includes surface electrodes and an ultrasound generator. The duration of development primarily requires methods of modulating electrical signals. While

II. Single Muscle Fiber Action Potentials Modeling

The time-varying fields associated with electromyographic currents are considered low frequency fields, and thus, the electrical field distribution can be approximated by a static field satisfying the

time-varying boundary conditions[8]. For a region with homogeneous conductivity σ the relationship between field potential, Φ, and the volume source density, I_v, can be stated using Poisson's equation as:

$$\nabla^2 \Phi = -\frac{I_V}{\sigma} \tag{1}$$

In general, the potential field at a particular point due to all current sources, in the volume conductor, can be expressed using Green's function as follows[1,6]:

$$\Phi(P) = \int_V G(P,P')I_v\,(P')dV(P') \tag{2}$$

where Green's function, *G(P,P')* produces the potential at point *P* due to a point charge at point *P'*. Due to the anisotropy of the extracellular medium, the field potential is expressed using modified Laplace's equation[24]. The extracellular action potential from a single muscle fiber is generated by the transmembrane current, flowing inward and outward during depolarization and repolarization[5,17]. The extracellular potential propagates from the neuromuscular endplate in both directions along the muscle fiber with a conduction velocity, v_c. If the muscle f
iber is parallel to the *z* axis and the neuromuscular endplate is located at *z = 0*, then the intracellular potential V_i at a particular point *z* and time *t* is expressed as follows:

$$V_i(z,t) = V_i(t - |z/v_c|) \quad and \quad V_i(z) = \alpha(\lambda z)^3 e^{-\lambda z} - \beta \tag{3}$$

where $V_i(z)$ is the profile of the intracellular action potential along the length of the muscle fiber in the *z* direction, α and β are constant values adapted from clinical data, and λ is a scaling factor[18,22]. At the neuromuscular endplate and muscle fiber endings, the transmembrane current is constrained by the excitation and extinction principles. Far from the neuromuscular endplate and fiber endings, the transmembrane current density is proportional to the second derivative of the intracellular potential:

$$i_m(z,t) = \frac{\sigma_i \pi d^2}{4} \frac{\partial^2 V_i(z,t)}{\partial z^2}$$

where σ_i is the intracellular conductivity and *d* is the muscle fiber diameter. In the particular case where the extra-cellular medium is considered infinite with cylindrical anisotropy, the potential recorded at the surface of the model when a current source travels along the muscle fiber from the neuromuscular endplate toward the tendons can be approximated by a potential function of a current source[22]. For an observation point (*x, y, z*) and a current line source located at (x_o, y_o, $z_o(t)$) in a medium with cylindrical anisotropy, the single fiber potential recorded at the surface of the model can be written as:

$$\Phi(x,y,z,t) = \frac{1}{4\pi\sigma_r} \int_{z_1}^{z_2} \frac{i(z',t)dz'}{\sqrt{\frac{\sigma_z}{\sigma_r}((x-x_0)^2 + (y-y_0)^2) + (z-z')^2}}$$

Where σ_r is the radial conductivity, σ_z is the longitudinal conductivity, and z_1 and z_2 are the coordinates of the ends of the muscle fiber. The coordinates (x_o, y_o) are the positions of a single muscle fiber. The single fiber potential can be presented by a convolution of the transmembrane current and the impulse response of the volume conductor as follows:

$$\Phi(x,y,z,t) = i_m(z,t) * h(x,y,z)$$

To account for the finite dimensions of muscle fibers and volume conductor distortions, the excitation and the extinction of the action potential are described as current sources at the neuromuscular endplate and fiber endings[13,22,24]. These compensating current sources are calculated such as the total current is zero over the active part of the fiber ($\int_{fiber} i_m(z,t)dz = 0$). The simplifier muscle model consists of a straight single muscle fiber located at depth *y* within the extracellular medium considered as infinite and with cylindrical anisotropy. The muscle fiber is assumed cylindrical with a length *L* and a diameter *d*. The neuromuscular endplate is located at the center of the fiber.

III. Simulation Design

For simulation, the diameter of the single muscle fiber is selected to be 60 μm. The muscle conduction velocity is set to 4m/sec. A surface electrode 10 mm away from the muscle endplate was simulated to record simulated single muscle fiber action potentials at a sampling frequency of 25 kHz. Many of the units were additionally scaled as to be easier to read and visualize. These include using time scales in milliseconds and making measurements in millivolts. Several options for the position of the muscle fiber have been implemented, including constant depth, linear variation in depth, and sublinear depth scaling. Additional models can be implemented, though these provide simple fiber arrangements that allow for the testing of a variety of simulation elements. The homogeneity of the muscle tissue, lack of anisotropy, and lack of consideration for temperature allow for a simplification in both programming the simulation and a reduction in time required for running the simulation. The model additionally does not note the effects of nearby blood vessels or bone on the readings. These limit the depth and accuracy of the simulation. The single fiber nature similarly reduces the complexity, though the analysis of this effect on a small scale is the intention. The presence of other fibers would introduce more complexity to the generated information.

Of the previously noted constants used, the α is 96, the β is -90mV, and the scaling factor is one. The intracellular conductivity is 1.01, the radial conductivity is 0.063, and the longitudinal conductivity is 0.330. The convolution method of modeling the system was used rather than the integration method. This resulted in a lack of necessity to integrate between two positions, allowing for the action potential to be determined at individual points without the remainder of the

path influencing the value. Of these components, modifications to each only alter the scale of the simulation values. The shape of each plot remains the same.

IV. Results & Discussion

Due to the primary focus on the alterations made by the presence of ultrasound, the experiment requires data from both a stationary fiber and an oscillating fiber. The stationary fiber acts as the control while the oscillating fiber has the ultrasound applied. By isolating the frequency spectrum of each, the broad impact can be more apparent regardless of the scale of the graphs, while also providing information on the broad behaviors of the signal.

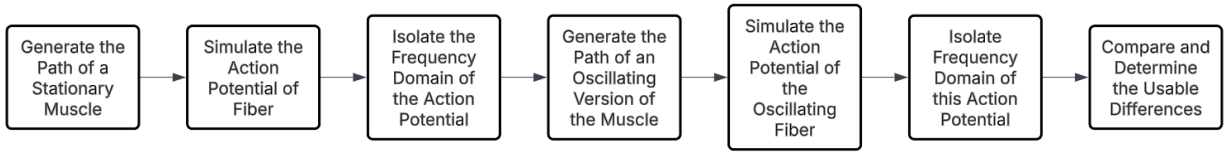

Figure 1: Flowchart of the steps taken in this experiment.

When implementing the simulation, the CMAP follows a pattern across many paths. This consists of a region of high activity near the probe prior to a decaying in magnitude at greater distances.

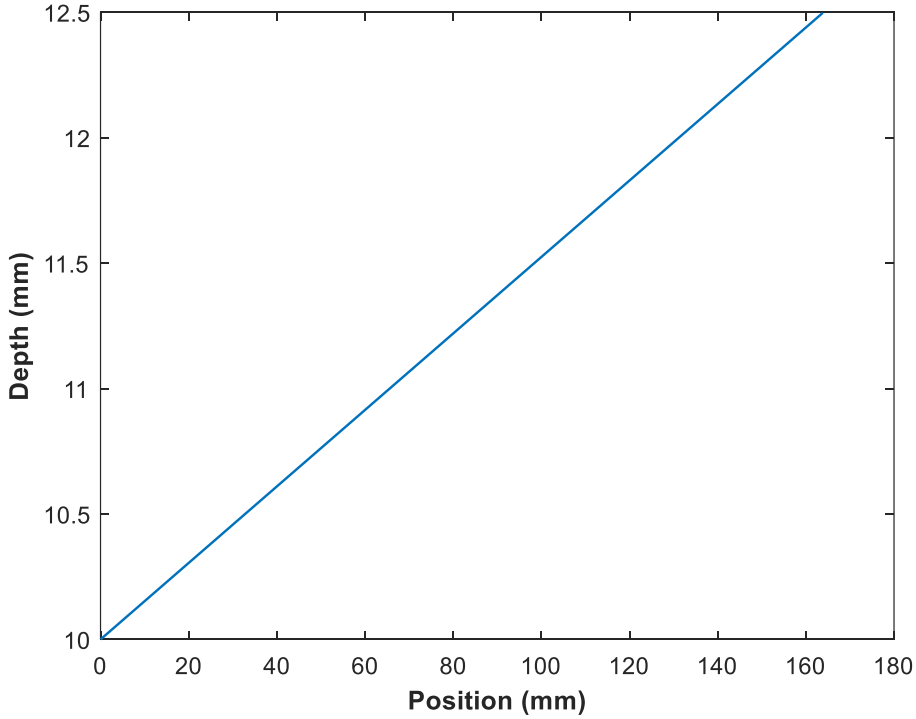

Figure 2: Illustration of the position of a simulated muscle fiber.

The muscle fiber is simulated as a line of variable depth. This is accomplished by creating a model which aligns the position with one of several functions. The above function linearly scales with distance from the endplate, though other paths are also employed. These include paths of constant depth and sublinear growth. Figure 2 indicates the path of a muscle fiber that follows a linear path.

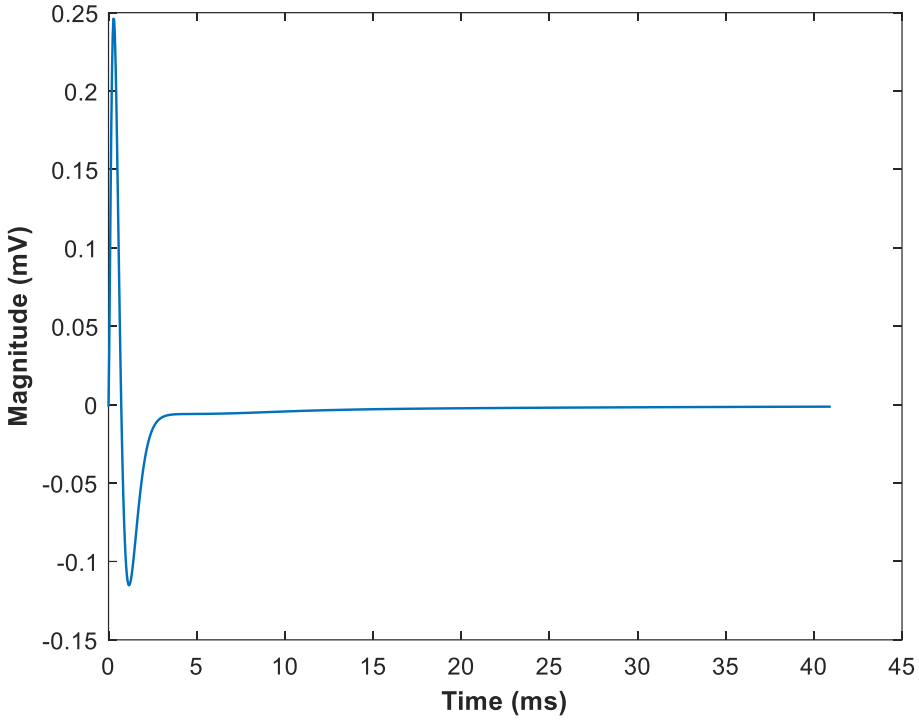

Figure 3: Illustration of the simulated action potential through time.

The action potential of the simulated muscle fiber is determined by the impulse response of the intracellular potential. By observing the behavior of the simulation, it is apparent that the action potential is most easily measured near the electrode's position. As indicated by Figure 3, greater distances correlate with the recorded action potential decreasing to a negligible magnitude. This is one of the primary components that is influenced by the motion of the fiber relative to the electrode. This is due to the impact of depth on the action potential.

The frequency spectrum of the action potential allows for simplified viewing by reducing a set of influences into separate components that can be understood visually. Figure 4 correlates to the muscle fiber following a linear path. Despite the variance of paths a muscle fiber may take, many have similar frequency domains. Figure 4 indicates a constant component is present, as well as a spread across the frequency domain.

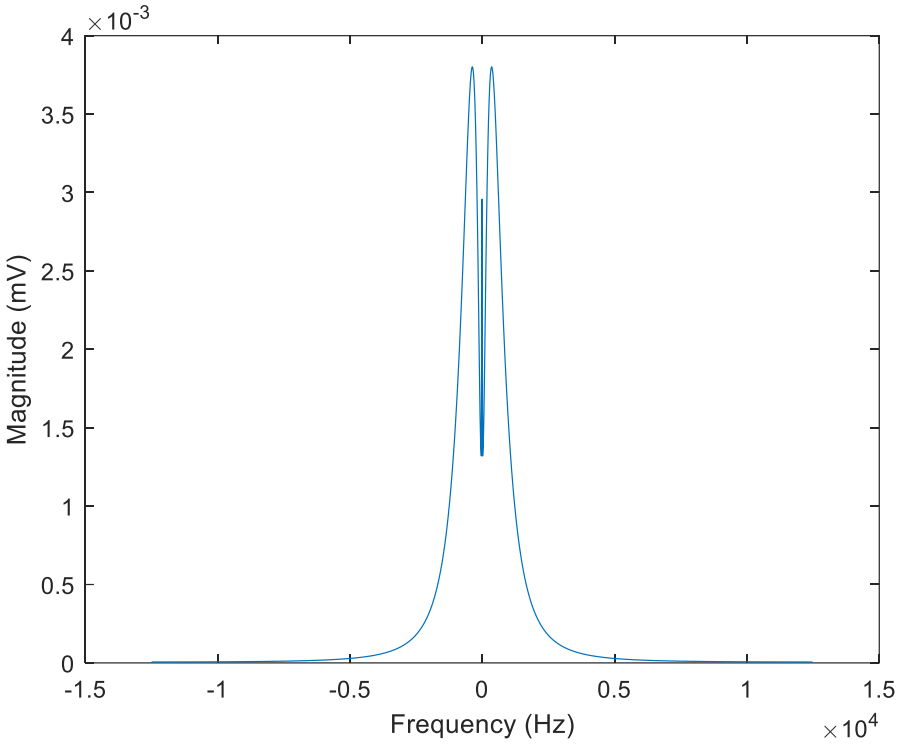

Figure 4: Illustration of the frequency spectrum of the simulated action potential.

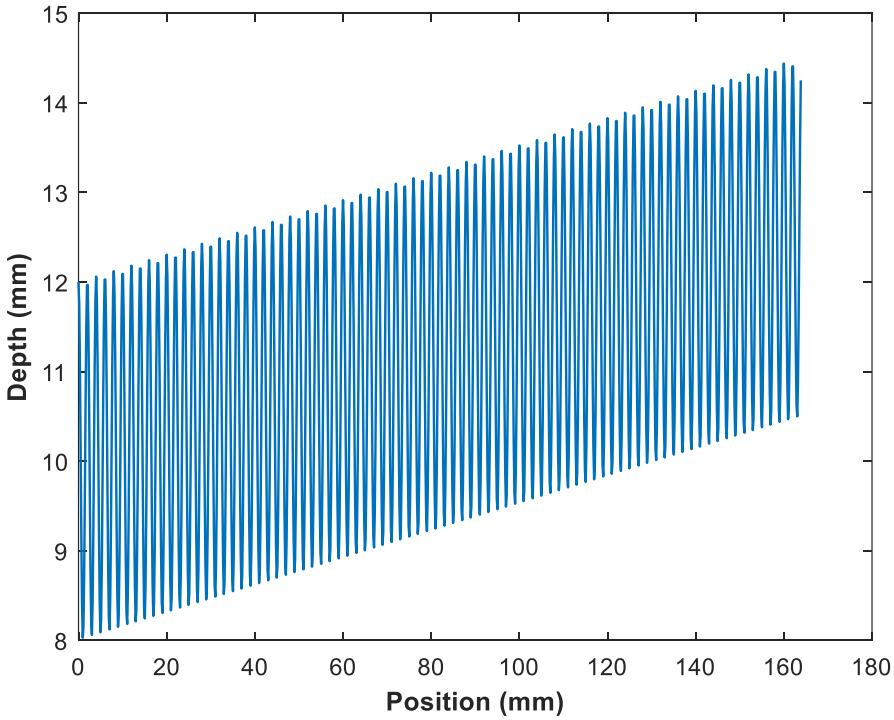

Figure 5: Illustration of the simulated muscle fiber at the position it is in as each point is measured.

The initial steps to determine the effects of ultrasound on the electrical properties of a muscle fiber is to apply the motion induced by the ultrasound generator. This is modeled as an oscillation of 2kHz on the muscle fiber. As the pulse of the action potential is functionally at a given point, the position of the fiber is modified in accordance with the oscillator. Figure 5 indicates the position of a linear muscle fiber as action potential pulses travel across it. The oscillation only affects muscle fibers at a depth that corresponds to the penetration depth of a given ultrasonic wave generator.

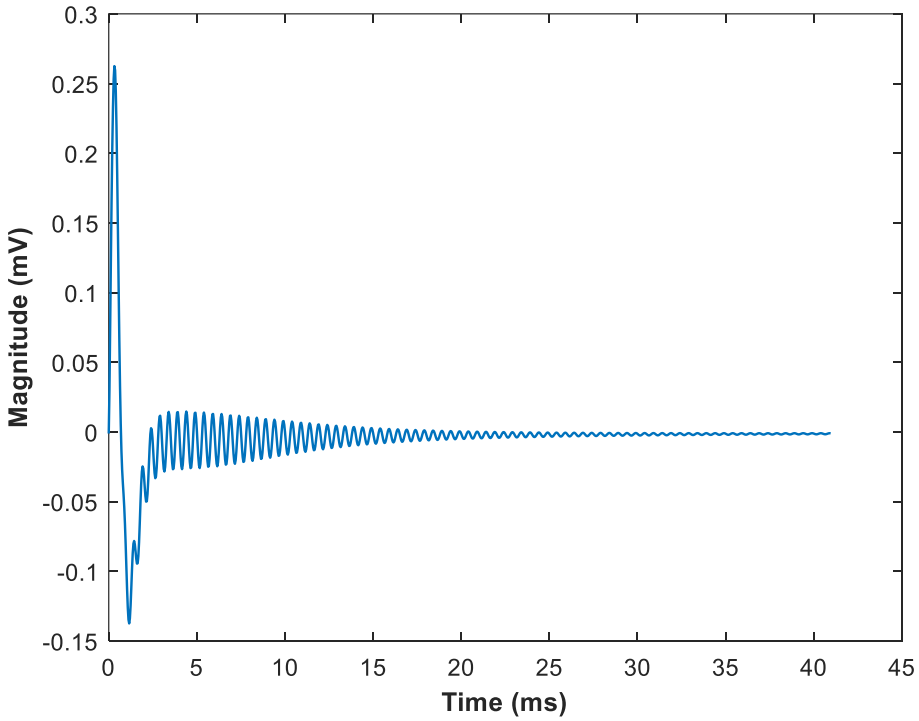

Figure 6: Illustration of the simulated action potential across an oscillating muscle fiber.

Figure 6 reveals several alterations to the action potential when the fiber is oscillating. Similarly to the position of the muscle fiber, the action potential appears to be modified through an additive process by the oscillator. This is not the case, however, due to the reduced impact as longitudinal distance increases. When the action potential wave is near the electrode, motion aligned to the normal of the skin is aligned with the depth of the fiber. This is the dominant component of the distance. As the distance from the endplate increases, the angle between the longitude of the fiber and the vector between the electrode and pulse approaches 0. The oscillation is perpendicular to this vector and is sufficiently small as to be a negligible shift in distance.

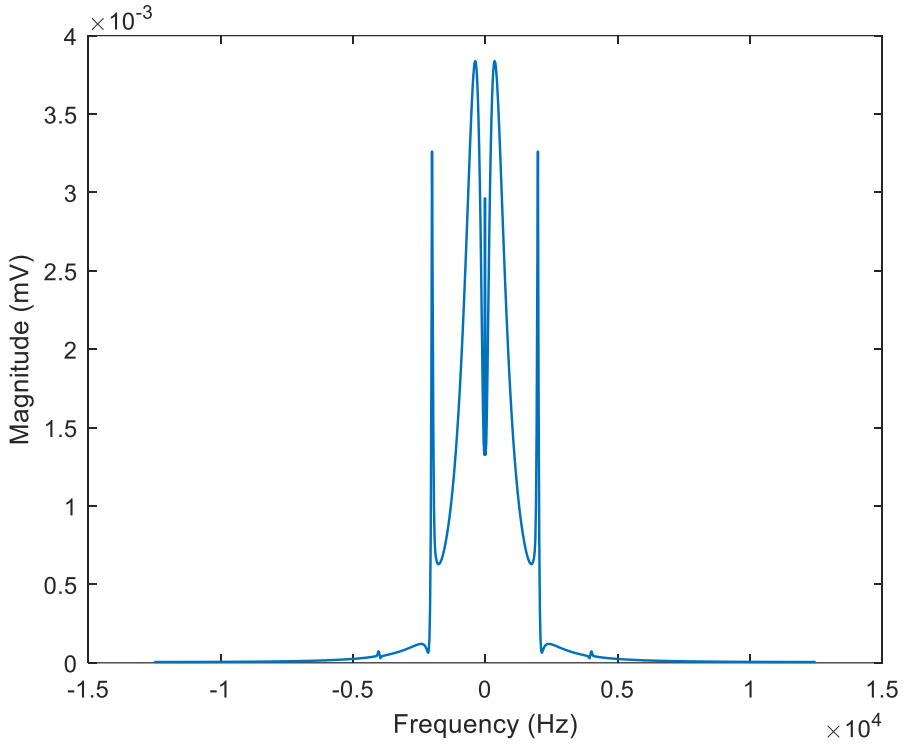

Figure 7: Illustration of the frequency spectrum of the simulated action potential.

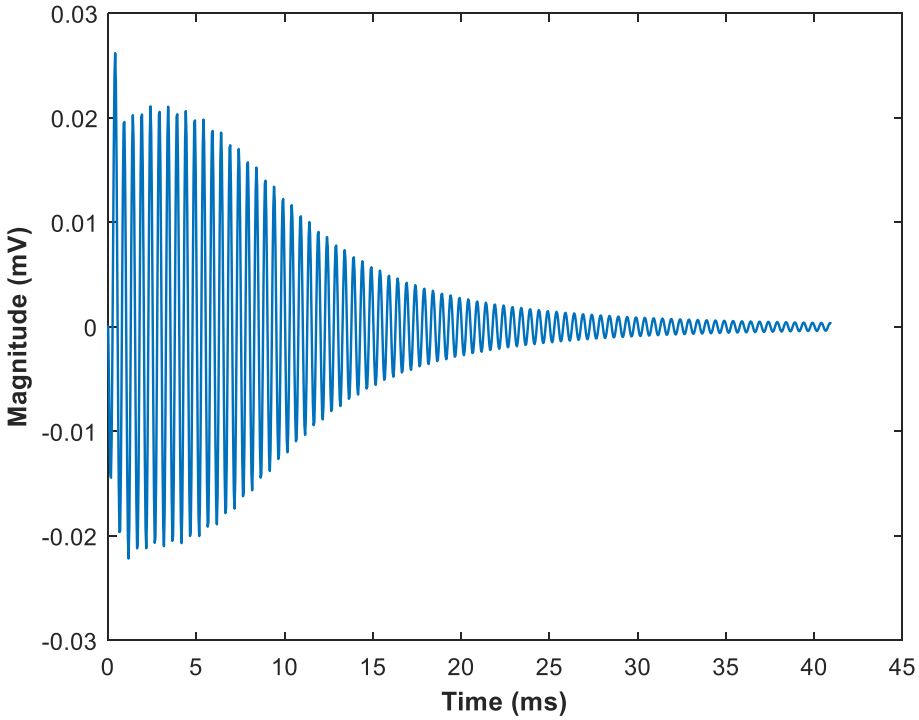

Figure 8: Illustration of the component of the simulated action potential caused by oscillation.

Figure 7 indicates the frequency domain of the action potential as modified by ultrasound vibrations. When compared to the frequency domain of the action potential for a stationary fiber, the primary observable difference is the inclusion of peaks located at the frequency of the oscillation. These are the primary components that allow for EMG readings through ultrasound. These oscillations are only present in muscle fibers at a particular depth.

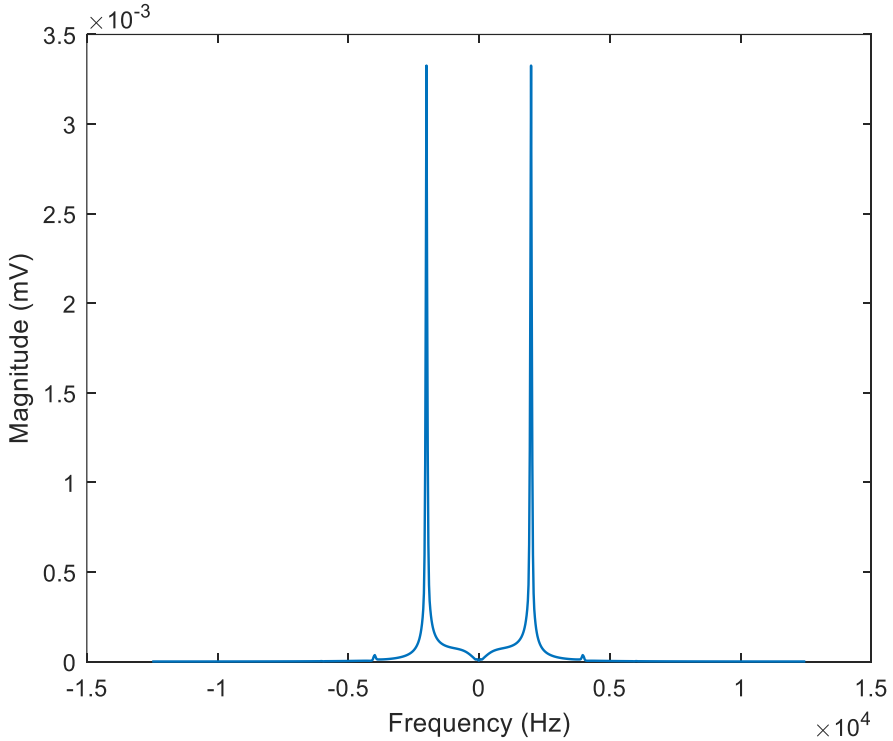

Figure 9: Illustration of the frequency spectrum of the simulated action potential caused by oscillation.

In Figure 8 and Figure 9, the difference between the stationary fiber and the oscillating fiber are determined. In Figure 8, the action potential consists of an oscillating signal with a decaying magnitude. Figure 9 illustrates the frequency spectrum. This reveals a dominant spike at the oscillating frequency, and a set of frequencies that follow due to the decay as the distance from the electrode increases. One of the primary differences between fiber position and the action potential Is the expansion from the oscillation frequency. This is due to the decay that is present in the

V. Conclusion

In this study, we have explored the behavior of single muscle fibers when an ultrasound frequency alters the position of the fiber. We accomplished this through the modeling of a muscle fiber and the electrical properties. This was accomplished by representing the fiber as a volume conductor and simplifying the behavior into a one-dimensional state. Ultrasound vibrations were applied, represented through a shift in the position of the fiber. This presented a clear indication of the effects of the ultrasound and how it influenced the muscle fiber. Through the process of

demodulation practices, it may be possible to detect a particular region of muscle tissue or set of fibers unobtrusively. Due to the limited region affected by ultrasonic vibrations and the effects of motion on the detected action potential, the readings of a fiber can be isolated and observed. Because of this, we believe the information that has been determined in this study will allow for further understanding of the use of ultrasound in studying and diagnosing nerves. It may potentially allow for non-invasive methods of observing nerves to enhance patient experience, prevent risk for injury, and provide methods to accurately observe muscle behavior without the same uncertainty caused by movement in needle electrodes. Additionally, the downsides of surface electromyography may additionally be overcome through our methodology.

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