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Research article

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RESEARCH

Parameterization of physical properties of layered body structure into equivalent circuit model

Jiho Lee^{1,2} and Sung-Min Park^{1,2,3*}

Abstract

Background: This study presents a novel technique to develop an equivalent circuit model (ECM) for analyzing the responses of the layered body structure to transcutaneous electrical nerve stimulation (TENS) by parameterizing electrical and geometrical properties. Many classical ECMs use custom meta-parameters instead of the physically driven parameters because of the difficulty in projecting physical properties directly into ECM. However, the difference in what parameters are customized hampers general agreement in modeling the responses to TENS. To overcome this limitation, we propose a tissue property-based (TPB) approach for the direct parameterization of the layered body structure.

Results: The proposed method was first validated through in vitro phantom studies and then was applied in-vivo to analyze the TENS on the forearm. The TPB-ECM calculated the impedance network in the forearm and corresponding the responses to TENS. In addition, the modeled impedance was in good agreement with well-known impedance properties that have been achieved empirically.

Conclusions: The TPB approach uses the physical parameters instead of meta-parameters, thus overcoming the disagreement problem of conventional ECMs. Therefore, the TPB-ECM has a potential for widely-applicable TENS analysis and could provide impactful guidance in the TENS parameter design.

Keywords: equivalent circuit model; electromagnetics; electrical nerve stimulation; multi-layer body impedance; direct parameterization of conductivity, permittivity, and thickness; layer-by-layer dynamic impedance

Background

Transcutaneous electrical nerve stimulation (TENS) has been in use as a clinical neuromodulation modality for decades. There are many FDA-approved TENS devices in the market [1–4]. While the TENS technique has shown clinically meaningful effects, the pulse parameters such as shape, intensity, and repetition rate have not been standardized for specific applications [5–7]. The lack of standardization on the stimulation parameter continues to be the problem leading to no universal agreement on the electrical responses of each body tissue layer to TENS. Determining the electrical responses of the body makes it possible to design the TENS parameters accurately to minimize the unwanted energy loss in the background tissues and thus overcome the safety concerns of TENS such as pain and skin irritation [8–10].

The equivalent circuit model (ECM) is an effective method to determine the TENS responses of a body impedance network [11–14]. Especially for the multi-frequency components in TENS pulses, the ECM provides efficient computation to analyze the comprehensive responses to the frequency-dependent body impedances, and thus is better compared to the volume conductor approaches (ex. finite-element method (FEM)), where 3D Maxwell's equations are solved at each frequency separately and then integrated to see the final effect [15–18]. Despite being very useful, the main limitation of conventional ECM is that it uses meta-parameters instead of the physically electrical and geometrical properties of each tissue layer [19,20]. Many researchers have performed in vitro and in vivo impedance experiments to optimize the meta-parameters [14,16,21]. However, the meta-parameter sets differ as to the equivalent circuit design and the impedance data. Due to a lack of agreement over customized parameter sets, it has been challenging to develop a generally applicable ECM. Thus, advances

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to the ECM are needed for analyzing the body responses by discovering the straightforward relationship between the body impedance and the physical tissue properties.

For parameterizing the physical tissue properties, the effects of 3D electromagnetic field distribution need to be projected into the 1D impedance elements. The main bottlenecks in this dimensional transition are the modeling of nonlinear effects of geometrical property on the impedance and the interactive effects among the electrical property of neighboring layers [22–25]. To overcome these challenges and develop the ECM with the physical parameters, we propose a novel tissue-property-based (TPB) method, which utilizes the hybrid techniques of computation and analytics. We modeled each tissue layer impedance as a complex capacitor parameterizing each layer's effective permittivity, which in turn is affected by the corresponding underlying layer. This interaction between the neighboring layers can't be modeled by existing ECM methods. Then, the network connection among capacitors was derived with the partial capacitance (PC) method [24]. We finally complete the model by computationally parameterizing the effect of geometrical tissue properties. We validated the developed model through in vitro experiments and then applied the model in-vivo to analyze the TENS on the forearm.

Methods

Analytical method to parameterize the effective electrical property into ECM

Due to the interactions among the neighboring layers, the effective electrical properties of each layer are different from their own properties. The following section describes analytical processes for parameterizing the effective electrical properties and importing them into ECM.

Complex permittivity to simplify the electrical property of a tissue layer

The complex permittivity simplify two representative electrical properties of conductivity and permittivity analytically ($\epsilon^* = \epsilon - j\frac{\sigma}{\omega}$, where ϵ^* is complex permittivity, ϵ is real permittivity, σ is conductivity, and ω is angular frequency) into a single property. Correspondingly, the RC impedances of a tissue layer can be simplified into a single complex capacitor. The impedance is given by:

$$Z_{tot} = R || \frac{1}{j\omega C}$$

$$\therefore = \frac{1}{j\omega(\epsilon - j\frac{\sigma}{\omega})\frac{A}{d}} = \frac{1}{j\omega\epsilon^*\frac{A}{d}} = \frac{1}{j\omega C^*}. \quad (1)$$

where $R(= d/\sigma A$, A is the area, and d is the distance between source electrodes) is the resistance, $C(= \epsilon A/d)$ is the real capacitance, and $C^*(= \epsilon^* A/d)$ is the complex capacitance. This simplification enables the application of an analytical method, called the partial capacitance. The method can model the effective complex permittivity of each tissue layer where the interactions among neighboring layers are considered.

Partial capacitance method to parameterize the effective complex permittivity into capacitor network

The partial capacitance (PC) method decomposes the total effective capacitance of the layered structure as the sum of each layer capacitance [22–24]. This physical projects to model a capacitor network of the structure with layer-by-layer partial capacitors. For each partial capacitor, the PC method decides its direction, whether parallel-to-layer (current shunting along with the layer) or serial-to-layer (current penetrating through the layer) (Fig. 1). The direction of a layer capacitor is dependent on the permittivity ratio between the layer and its underlying layer. In the first category, the parallel partial capacitance (PPC) condition is the case when the upper layer (Layer₁) has a higher permittivity than the lower layer (Layer₂) with layer interface under Neumann boundary condition (Fig. 1a) [22]. The PPC for the total capacitance is defined as:

$$C_{tot} = C_1 + C_2$$

$$= (\epsilon_1 - \epsilon_2)g_1 + \epsilon_2g_2, \quad (\epsilon_1 > \epsilon_2) \quad (2)$$

where C_{tot} is the total capacitance, C_i is the Layer_{*i*} capacitance ϵ_i is the Layer_{*i*} permittivity, and g_i is PPC geometric factor of Layer_{*i*}. The geometric factor (GF) of each layer is defined as:

$$g(x_1, x_2, x_3, \dots) = \frac{C}{\epsilon} \quad (3)$$

where x_i is the geometric variables, such as the thickness of layer, the electrode size, and the distance between electrodes. In the second category, serial partial capacitance (SPC) is the case when Layer₁ has a lower permittivity than Layer₂ with satisfying the interface under Dirichlet boundary condition (Fig. 1b). The SPC for the total capacitance is defined as:

$$\frac{1}{C_{tot}} = \frac{1}{C_1} + \frac{1}{C_2}$$

$$= \left(\frac{1}{\epsilon_1} - \frac{1}{\epsilon_2}\right)\frac{1}{g_1} + \frac{1}{\epsilon_2g_2}, \quad (\epsilon_1 < \epsilon_2) \quad (4)$$

Figure 1 Principle of PC method. a) Parallel partial capacitance (PPC), b) Serial partial capacitance (SPC).

Figure 2 Complex permittivity measurement of salted and pure agar material and comparison of phantom capacitances. a) magnitude of the complex permittivities, b) phase of the complex permittivities, c) PPC phantom capacitance comparison, and d) SPC phantom capacitance comparison. (mean ± 2 std for b, c, and d)

where g_i' is SPC GF of Layer $_i$, and the other variables are similar to PPC formula. Moreover, the researches have demonstrated that the approach is valid when using the complex capacitors [24]. The PC method formulas in equation (2) and (4) implies that the effective (complex) permittivity of Layer $_1$, which is affected by Layer $_2$, can be described as $\varepsilon_1 - \varepsilon_2$ for PPC and $1/\varepsilon_1 - 1/\varepsilon_2$ for SPC. The method also describes the connection information between the capacitors of two neighboring layers, and thus parameterizes the effective complex permittivity of each layer to the ECM network consisting of complex capacitors.

Computational method to parameterize the effects of geometrical property

To parameterize the remaining effects of geometrical properties into the PC method formulas in equation (2) and (4), 3D electromagnetic simulations were utilized because the pure analytical derivation of the 3D GF terms requires redundantly complicate processes (the electric field distribution follows the elliptic functions and their integral would be almost impossible). Therefore, to avoid unnecessary complexity but complete the TPB-ECM, we used regression techniques on 3D 1simulation results to find the nonlinear geometrical effects. The detailed idea and process are the following.

Interdimensional factor (IF): modified GF for error complement during the 3D expansion of PC method

Our preliminary results showed that the GF terms have secondary fluctuation errors when expanding the original 2D PC equations to 3D (see Supplementary section 1). Even though the errors are secondary effects, they still reduce the accuracy of the 3D PC equations. As to eliminate the errors and increase the accuracy of the TPB model equations, a concept of modified GF was newly defined and named as an interdimensional factor (IF). The IFs are assumed to be affected by both Layer1 thickness (h_1) and the permittivities ratio between Layer1 and Layer2 ($\varepsilon_2/\varepsilon_1$) because the fluctuation errors in 2D-to-3D expansion of PC method depend on the layer permittivities (see Supplementary section 2).

Regression for IF and final TPB model equations

3D simulation data with the inputs of ($h_1, \varepsilon_1, \varepsilon_2$) and the output of (C_{PPC} or C_{SPC}) were applied to the regression model to formulate IFs. Using the formulas of 2D Layer $_1$ GF (PPC: $\frac{C_{PPC} - \varepsilon_2 g_2}{\varepsilon_1 - \varepsilon_2}$, SPC: $\frac{1/\varepsilon_2 - 1/\varepsilon_1}{1/C_{SPC} - 1/\varepsilon_2 g_2}$), the regressions were performed for each of PPC IF and SPC IF. For PPC IF, the logarithmic regressions were performed and the final PPC model is described as:

$$C_{PPC}^{3D} = (\varepsilon_1 - \varepsilon_2) I_{PPC}^{3D} + \varepsilon_2 g_2$$

$$\text{with } I_{PPC}^{3D} = a_0 + a_1 \ln(\varepsilon_2 - \varepsilon_1) + a_2 \ln(h_1) \quad (5)$$

where I_{PPC}^{3D} is the PPC IF and a_i ($i=0, 1, 2$) are regression coefficients. For SPC IF, inverse-linear regression on h_1 and the logarithmic regression on $\varepsilon_2/\varepsilon_1$ were performed. Then the final SPC model is described as:

$$\frac{1}{C_{SPC}^{3D}} = \left(\frac{1}{\varepsilon_1} - \frac{1}{\varepsilon_2} \right) \frac{1}{I_{SPC}^{3D}} + \frac{1}{\varepsilon_2 g_2}$$

$$\text{with } I_{SPC}^{3D} = \frac{b_0 + b_1 \ln\left(\frac{\varepsilon_2}{\varepsilon_1}\right)}{h_1} + c_0 \quad (6)$$

where I_{SPC}^{3D} is the SPC IF and b_i ($i=0, 1$) and c_0 are regression coefficients. All regressions used the least-square method to calculate the coefficients.

As for the simulation environment, a FEM simulator (Electrostatic solver, Maxwell 3D, ANSYS, Ansys Inc.) with calculated the outputs for double-layered 3D virtual phantoms. The h_1 varied from 2 to 6 mm in 2 mm-step. The Layer2 thickness was 5 cm enough to assume infinity. The ($\varepsilon_1, \varepsilon_2$) were in 10 different pairs for each of PPC and SPC (Table 1). In addition, the size of the simulated electrodes and the distance between them was 1 cm x 1cm area and 1 cm length in constant.

Table 1 Relative permittivity ($\varepsilon_1, \varepsilon_2$) pairs for 3D simulation.

In vitro Validation of the TPB-ECM

In vitro experiments using double-layered agar phantoms were performed to validate the TPB model. The

Figure 3 Regression for interdimensional factor (IF) for importing the geometrical tissue properties into PC method formulas.

modeled capacitances of the agar phantoms were compared to the measured values (Fig. 2a).

Agar fabrication and complex permittivity measurement

To prepare phantom, agar powder (BD Difco Plate Count Agar, Fisher Scientific) was first mixed with DI water at a concentration of 23.5 g/L. Two different types of agar were used to construct the double-layered phantoms. One type (Agar_{Salt}) used DI water with 10 g/L NaCl (Sodium Chloride, Sigma-Aldrich), and the second type (Agar_{Pure}) used pure DI water. For Agar_{Salt}, the conductive ions of NaCl increased the conductivity [26], thus resulting in higher complex permittivity for compared to Agar_{Pure}. The mixture was then heated on a hotplate (MSH-20D, DAIHAN Scientific) to 80-90 celsius degrees to dissolve the powder. The heated solution was poured into a mold (glass, diameter 10 cm) and was cooled for more than 3 h to make it hard.

Complex permittivities of Agar_{Pure} and Agar_{Salt} were measured to use as inputs to the TPB model. Impedances of single layer structures of each agar (cylinders, 5 cm radius, 5cm thickness) were measured using an impedance analyzer (E4990A, Keysight Technology). The measurement ranged from 0.1 to 10 kHz in a log scale with 201 samples in between. Using equation (3), The complex permittivities were calculated for IF of structures with the FEM simulator. The measurement electrodes were 1 cm width x 1cm length, separated by 1 cm, and slightly coated with conductive gel (Spectra 360, Parker lab.).

In vitro experiments: complex capacitance of double-layered agar phantom

The developed TPB model was validated by comparing the total capacitances of the modelled and the measured double-layered agar phantoms for both PPC and SPC. The phantoms were cylindrical, and their dimensions were 5cm in radius, and 2 mm and 5 cm in thicknesses for Layer1 and Layer2, respectively. The experimental capacitances were calculated with the measured impedance using equation (1). The modelled capacitances of the phantoms were calculated using the measured agar permittivities with equations (5) and (6). Both the real and imaginary parts of the complex capacitances were compared to verify that both resistive and capacitive properties of the modeled capacitances agreed well with the experiments.

Representative application of TPB-ECM on forearm

As a representative application, in the forearm the impedance network was analyzed when the tissue properties were variable and the responses to TENS were also analyzed when the TENS pulse parameters were variable.

Impedance network analysis in forearm

The forearm tissue layers were classified into three categories: 1) skin, 2) fat, 3) muscle. The variances in the electrical and geometrical properties were applied: thicknesses of 1 to 3 mm in 0.5 mm-step for skin and fat (muscle was assumed to be infinite); permittivities of -20% to $+20\%$ in 10%-step for each tissue. The referred complex permittivities ($\pm 0\%$) were set as the standard dataset [27]. Moreover, the conductive gel impedance was assumed as 160Ω was simply added to the network (preliminary measured value not presented).

Response to TENS in the forearm

The electrical responses of each layer in the forearm were analyzed to square pulse TENS Using the calculated network. A voltage pulse (or current) to drive stimulation was used: square wave of intensity 1 V (or 1 mA) / pulse width 3 ms / stimulation frequency 10 Hz / ramping time 1 μ s. Since the pulse signals have multiple frequency components, the electrical responses, such as the current flowing through the network, were calculated in the frequency domain. The response analysis was completed in the time domain after the inverse FFT. In addition, the change in total current was investigated by the changing pulse width (0.05 to 5 ms) and stimulation frequency (1 to 300 Hz).

Results

ECM developed with TPB approach

The 2D-to-3D expansion of PC method showed its reliability by verifying the GF terms are dominantly affected by tissue layer thickness, the geometrical tissue property (Fig. S2). The regression of the coefficients in the expanded PC equations to the tissue properties fitted to the 3D simulation data (Fig. 3). The determined coefficients for equation (5) and (6) were: $(a_0, a_1, a_2) = 10^{-3}(4.69, 1.12, 4.40)$ for PPC; $(b_0, b_1, c_0) = 10^3(2.83, 1.00, 1.30)$ for SPC. Consequently, the TPB model was developed by the analytical PC principles complemented by the computational regression of IFs.

Figure 4 The forearm complex permittivity analysis, the impedance, and the effects of tissue property variance on the impedance. a) the complex permittivity for skin, fat, and muscle, b) layer-by-layer impedance and the total impedance, c) the total impedance change due to tissue thickness, d) the total impedance change due to tissue permittivity.

Figure 5 Electrical responses of the forearm impedance network to voltage-driven pulse stimulation and expected current change according to pulse width (W) and stimulation frequency (F) of the voltage-driven stimulation input. a) voltage-driven pulse signal (intensity 1 [V]/ pulse width 3 [ms]/ stimulation frequency 10 [Hz]), b) current response to the stimulation, c) loaded voltage on skin layer, d) loaded voltage on fat layer, e) loaded voltage on muscle layer, f) the current change versus the stimulation frequency change (W=3 [ms]), g) the current change versus the pulse width change (F=10 [Hz]), h) total map of the current change.

TPB-ECM agreeing to in vitro experiments

The modeled capacitances in ECM constructed using the TPB approach well-agreed to the in vitro results. The capacitances of double-layered agar phantoms were calculated by the developed model using the measured permittivity of Agar_{Salt} and Agar_{Pure} (Fig. 2b). The magnitude of permittivities verified that the PPC phantom was Agar_{Salt}-on-Agar_{Pure}, and the SPC phantom was Agar_{Pure}-on-Agar_{Salt} due to the permittivity increase with the injected NaCl ion in Agar_{Salt}. For both PPC and SPC phantoms, the real part of measured capacitances agreed well with the modeled values, but errors were present in the low-frequency components of imaginary parts (Fig. 2c and 2d). Except for the low-frequency imaginary error, there were the overall tendency matches between the model and the experiment.

Forearm impedance and response to TENS analyzed with TPB-ECM

We applied the TPB-ECM to analyze the multilayered forearm structure. The impedance network of the forearm and its electrical responses to square-pulse TENS were calculated. Permittivities of forearm tissue structure satisfied the SPC condition ($|\epsilon_{muscle}| > |\epsilon_{fat}| > |\epsilon_{skin}|$) (Fig. 4a). Compared to the skin impedance (≈ 30 k Ω), the impedance magnitude of fat (≈ 1 k Ω and muscle (≈ 0.1 k Ω layers were relatively small. Thus, the total impedance followed the skin impedance tendency as expected from the references of empirical measurements [28] (Fig. 4b). In addition, the analysis of permittivities with varying tissue layer thicknesses clearly showed that the skin thickness was the dominant factor for total impedance compared to other variables. The skin thickness changed the total impedance from 30 to 80 k Ω , whereas the others did not induce significant change (Fig. 4c and 4d).

The square-wave voltage pulse applied to the impedance network induced the tiny current, and the current was distorted when penetrating through the serially connected complex capacitor network (Fig. 5a and 5b). The shape and intensity of voltage loaded

on the skin were similar to the source, whereas the voltages loaded on the fat and muscle were very small and shaped similarly to the total current (Fig. 5c-5e). The variance in pulse width changed the amount of total current remarkably, but the stimulation frequency did not induce significant change (Fig. 5f-5h). Thus, even though the large drop of the supplied voltage near the skin resulted in tiny current flowing into the body, it was possible to increase the current penetrating into the body dominantly by adjusting the pulse width. Similarly, analyses of the forearm responses to the current-driven TENS showed that high voltage (≈ 40 V) was required to flow the unit current inside the body (1 mA). However, the current source could also avoid the effect of high skin impedance and thus reduce the high voltage load at the source by decreasing the stimulation frequency and increasing the pulse width (Fig. S1).

Discussion

The ECM is very useful in modeling the impedance of body and in analyzing the response to TENS. It calculates the comprehensive response to multi-frequency components of the pulsed-TENS better than other methods such as the numerical volume conductor approach. However, the current ECMs use meta-parameters for modeling the body impedance network instead of the physical properties of tissue layers. This indirect parameterization often leads to the problem of non-unified parameter sets and correspondingly hinders the development of a generally applicable TENS analysis method.

In this study, we designed a new approach for modeling the body impedance network, named as the TPB, to directly parameterize the physical properties into the ECM. The TPB approach employs the hybrid technique and incorporates both analytic and computational methods to overcome the difficulty of projecting 3D electromagnetic field distribution into 1D equivalent circuit. The TPB models effects of both electrical and geometrical properties. The analytic PC method can even address the interaction of complex permittivity between neighboring layers and the connection

between their complex capacitors. The 3D FEM simulation complements the PC method equations by parameterizing the nonlinear layer-by-layer geometrical properties.

The developed TPB-ECM was validated through in vitro experiments on phantoms. The real and imaginary components of the measured complex capacitances in phantoms were compared against the modeled values. In other words, the model was verified by evaluating both the capacitive and resistive components of the phantom's impedance. The results showed that the model generally agreed well with the measurement. At low frequency, there were underestimations in imaginary parts, which implies overestimation in the modeled conductivity. This error is most likely due to electrode-skin impedance being high at low frequency [29,30], which was not considered in this study.

Finally, to evaluate the model's usefulness in-vivo, we applied the TPB-ECM to multilayered forearm structure and compared them to well-known reference data, which have been derived empirically. The skin impedance was similar to the RC low pass filter and was the dominant factor in the total impedance. Among the physical parameters, the skin thickness had more influence on impedance than other parameters. These findings agree well with the known experimental references [31, 32]. Similar to the in vitro error, the modeled low-frequency impedances were relatively lower than the reference, mainly due to the electrode-skin interfacing impedance. By adding the conventional electrolyte-skin impedance model, the modeled impedance of TPB-ECM can be made to have high accuracy in both low frequency as well as high frequency. Using the network of variable impedances in forearm, the forearm responses were analyzed using the conventional square pulse TENS. The applied source voltage was largely consumed at the skin surface. By shortening pulse width, the voltage drop at the skin surface can be mitigated, which is consistent with the reported experimental results [19,33,34]. These results validate the TPB-ECM method can model the layered body structure more accurately and efficiently than the current ECM and FEM methods.

Conclusion

The TPB-ECM approach proposed here to analyze the response TENS provides a method for direct physical property parameterization, including the electrical interaction between adjacent layers of body structures and the nonlinear geometrical effects of each layer. Compared to the conventional meta-parameter fitting, TPB-ECM avoids the risk of non-unification and thus can be applied widely to analyze the response to TENS. The TPB-ECM results agree well with the

experimental results in in vitro experiments and with in-vivo results derived empirically. We believe, TPB-ECM has the potential to become a commonly used platform to analyze the response to TENS and could provide impactful guidance in the standardization of TENS parameter design.

Abbreviations

TENS: Transcutaneous electrical nerve stimulation; ECM: Equivalent circuit model; TPB: Tissue-property base; PC: Partial capacitance; GF: Geometric factor; PPC: Parallel partial capacitance; SPC: Serial partial capacitance; IF: Interdimensional factor;

Declarations

Ethics approval and consent to participate

Not applicable.

Consent for publication

Not applicable.

Availability of data and materials

The datasets used and/or analysed during the current study are available from the corresponding author on reasonable request.

Competing interests

The authors declare that they have no competing interests.

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Authors' contributions

All authors contributed to the study concept and design. The in silico and in vitro experiments, data analysis, writing the draft of manuscript were performed by J.Lee. Critical revision of the manuscript was performed by S-M. Park. All authors read and approved the final manuscript.

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Additional Files

Additional file 1 — Supplementary script

The supplementary script describes the in-silico experiments verifying the expansion of the the original 2D PC principles into 3D condition. The results show the indicator dependent on geometrical property in 2D is also dependent on the geometrical property in 3D (Fig. S2).

Additional file 2 — Figure S1

Electrical responses of the forearm impedance network to current-driven pulse stimulation. Using the developed TPB-ECM, the forearm impedance network and electrical responses to current-driven TENS were analyzed.

Additional file 3 — Figure S2

GF indicator calculated from in silico experiments using FEM simulator. a) PPC GF indicator, b) SPC GF indicator.

Table 1. Relative permittivity ($\varepsilon_1, \varepsilon_2$) pairs for 3D simulation.

For PPC ($\varepsilon_1 > \varepsilon_2$)		ε_2			
		10	20	30	40
ε_1	20	(20, 10)			
	30	(30, 10)	(30, 20)		
	40	(40,10)	(40, 20)	(40, 30)	
	50	(50, 10)	(50, 20)	(50, 30)	(50, 40)
For SPC ($\varepsilon_1 < \varepsilon_2$)		ε_2			
		20	30	40	50
ε_1	10	(10, 20)	(10, 30)	(10, 40)	(10, 50)
	20		(20, 30)	(20, 40)	(20, 50)
	30			(30, 40)	(30, 50)
	40				(40, 50)

Figures

Figure 1

Principle of PC method. a) Parallel partial capacitance (PPC), b) Serial partial capacitance (SPC).

Figure 2

Complex permittivity measurement of salted and pure agar material and comparison of phantom capacitances. a) magnitude of the complex permittivities, b) phase of the complex permittivities, c) PPC phantom capacitance comparison, and d) SPC phantom capacitance comparison. (mean ± 2 std for b, c, and d)

Figure 3

Regression for interdimensional factor (IF) for importing the geometrical tissue properties into PC method formulas.

Figure 4

The forearm complex permittivity analysis, the impedance, and the effects of tissue property variance on the impedance. a) the complex permittivity for skin, fat, and muscle, b) layer-by-layer impedance and the total impedance, c) the total impedance change due to tissue thickness, d) the total impedance change due to tissue permittivity.

Figure 5

Electrical responses of the forearm impedance network to voltage-driven pulse stimulation and expected current change according to pulse width (W) and stimulation frequency (F) of the voltage-driven stimulation input. a) voltage-driven pulse signal (intensity 1 V/ pulse width 3 ms/ stimulation frequency 10 Hz), b) current response to the stimulation, c) loaded voltage on skin layer, d) loaded voltage on fat layer, e) loaded voltage on muscle layer, f) the current change versus the stimulation

frequency change ($W=3$ [ms]), g) the current change versus the pulse width change ($F=10$ [Hz]), h) total map of the current change.

Supplementary Files

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