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Analysis of tissue electrical properties on bio-impedance variation of upper limps

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Abstract: Upper limb loss has a significant impact on individual socioeconomic life. Human-machine interface (HMI) using surface electromyography (sEMG) establishes a link between the user and a hand prosthesis to recognize hand gestures and motions which allows the control of robotic machines and prostheses to perform dexterous tasks. Numerous methods aimed to enhance hand gesture and motion recognition toward an HMI. Bio-impedance analysis (BIA) is a noninvasive way of assessing body compositions and has been recently used for hand motion interpretation using 'brute force' pattern recognition. The impedance variation in the body mostly depends on the precise stimulation using appropriate electrical features of the associated tissue layers. It has been reported that the electrical properties of these layers varied significantly. Thus, it is essential to investigate the influence of these variations on the stimulator design for the hand motion interpretation. This may not be possible using experimental approaches. Alternatively, using highly advanced computational models, this can be readily investigated by attaining the available range of the electrical properties of each tissue layer and applying appropriate boundary conditions and simulation settings. The computational models are composed of a volume conductor of the human arm model and electrode settings. Also, two different computational study methods were used to determine the influence of the tissues' dielectric properties on the results. The quasistatic approximation was used by only considering the resistivity of the anatomical layers and the transient simulation was used to analyze the capacitive impact on the results. Finite element (FE) models were developed to simulate the potential distribution inside the skin, muscle, and bone layers of the upper arm for given electrode settings. Then, simulation results were recorded for various electrical properties and different study types. It was shown that the capacitive influence of the tissue may not be ignored for certain conditions due to significant variation in the induced electrical potential variation along with the target muscle. Also, the influence of the individual tissue's electrical properties was investigated using a set of dielectric parameters. The results showed that the skin and muscle layers have a significant impact on the electrical potential variation across the muscle length.

Key words: Bio-impedance analysis, computational human arm model, dielectric property, finite element simulation, human-machine interface, upper limb loss

1. Introduction

The human hand is a powerful tool and allows human beings to accomplish sophisticated movements. Upper limb loss can significantly affect the capability of performing daily living, working, and social activities [1]. This is a global constraint and over 3 million people worldwide suffer from it, and this number is expected to double

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by 2050 [2]. Using prostheses may improve people's quality of life who suffer from upper-limb loss [3]. Surface electromyography (sEMG) is the human-machine interface (HMI) method used in myoelectric prostheses [3, 4]. Electrodes are placed on the desired skin surface to record useful signals during target muscle contraction. While myoelectric prostheses are promising, user acceptance of these devices remains low due to addition of the noise signal to the main EMG signal [5]. Alternative noninvasive ways of recording raw bio-signals may provide further advancement in the field. Transcutaneous bio-impedance analysis (t-BIA) for hand gestures is an alternative noninvasive technique that applies an electrical current through the body for body composition measurements and assessment of clinical conditions [6, 7]. This method reduces the risks and costs associated with implanted devices. t-BIA injects a current and measures resulting voltage potentials as shown in Figure 1. Such current-induced voltage signals have a better signal-to-noise ratio (SNR) and are related to the body compositions that are underneath the electrode. t-BIA for HMI has been recently used for hand motion interpretation using 'brute force' pattern recognition [8, 9]. In such an HMI system, the electrodes are placed on the forearm and the BIA dataset is analyzed to reflect upper limb movement.

It has been shown that many factors may have an influence on t-BIA [1]. The variations in the electrical properties of the biological tissue may have a significant impact on the efficient design of the t-BIA system for HMI. When the current is applied from the electrodes that flow through the skin and underlying tissues where a potential field is generated depending on tissue electrical properties (e.g., resistivities and permittivities (capacitance)). In particular, it has been reported that the electrical features of the fundamental tissues, such as skin, muscle, and bone, are varied significantly based on different prosthesis' electrical parameters (e.g., simulation pulse duration) due to bio-impedance frequency-dependency [10–12]. Some studies included capacitive effect using transient models, while others neglected this feature by using static models [13–15]. The transient simulation incorporates the capacitive effects of the tissues, while the quasistatic approximation of the electrical potential distribution enables Maxwell's equations to be simplified by neglecting the capacitive, inductive, and wave propagation effects. The pioneers of this simplification used this for a certain frequency range (e.g., 1kHz) and they mainly focused on applications where the activity of tissue is recorded at the body surface (e.g., EMG) [13].

The quasistatic solution can be used for certain conditions and may not always be appropriate as tissue acts as a frequency filter [13, 16]. Also, it has been stated that electrical potential variation is a nonlinear function, and differences in the spatiotemporal distribution of the potentials may affect stimulus efficacy. Thus, the transient solution has been suggested to apply for the frequency range beyond 1kHz [16, 17]. Although this was investigated for various applications [16, 18], it has not been addressed by any previous examination of the transient and quasistatic solutions for the electrical potential variation of hand motion interpretation. Thus, the capacitive effects of the bulk tissues may have an effect and need further investigation before being neglected. Therefore, there is a need to investigate the impact of these variations on the t-BIA to obtain the optimal stimulation strategy.

The purpose of this investigation is to analyze the impact of the dielectric features of the fundamental tissue layers of the human arm on the bio-potential variations of the hand motion interpretation for upper limb loss using different solution methods. It is required to design an efficient and systematic method to parametrize the electrical features of these tissue layers. This may not be feasible using preclinical and clinical studies. Alternatively, computational modeling provides an important toolset for designing and analyzing these parameters to design enhanced bio-medical devices for biological disorders and diseases [14, 19, 20]. Current

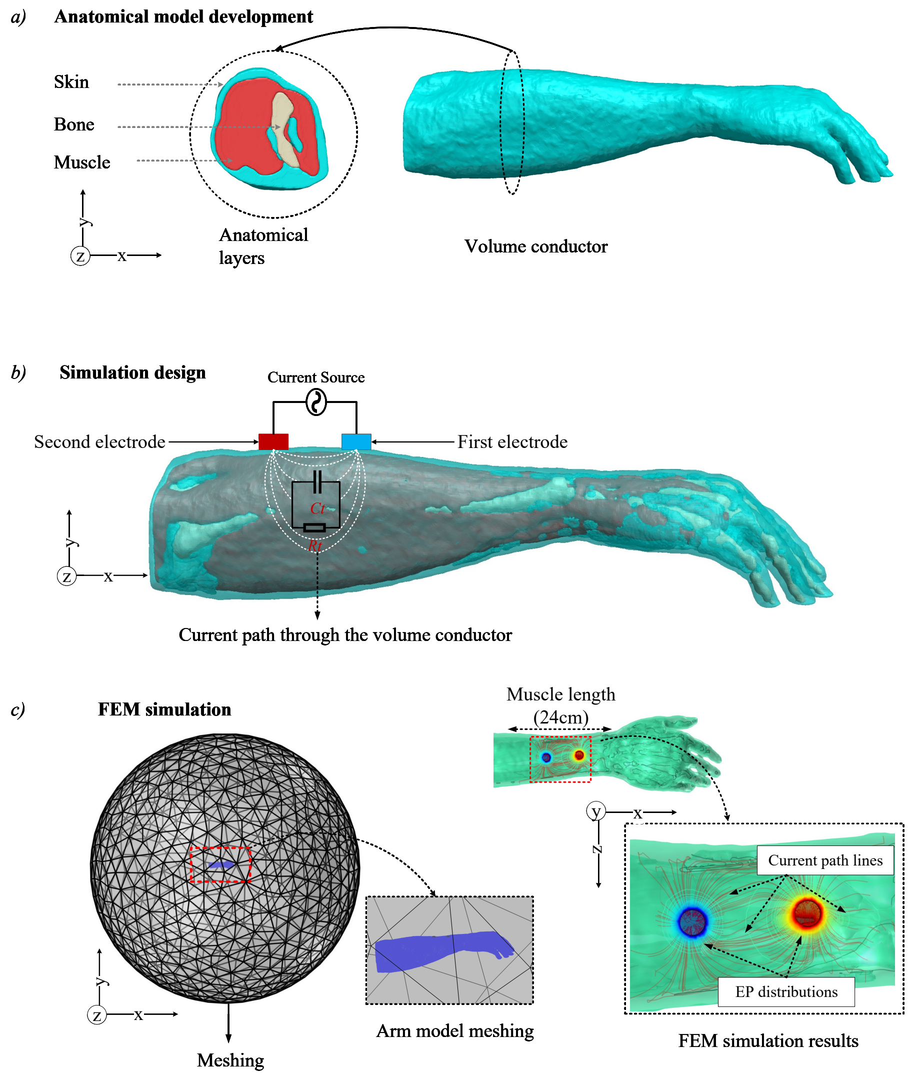

Figure 1. (a) Showing generation of volume conductor of arm model. Associated tissue layers are highlighted and labelled. (b) Showing simulation design. The current path through the volume conductor is highlighted in white. C_t and R_t show capacitance and resistance of the associated tissue, respectively. (c) Showing FEM simulation. Discretization of the whole model was shown and the current and electrical potential (EP) distribution within the arm model is highlighted. The potential variation was calculated along the target muscle length (24 cm).

commercial finite element method (FEM) software packages (e.g., COMSOL Multiphysics, ANSYS) allow calculating electrical potential distributions in the computational models. FEM has matured as a numerical approach for solving bioelectric field problems with complex geometric features and anisotropic tissue properties [21, 22]. The typical workflow for computational modeling of a t-BIA device involves a three-dimensional FEM of the electrode and nearby tissues to calculate the distribution of electric potentials in the tissues using appropriate boundary conditions as shown in Figure 1. The three-dimensional (3D) volume conductor of the arm model was generated based on the Duke model for the transcutaneous electrical stimulation (TES) that also incorporates the capacitive effects of the most important tissues as shown in Figures 1a and 1b. The available range (maximum, standard, and minimum) of the conductivities and permittivities of each tissue layer of the arm model were used to compare the quasistatic FE approximation and transient FE solutions. This led to deciding which solution method is optimal for further investigation. The findings showed that the capacitive effect cannot be neglected for certain conditions. Thus, the transient solution was used to determine each tissue's influence on the electrical potential distribution for hand motion interpretation. Then, the results were calculated using three ranges (minimum, standard, maximum) of the dielectric properties of these layers. These ranges were parametrized to examine each tissue's influence on the electrical potential distributions along the target muscle length. Each model was simulated and results along the muscle length (24 cm) were recorded as a sample shown in Figure 1c.

2. Methods

All simulations have been carried out using the FE tool in COMSOL Multiphysics, which is a module-based FE software widely used in physics and engineering design and optimization strategies. The AC/DC and Design modules were used for the electric simulations. The stationary solution method was used to calculate quasistatic approximation results and the transient solution was applied to analyze the evolution of the impedance variation inside the human upper arm over time. These are detailed as follows.

2.1. Electrical stimulation design

The current pulse was applied to the electrode through associated anatomical layers to analyze the impact of the electrical features (e.g., conductivity and relative permittivity) of anatomical layers on the t-BIA. Electrical impedance on the muscle was estimated by measuring the electrical potential distribution across the muscle length. Electrical stimulation was designed using a set of electrodes in contact with the skin as shown in Figure 1b. The current was applied at the surface of these electrodes which leads to an electric field being set up in the underlying tissue. Since the electrode features (e.g., electrode size, electrode spacing) may have a significant impact on the electrical potential variation across the target muscle, the single size of the electrode (electrode diameter = 12 mm, electrode spacing = 50 mm) was designed to investigate the influence of the different types of the electrical stimulation and electrical properties on the impedance variation on the associated muscle as shown in Figure 1b. In the TES system, different electrode shapes can be used. In this study, a circle smooth electrode was designed to prevent any possible edge effects. In all the following cases, anodes were set as the source of a total of 1 mA while the same amount was set to sink in cathodes. This was implemented in COMSOL by terminal current of negative value for the cathodes and positive value for the anodes. Two different types of simulations have been considered; quasistatic simulation and transient time-domain simulation as detailed in the following sections. Also, it has been shown that the significant differences in electrical parameters for muscles in a different direction (the axial direction compared to the radial direction) may have a notable influence on the simulation results [23]. Thus, it is important to include anisotropy of the electrical properties of the associated materials to get a qualitative understanding of the processes involved, and accurately interpret the outcome of the simulations accordingly. A set of properties for biological tissue have been chosen for simulations presented in this report based on comparing parameters from different sources for TES [10–12, 17] as shown in Table 1.

2.2. Quasistatic simulation

The quasistatic simulation was applied using COMSOL Multiphysics settings while considering the quasistatic approximation of Maxwell equations demonstrated by the Laplace equation as shown in (3). Using this

Table 1. Tissue electrical properties. The standard values were taken from [17] based on 5 kHz and minimum and maximum properties were extracted from [10–12] based on 500 Hz and 50 kHz, respectively.

Tissue	Cond (S/m)			Permitivity (F/m)		
	Min	*Std*	*Max*	*Min*	*Std*	*Max*
Skin	0.00016	0.0014	0.002	1e3	6e3	3e4
Muscle axial	0.2	0.33	0.5	1e5	1.2e6	2.5e6
Muscle radial	0.066	0.111	0.166	3.3e4	4e4	8.3e5
Bone	0.016	0.02	0.025	2.4e3	3e3	3.6e3

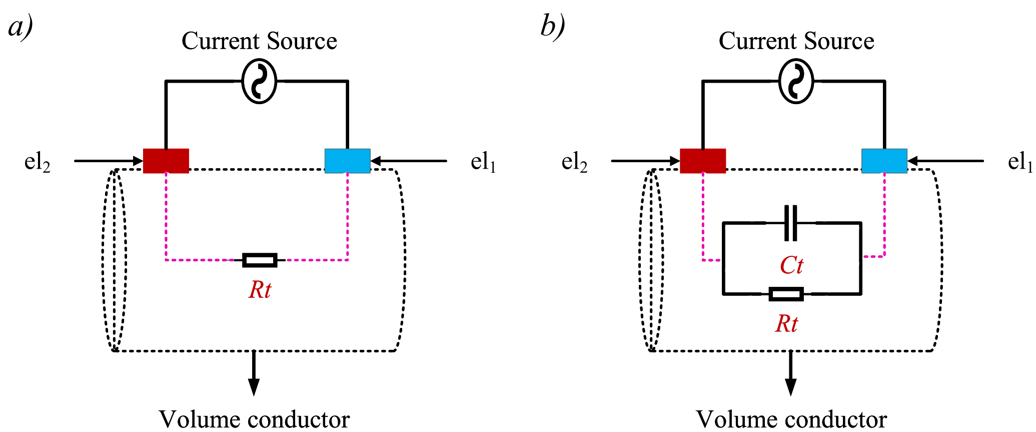

Figure 2. Two different simulation methods are shown. (a) The quasistatic method only considers resistance (R_t) of the tissue during simulation. (b) Using transient simulation allows considering the capacitive impact of the tissue on the results by taking account of C_t. The current pulse is applied through electrodes to the volume conductor. el: Electrode.

approximation resulted in neglecting the influence of capacitive fields on the electric field by setting the existence of any free charges to zero in the volume conductor as shown in Figure 2a. The software solves the following Maxwell's equations using stationary in electric currents settings to simulate the electric potential in the model for the quasistatic simulation.

$$J = \sigma \cdot E + J_e \tag{1}$$

$$\nabla \cdot J = Q_j \tag{2}$$

$$\nabla \cdot (\sigma \nabla V) = 0 \tag{3}$$

where is J the current density, Q_j is the current source, E is the electric field, and J_e is the external current density. To obtain the electrical potential variation on the target muscle based on the quasistatic simulation, Q_j and J_e were set to zero everywhere in the model. Thus, the electrical potential is mainly varied depending on tissue conductivity (σ). A range of the values that were reported for the conductivity of skin, muscle, and bone was shown in Table 1. The set of the conductivity of these layers was attained for each simulation. It is noted that anisotropy properties of the muscle were used for more accurate simulation. A comparatively large nonconductive ($\sigma = 10e^{-10} Sm^{-1}$) sphere was defined as an external boundary and the Dirichlet boundary condition (V = 0) was applied, which was considered an approximation of the ground at infinity.

Initially, three different scenarios were applied using minimum, standard, and maximum tissue electrical properties to compare the transient and quasistatic simulation results. These scenarios are shown below.

$$\textbf{Simulation scenarios} \Rightarrow \begin{bmatrix} Skin_{min(\sigma),max(\epsilon_r)} & Muscle_{min(\sigma),max(\epsilon_r)} & Bone_{min(\sigma),max(\epsilon_r)} \\ Skin_{std(\sigma,\epsilon_r)} & Muscle_{std(\sigma,\epsilon_r)} & Bone_{std(\sigma,\epsilon_r)} \\ Skin_{max(\sigma),min(\epsilon_r)} & Muscle_{max(\sigma),min(\epsilon_r)} & Bone_{max(\sigma),min(\epsilon_r)} \end{bmatrix}$$

Each row (e.g., $Skin_{min(\sigma),max(\epsilon_r)} Muscle_{min(\sigma),max(\epsilon_r)} Bone_{min(\sigma),max(\epsilon_r)}$) shows one simulation and the other rows indicate the remaining simulation scenarios.

FEM was used to calculate the electrical potentials within the volume conductor by discretizing the domains using free tetrahedral elements. The anatomical layers and electrodes were more finely meshed, while the ground domain (sphere) was relatively coarsely meshed to obtain more accurate potential distributions in a reasonable time. This resulted in about 1.5 million tetrahedral elements and about 2 million degrees of freedom.

2.3. Transient (capacitive) simulation

$$E = -\nabla V \tag{4}$$

$$\nabla \left[J + \frac{\partial D}{\partial t} \right] = 0 \tag{5}$$

$$\nabla \cdot \left[\sigma \nabla V - \epsilon_0 \epsilon_r \nabla \frac{\partial V}{\partial t} \right] = 0 \tag{6}$$

The electrical field in the biological volume conductor can be described by equation (4). Since only the electrical potential variation is of interest, wave propagation and inductive effects in the volume conductor can be neglected [13]. Thus, the equations can be simplified using the continuity equation (5). In this way, the transient differential equation (6) that is describing the transient electric scalar potential V in a volume conductor can be derived. This is composed of resistive (σ) and the dielectric properties ($\epsilon = \epsilon_0 \epsilon_r$) of the tissues as shown in Figure 2b.

The influence of the capacitive effects on the impedance variation along the muscle was investigated by changing the permittivities and conductivities within the range of reported values in Table 1 which was designed based on the range of values that is available in the literature that can be expected in practical applications of TES [23]. The charge-balanced biphasic current pulse with 10μs, 100μs, and 1000μs was used to compare solution methods. The simulation frequencies, in turn, were 50 kHz, 5 kHz, and 500 Hz. It is noted that the anisotropy conductivity and dielectric properties of the muscles were considered to measure more accurate results.

It was assumed that the magnitude of the electrical potential does not vary based on the stimulus pulse. The impact of each tissue layer on the impedance variation was investigated by considering the following criterion using 10μs pulse for each simulation. For example, the influence of the skin was analyzed by keeping constant electrical properties (using standard values) of the muscle and bone and altering the value for the skin. In the following example, the standard (std) value of the electrical features (σ, ϵ_r) was used for the muscle and bone and, in turn, minimum, standard, and maximum values of the electrical properties of the skin were used to analyze the impact of the skin on the results. The same method was followed to investigate the influence of the muscle and bone. The results were recorded accordingly to examine which tissue mostly affects the electrical potential distribution across the muscle length.

$$\textbf{Skin's influence} \Rightarrow \begin{bmatrix} Skin_{min(\sigma),max(\epsilon_r)} & Muscle_{std(\sigma,\epsilon_r)} & Bone_{std(\sigma,\epsilon_r)} \\ Skin_{std(\sigma,\epsilon_r)} & Muscle_{std(\sigma,\epsilon_r)} & Bone_{std(\sigma,\epsilon_r)} \\ Skin_{max(\sigma),min(\epsilon_r)} & Muscle_{std(\sigma,\epsilon_r)} & Bone_{std(\sigma,\epsilon_r)} \end{bmatrix}$$

3. Results

3.1. Compare transient and quasistatic solutions

The results based on electrical potential versus muscle length for different simulation methods are shown in Figure 3. The results for the different electrical properties of the tissues are highlighted in different colors. Figures 3a and 3b illustrates the transient and quasistatic solutions, respectively, for a given set of properties for biological tissues. As it is observed in the plots, the maximal potential value of each simulation appeared in proximity to the stimulating electrodes for both simulations. It is clearly shown that there is a difference between each simulation for both solution methods when the induced electrical potential ranges are compared. The variation in the induced electrical potential on the muscle length is proportional to tissue's conductivity for all simulations in Figure 3a as agreeing with (6). The maximum electrical potential is observed when the maximum tissue's conductivity was used. Also, if the minimum tissue's conductivity is used, a comparatively less electrical potential is calculated. The maximal potential for each simulation for maximum, standard, and minimum tissue's conductivity is, in turn, 1.2 V, 0.9 V, and 0.18 V based on the transient solution method. Thus, the maximal electrical potential difference is about 1 V.

The electrical potential variations in Figure 3b indicate inverse-proportional variation regarding the electrical conductivity of the tissue. Using maximum tissue's conductivity provides relatively minimum electrical distribution and using minimum tissue's conductivity induced maximum electrical potential variation along the muscle length as agreeing with (3). Although the results are nearly the same when the maximum and standard electrical properties of the tissue were used, there is a significant induced electrical potential along the muscle length if the minimum electrical conductivity was used based on the quasistatic solution method. The maximal electrical potential value for the maximum, standard, and minimum conductivity is 1.25 V, 1.8 V, and 10 V, respectively.

It can be deducted from the results that there is a significant difference in the induced electrical potential between transient and quasistatic solutions. The induced electrical potential variation range is –0.7 V to 1.2 for the transient solutions, while this range is –1.2 V to 10 V for the quasistatic solutions. It can be derived that the relative permittivity effects (capacitive effect) on the results cannot be ignored. Although the transient solutions relatively provide accurate results, it is time-consuming compared to the quasistatic solutions. The solutions, on average, were obtained 45 min for the transient solution while this was approximately 1 min for the quasistatic solutions (not shown).

3.2. Impact of tissue electrical properties

As it was shown that the tissue capacitive effect cannot be neglected and using the transient solution provides more. Thus, this solution method was solution method was used to determine the influence of each tissue's

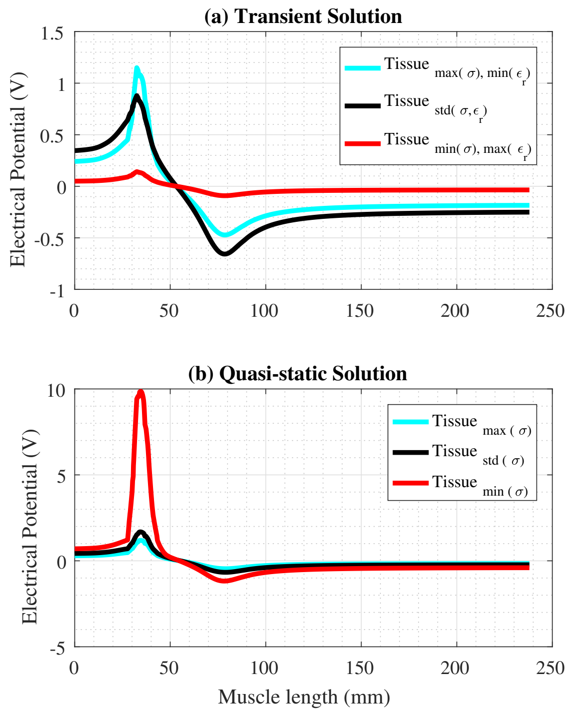

Figure 3. Different solution methods for the FEM of the arm. (a) Showing electrical potential variation across the muscle length for different electrical properties of the anatomical tissue for the transient solutions. The current pulse 10 μs was used for the maximum, 100 μs was used for standard and 1000 μs was applied for minimum electrical conductivity (σ). (b) Showing electrical potential variation across the muscle length for different electrical properties of the anatomical tissue for the quasistatic solutions. The variations are highlighted. ϵ_r shows the relative permittivity of the tissue within the volume conductor.

electrical properties on the results. The impact of the dielectric properties of the skin, muscle, and bone is shown in Figures 4a–4c, respectively.

It is shown that each tissue affects the results differently. Although the variation in the potentials is proportional to the skin's conductivity, this is not valid for the muscle and bone for all simulations. The maximum electrical potential variation along the muscle is recorded when the standard dielectric properties of the muscle and bone were used while this variation for the skin was recorded based on the maximum tissue's conductivity. It is also shown that although the variation in the dielectric properties of the skin and muscle has an impact on the results, the electrical potential variation is not affected by the parametrization of the bone's dielectric properties. The maximal electrical potential is roughly the same when the electrical properties of the bone were changed. It is noted that the variation in the electrical potential variations is comparatively minimum (as highlighted in red in subfigures in Figure 4) for all the tissue parametrizations when the minimum electrical conductivity of the tissue was used. The variation of the skin's dielectric properties resulted in different maximal electrical potentials that range from 0.5 V to 1.2 V. This range is 0.25 V to 0.9 V for the muscle. The range for the bone is approximately the same based on different electrical features.

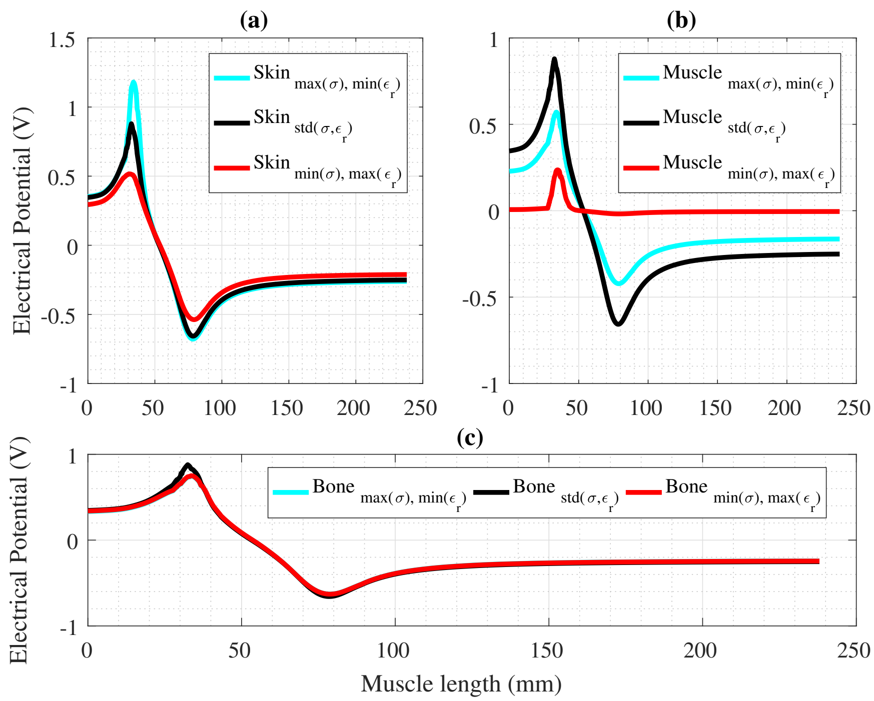

Figure 4. Analysis of the impact of the electrical properties of the tissue on the electrical potential variation. (a) Showing the influence of the skin on the results using standard electrical properties of the skin and using parametrization of the remaining tissues' electrical properties. (b) Showing the effect of the muscle on the results using standard electrical properties of the muscle and using parametrization of the remaining tissues' electrical properties. (c) Showing the impact of the bone on the results using standard electrical properties of the bone and using parametrization of the remaining tissues' electrical properties. The electrical potential variation is highlighted in a different color for each simulation. σ shows electrical conductivity, ϵ_r shows relative permittivity of the tissue.

4. Discussion

The fundamental goal of this study was to perform a quantitative evaluation of a range of electrical properties of the tissue on the impedance variation for t-BIA of HMI using different simulation strategies. Since the results that have been produced so far are mixed [24–26], there is a need to analyze the impact of the electrical properties and simulation strategy on the impedance variation for the hand motion interpretation to optimize stimulation devices. A computational anatomical model of the arm was generated to readily quantify the influence of these parameters on the defined muscle length.

The current was injected through the electrodes and the induced electrical potential on the muscle was calculated for each simulation. In the first scenario, the transient and the quasistatic simulations were compared based on a different range of the tissue's dielectric properties. The results showed that there was a significant difference between the two solution methods. In particular, there was a considerably electrical potential difference when the minimum conductivity of the tissue was used as shown in Figure 3. This can be related to equation 3. As can be derived from the equation the conductivity of the tissue is inverse-proportional with the electrical potential variation. Thus, using minimum conductivity resulted in maximum electrical potential variation based on the quasistatic solution. However, this is not valid for the transient solution of the FEM. Although there was no notable difference in potential variation across the muscle length for 100 μs (5 kHz), 1000 μs (500 Hz) current pulses, there was a significant difference between the two solutions methods for

10 μs (50 kHz). These results are proven and are in agreement with those of the previous studies [8, 16, 17] that the electrical properties of the tissue are frequency-dependent; thus, the capacitive effect of the tissue may not be neglected after a certain frequency range [16].

In the second scenario, the transient response of the given current level was quantified using the standard dielectric properties of each tissue and using the parametrization of the remaining tissue's dialectical properties based on available literature [11–13]. The results suggested that there is a significant impact of the tissue's dielectric properties on the variation of the electrical potential along the muscle length. It was shown that the variation in the skin's dialectical properties has resulted in nearly 300% and this was about 400% for the muscle's electrical properties. These significant variations may indicate that skin and muscle tissue layers are the most important ones during designing HMI for the BIA of hand motion interpretation. However, it was shown in Figure 4c that the variation in the bone dielectric properties may not affect the outcome. This is maybe related to the territory of the muscle with respect to the muscle length as it is located to the inner of the arm. Also, it has been shown in the muscle that there has been a combination of both short rise times of the potential long rise times in the FE model [17].

Overall, the results of this study suggested that the capacitive effect of the tissue should not be ignored when HMI is designed for the BIA. Thus, although the transient solution method is time-consuming compared to the quasistatic methods, it is essential to use this method to obtain more accurate results and design reliable and safe biomedical devices accordingly. Also, it was shown that it is important to consider that the variation in the dielectric properties of the tissue may affect the results significantly.

Since the aim of this study was to investigate the impact of impedance variation based on different electrical features of the upper limb layers using different computational study methods, the statistical analysis of the results was not studied due to insufficient samples.

In this study, three different rages of the electrical properties of each tissue were used to investigate their impact on the electrical potential variation across the muscle length. Although the results are promising and indicated that there was a significant influence of these parameters, it may be required to investigate this for a different range of these parameters. Since there are large variations in the properties of the biological materials based on the available literature, more accurate knowledge of electrical and biological tissue properties is needed to draw a conclusion.

5. Conclusion and future work

The present work has illustrated a comparison between computational solution methods and examined the influence of the electrical properties of the anatomical layers in the volume conductor for hand motion interpretation using a set of the electrical properties of the associated tissue layers of the arm. The computational results showed that an important qualitative conclusion can be drawn. In the first step, the transient FEM solutions were compared to the quasistatic FEM results using three different scenarios of the electrical properties of the tissue layers. It was shown that there was a significant difference in the induced electrical potential variation across the muscle length. Thus, the capacitive effect of the tissue may not be neglected for certain conditions. Therefore, the results suggested that the transient solution should be used to obtain more accurate results for designing safe and effective HMI of BIA for hand motion interpretation.

In the second step, the influence of the individual tissue layer was analyzed using their dielectric properties based on the transient FE simulations. To examine the influence of a tissue layer on the electrical potential variation, the standard electrical properties were used for this tissue layer and the dielectric properties were parameterized for the remaining tissue layers. It was shown that although the electrical potential variation was not changed significantly when analyzing the influence of the bone on the result, the results showed that the skin and muscle considerably have an impact on the electrical potential variations.

More accurate results and detailed conclusions may be drawn by considering a more elaborate range of the dielectric properties of the tissue and using different sets of electrode sizes. Then, the computational results should be validated with the experimental tests.

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