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Investigating Volume Conduction Effect in MMG and EMG during Action Potential Recording

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***Abstract*— The study and measurement of the magnetic field from the skeletal muscle is called Magnetomyography (MMG). These magnetic fields are produced by the same ion currents which give rise to the electrical signals that are recorded with electromyography (EMG). Layers between the muscle and skin surface, known as volume conduction, play a critical role during the measurement. This paper presents the volume conduction effect on the electrical and magnetic signals with the finite-difference time-domain simulations using Sim4Life. The effects of 1 mm fat on the recorded electrical and magnetic signals from the skin surface have been evaluated in both EMG and MMG. The results indicate that due to 1 mm fat, the electrical signals decrease over 60% through traveling across layers between the muscle and skin surface, while these layers are transparent to the magnetic field. In a similar simulation procedure, when the new fibers are recruited, the interference among electrical signals makes the strength of recorded signals behave non-linearly proportional to the increasing number of active muscle fibers. Sim4Life simulations show that the recorded magnetic signals do not have the same trajectory as electrical signals. Hence, the changes in EMG signals caused by the volume conduction effect can result in signal misinterpretations.**

Keywords—Electrical signal, Electromyography, EMG, MMG, Magnetic signal, Magnetomyography, Volume conduction effect

I. Introduction

Smooth control of muscle movements relies on the complex interaction between the neuronal system and the skeletal muscle, which is called the neuromuscular system. In short, a motor unit is a basic functional unit of the neuromuscular system, which consists of a motor neuron and all the innervated muscle fibers. Contraction of muscle fibers in a motor unit is initiated by a traveling action potential from an innervating motor neuron, and neuromuscular junction to the entire length of the target muscle fibers membrane. Propagating action potential along the muscle fiber length causes both axial and transmembrane electrical currents. This time-varying transmembrane electrical current elicits electrical field changes in extracellular tissue, which can be measured both invasively through needle electromyography and non-invasively from the surface of the skin by employing surface electromyography. While electromyography is a long-standing method in many fields such as human-machine interface, sport, ergonomics, prosthesis, and biofeedback, it suffers from several limitations. Firstly, the recorded signal is a compound motor unit action potential as it is the spatiotemporal summation of all the active muscle fibers within the electrode pickup area. Additionally, these signals are changed by the volume conduction effect, which means that they may be distorted, dispersed, and filtered by traveling

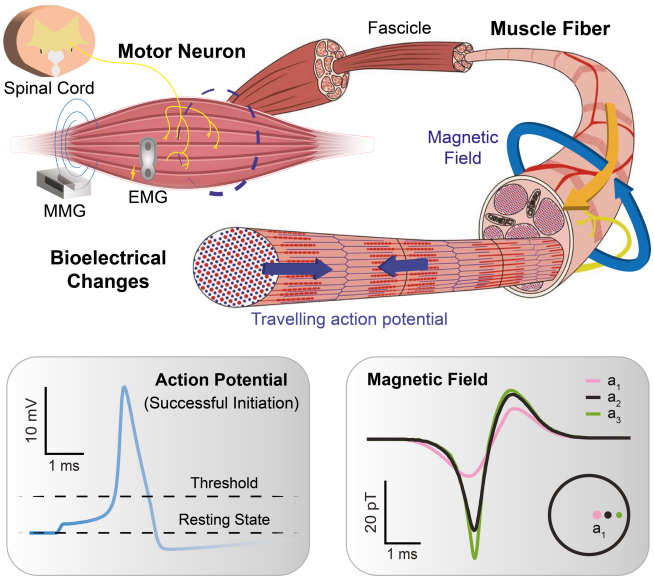


Fig. 1. Overview of EMG and MMG sensing technologies.

through various biological tissues between muscle fibers and the skin surface. So, the recorded signals suffer from low spatial resolution and rarely give information from a signal source [1]. Moreover, the electrical contact, which is essential in both surface and needle EMG in recording electrical fields, can cause challenges in recorded signal characteristics as they can be changed by electrode location, interelectrode distance, and physical electrode size [1].

The above-mentioned limitations are drivers for the development of Magnetomyography (MMG) which is the magnetic measurement of muscle activity and was firstly proposed by Cohen and Gilver in 1972. This magnetic field is generated by the electrical current caused by the traveling action potential in muscle fibre's length, as shown in Fig. 1. The correspondence between electrical signals and their counterpart magnetic signals derives from Maxwell–Ampère law. Although MMG could overcome EMG drawbacks, there are a few challenges that limit its practical use. Most importantly, the magnitude of biomagnetic signals is low, in the range of pico (10^{-12}) to femto (10^{-15}) Tesla (depending on the experimental set-up) [2], which requires high technical demands concerning sensitivity, level of detection, portability, cost, and noise cancellation as magnetic signals can be easily polluted by the environmental magnetic background [3, 4]. Although the volume conduction effect is one of the limitations of EMG, which causes phase cancellation and acts as a low-pass filter in recorded signals, it makes extracellular electrical measurements possible even at distance from the

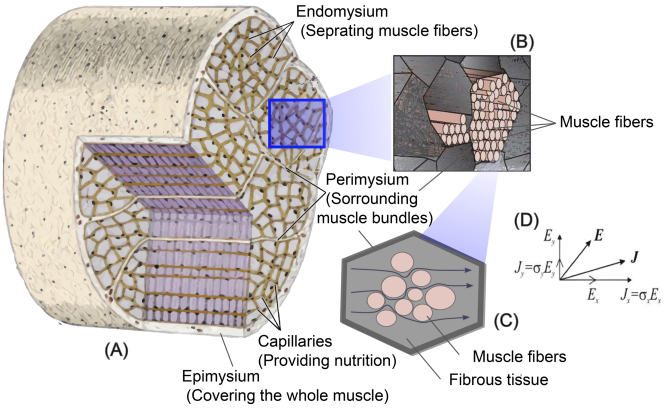


Fig. 2: Anisotropy of electrical signals. A) Skeletal muscle B) Muscle fibers (Longitudinal view) C) Cross-section of muscle fibers and transverse currents which needs to cross the fibrous tissue to reach the surface D) Current density (J) and electrical field direction (E) in an anisotropic medium with different electrical conductivity σ.

signal source by passively conducting current between the regions of potential differences. This mixing effect of volume conduction can be hardly explained quantitively without implementation in models. Since lack of proper knowledge results in misinterpretation of EMG signals, it is the major cause of most model development in the electrical field measurement [5].

Anisotropy is one of the concepts used to explain the volume conduction effect which is the difference between the electrical conductivity in the fiber direction with the one perpendicular to the fiber direction. As illustrated in Fig. 2, since the muscle fibers membrane has a high impedance, a small number of currents can travel through the cell membrane, and they have to go through the surrounding fibrous tissue which causes a high resistance for transverse current. However, in the fiber direction, an intracellular area with relatively small resistance is available for electrical conduction. Consequently, the current density (J) is not the same as the electrical field direction (E) which means that as a result of anisotropy, the same electrical field strength will cause the different current densities in different directions that can be written as:

$$J = \sigma E \tag{1}$$

where σ is the electrical conductivity which is a mathematical matrix rather than a scalar value.

II. Methodology

A. Sim4life Platform

Sim4life (V7.0.1) is a Finite-Difference Time-Domain (FDTD) simulation platform that combines powerful physics solvers with advanced anatomical models of humans and animals. Hence, it is well-suited for testing the efficiency and safety of medical devices and treatments in a realistic physiological environment, compared to COMSOL Multiphysics for device modelling [6]. Moreover, Sim4life facilitates all steps, from modelling, simulating, and analysing to result in visualization with high flexibility. This commercially available FDTD simulation platform permits a direct solution of Maxwell's curl equations to be derived in the time domain [7] and is discretized with automatically grid for the tissue model.

B. Modelling framework

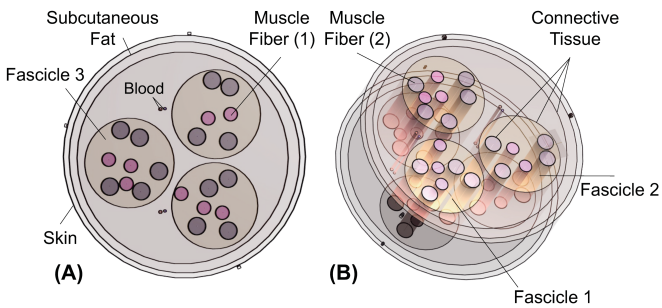
Skeletal muscle was modelled as a cylinder that consists of three fascicles that contain a variable number of muscle fibers in two different colours. Each colour shows that the muscle fiber is innervated with different motor neuron. As depicted in Fig.3B, Blue muscle fibers with a 52 µm diameter and 500 µm length are innervated by larger motor neurons (diameter: 7.3 µm), whereas pink muscle fibers with a 43 µm diameter and 500µm length are innervated by smaller motor neurons (diameter: 5.7 µm). Small motor neurons innervate relatively few and small muscle fibers yielding in generating small muscle force (8 muscle fibers innervated by the small motor neuron), while large motor neurons supply a greater number of muscle fibers with higher diameters, resulting in powerful muscle force (larger motor neuron innervates 13 muscle fibers). Different sizes of motor units follow the size principle [8] which states small motor units are recruited first and progressively larger motor neurons are recruited to generate forceful contraction. Therefore, in this paper, first small muscle fibers are recruited, then according to size order, larger muscle fibers are added to the simulation. Besides, as summarized in Fig. 3, connective tissues like Epimysium, perimysium, and endomysium are modelled as surrounding the skeletal muscle, fascicles, and muscle fibers respectively. Then subcutaneous fat, skin, and blood vessels are added to the model to complete the structure of the muscle.

C. Electrical Field Distribution

For assessing the volume conduction effect on the transverse currents, first, all small muscle fibers are activated, after that, four large muscle fibers from fascicle 1 were added to the simulation. The electrical and magnetic signals over the skin, in both scenarios, are measured to evaluate the electrical and magnetic signals interference among activated muscle fibers. Further, for studying the influence of fat, the electrical and magnetic fields have been simulated in two new conditions: without fat and with 1 mm fat. In each state, large muscle fibers following the size principle were recruited to investigate the influence of fat on the recorded electrical and magnetic signals in a different number of active muscle fibers.

D. Magnetic field Distribution

The magnetic field which is generated around the muscle fibers has been expected to be mainly due to the axial intracellular current rather than the transmembrane currents [9]. As such, for measuring the magnetic field distribution, an intracellular axial current is applied to each muscle fiber. For measuring the axial intracellular current:



- Muscle Fiber (1): Recruited first and innervated by small motor neuron
- Muscle Fiber (2): Recruited progressively and innervated by larger motor neuron

Fig. 3: Sim4life Modelling of the cross-sectional area of skeletal muscle: (A) transvers view and (B) oblique view.

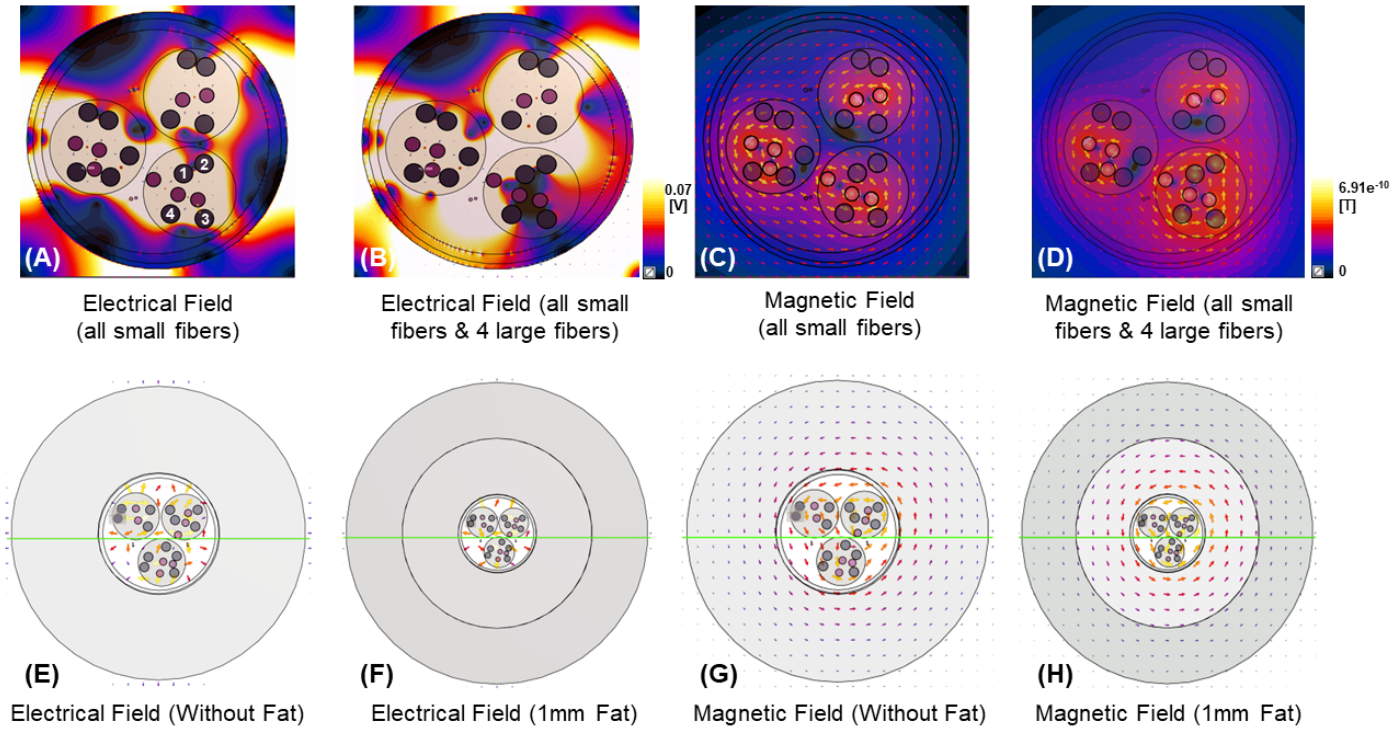


Fig. 4: All small muscle fibers are activated: (A) electrical and (C) magnetic field distribution. Four large muscle fibers from fascicle 1 shown with numbers 1 to 4, were added : (B) Electrical and (D) Magnetic field distribution respectively. Electrical field distribution: (E) without fat and (F) with 1 mm fat and Magnetic field distribution: (G) without fat and (H) with 1 mm fat.

$$I_i = \frac{V_{i+1} - V_i}{R_i} \tag{2}$$

where I_i, V_i and R_i are axial intracellular current, membrane potential, and axial resistance in each compartment of i respectively.

R_i can also be calculated with:

$$R_i = \frac{\rho \cdot l}{A} \tag{3}$$

in which ρ, l, and A are expressed as specific resistivity, length, and cross-section of the compartment.

According to these two equations, the axial intracellular current is proportional to the muscle fiber radius. Since the peak-to-peak amplitude ranges from 50 to 100 nA [10], the intracellular axial currents in muscle fibers were assumed to be 55 and 60 nA for small and large muscle fibers respectively. So, first, all small muscle fibers are activated by applying 55 nA in each one and then following the size principle and applying 60nA, four larger muscle fibers from fascicle 1, are recruited. The magnetic fields are measured on the skin surface over fascicle 1, to assess the magnetic field changes while activating new muscle fibers.

For evaluating the volume conduction effect in the magnetic field, magnetic field distribution was recorded without and with 1 mm fat in two different numbers of muscle fibers following the size principle. Hence, the influence of fat on the measued magnetic signals, while recruiting new muscle fibers, can be evaluated.

III. Results and Discussion

This work is implemented to show the volume conduction effect in the recorded magnetic and electrical field over the skin. Simulations with Sim4life platform have been started by recruiting new muscle fibers with the same diameter of connective tissues, subcutaneous fat, and skin to see the electrical and magnetic field changes on the skin. As illustrated in Fig. 4A and 4C, all small muscle fibers from different fascicles are recruited and the electrical and magnetic fields over the activated fibers have the highest value. However, by activating the large muscle fibers from fascicle 1, the electrical field just over the fascicle 1 (Fig. 4B) is not increased proportional to the number of active muscle fibers which means that the electrical field of the active adjacent muscle fibers neutralizes and interferes with one another.

We repeated the above-mentioned steps for magnetic field distribution. As shown in Fig. 4D, recruiting large muscle fibers from fascicle 1 has not the same effect as an electrical field since the magnetic field is higher on the skin over the fascicle 1. Then, the same steps have been followed in two conditions, without fat, and with 1 mm fat with the same thickness for connective tissues and skin. The arrows in Fig. 4E and 4F, represents the electrical field intensity and direction which has decreased by adding 1mm fat to the simulation. However, as depicted in Fig. 4G and 4H, by placing 1 mm fat, the magnetic field intensity has not changed.

Next, the recorded electrical and magnetic fields along the green line in Fig. 4 have been demonstrated in Fig. 5. We compare the signals over the skin which would be 2 mm (without fat) and 3 mm (1 mm fat) to evaluate the effect of fat on the signals. As shown in Fig. 5A and 5C, when all small

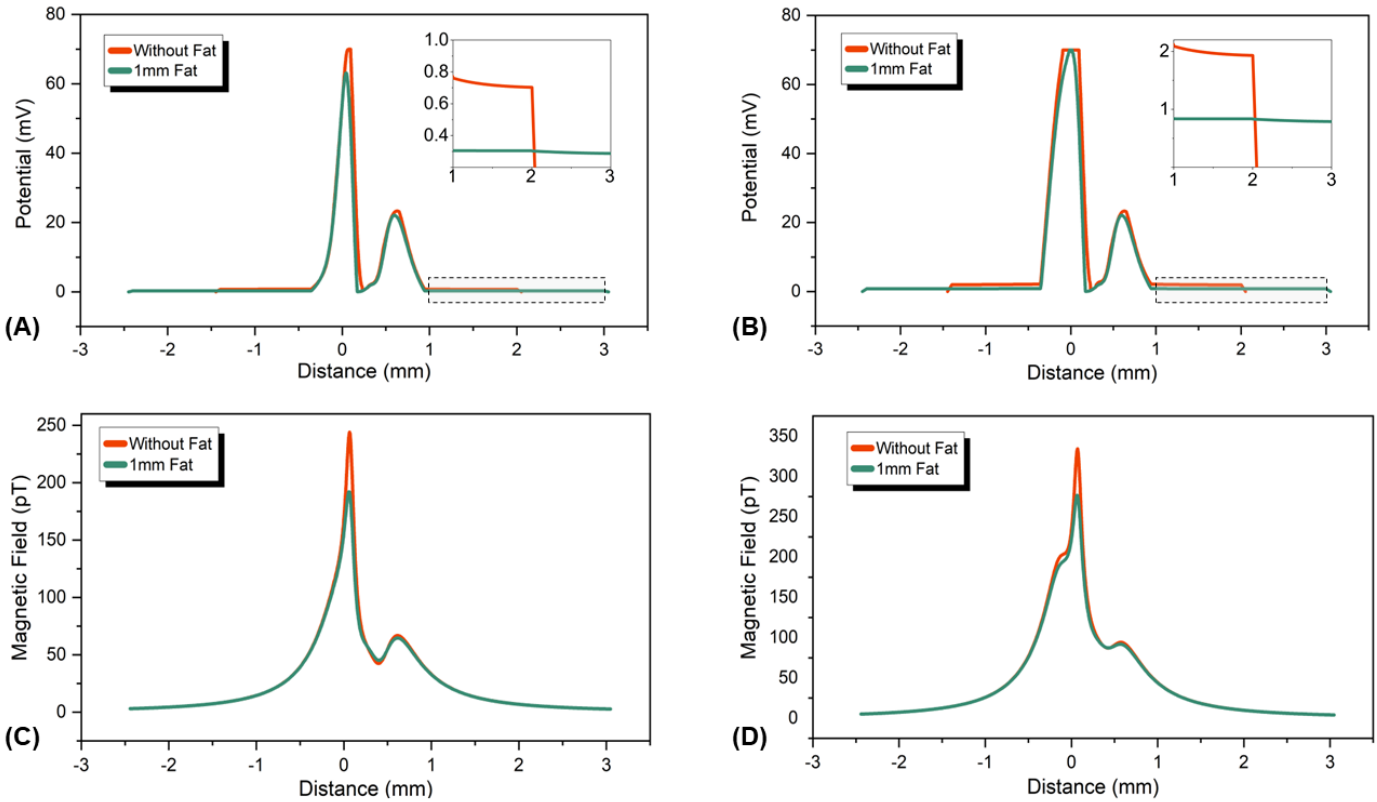


Fig. 5: Electrical field distribution without and with 1 mm fat: (A) activated all small muscle fibers and (B) four large muscle fibers were added. Magnetic field distribution without and with 1 mm fat: (C) activated all small muscle fibers and (D) four large muscle fibers were added.

muscle fibers are activated, by adding 1mm fat, while the electrical signal is declined by 60%, the magnetic field doesn't change at all. The same result also can be seen in Fig. 5B when four large muscle fibers from fascicle 1 are recruited, though, the electrical field is decreasing by 58%, the magnetic field is not changing while going through 1 mm fat.

IV. Conclusion and Future Work

In this paper, first, eight small muscle fibers were activated and then four large muscle fibers following the size principle, were recruited. Then, the recorded electrical fields and its magnetic counterpart on the skin are compared to evaluate the effect of recruiting new muscle fibers on the recorded signals. Recorded electrical signals are not linearly proportional to the increased number of muscle fibers as they interfere with each other. However, magnetic signals do not interfere with one another. Moreover, the effect of several layers between skin and the source on the recorded electric and magnetic signals is also discussed which shows that the layers separating the skin and signal source act as a "way-open" for the magnetic field but change the measured electrical field. Further work would be investigating the effects of several fundamental scenarios (e.g., fat and connective tissue thicknesses, number and geometry of muscle fibers and sensor locations) on EMG and MMG signals.

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