Simulating, Modeling, and Sensing Variable Tissues for Wireless Implantable Medical Devices

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Abstract—Wirelessly-powered implantable medical devices require efficient power transfer through biological tissue within safety constraints on energy absorption, often in the presence of environmental variability. Accurate modeling of the tissue medium is essential to evaluate the performance and sensitivity of transcutaneous powering systems. Here, we investigate loop and dipole antenna topologies in proximity to simulated tissue models and experimental phantoms, with emphasis on representing heterogeneous tissue with functionally-equivalent simplified models, and modeling variability in tissue properties for sensitivity analyses. We first present a modified phantom formulation that provides greater control over frequency-dependent properties. We then show that homogeneous phantoms have limited use at representing input impedance and energy absorption at ultrahigh operating frequencies (UHF) by analyzing each antenna topology in proximity to layered or homogeneous tissue across frequency. We compare loop and dipole antenna topologies in terms of specific absorption rate (SAR) and impedance, and show that frequency-dependent tissue behavior must be considered even at fixed operating frequencies. Finally, we discuss the dual utility of a transmitting antenna as a resonator to detect changes in tissue properties in addition to powering an implanted device.

Index Terms—Implantable medical devices, tissue dielectric properties, tissue phantoms, wireless power transfer

I. INTRODUCTION

WIRELESS transcutaneous electromagnetic systems must reliably and safely transfer power through a dynamic tissue medium [1]. Electromagnetically, tissue is a lossy dielectric with electromagnetic behavior and power transmission mechanisms described by permittivity and conductivity [2], [3]. Tissue dielectric properties are functions of smaller-scale tissue structure: conductivity describes charge conduction within a material in response to an applied field, where ions act as charge carriers and produce current; permittivity describes polarization within a material in response to an applied field, where cell membranes and other interfaces lead to charge buildup and capacitance, or at higher frequencies (>100 MHz) where polar molecules (such as water) align with an applied field [2], [4]–[7]. Electronic and atomic polarization occurs at even higher frequencies [8]. Mathematically, permittivity can be represented as a complex quantity to include effects on field amplitude and losses associated with polarization mechanisms. However, losses due to polarization and ionic mobility are often combined into a single loss term.

There are many reports of measured tissue properties, either as conductivity and permittivity or as resistance and phase angle (capacitance) [9]. These measurements are typically over a range of frequencies to examine resonance and dispersion behavior. Gabriel et al. performed a comprehensive review of measurements and developed an empirical model that has since been widely used for nominal property values in the wireless electromagnetic powering literature [7], [10], [11]. From the reported measurements of tissues, it is clear that tissue properties vary among samples even at a single frequency due to differences in structure and composition [7], [10], [12]–[14]. Gabriel et al. [7] estimated a variation of 5-10% above 100 MHz, but Paulides et al. [15] note a variation of 30% or more in breast and brain tissues, and Balidemaj et al. [14] found 14% higher conductivity in their in vivo measurements of muscle tissue. Likewise, Halter et al. [13] reported higher conductivity and permittivity in vivo than those measured in ex vivo samples, in a study motivated by the difficulty of estimating dielectric properties on an individual basis. Therefore, designing a system for nominal tissue properties using the empirical model is a starting point, but additional consideration of variability is necessary for a comprehensive evaluation of the potential impacts on system performance and safety.

With the exception of measurements relative to frequency or temperature, there are relatively few reports quantifying tissue structure and composition as they correspond to measured dielectric properties [4]–[6], [11]. Finding literature relating tissue property measurements and tissue parameter values is difficult presumably due to the difficulty of in vivo measurements of human tissue properties [13], and alternately the difficulty of quantifying tissue parameters without breaking down a tissue sample into its material constituents.

General statements can be made about the relationship between dielectric properties and parameters such as water content (e.g., bone, fat, muscle, skin) or cellular structure (e.g., gray matter versus white matter, blood versus tissues) [2], [6], [7]. One of the main resources for such relationships is the bioimpedance literature, where tissue properties are used as indicators of tissue parameters such as water content [12], [16]. Schwan et al. [12] discussed a model to estimate protein-bound water from constituent dielectric properties. Schepps and Foster [4] used the Fricke mixture model and Debye functions to fit measurements of dielectric properties versus frequency for tissues of varying water content (tumor, liver, spleen, fat, and muscle).

The representation of small-scale tissue structure with bulk dielectric properties is integral to modeling heterogeneous tissue with simplified structures, using discrete layers or
TABLE I
INGREDIENT RATIOS OF EACH PHANTOM, REPRESENTING TWO FORMULATIONS FOR EACH OIL RATIO.

<table>
<thead>
<tr>
<th>Oil Ratio [17]</th>
<th>Water [mL]</th>
<th>Gelatin [g]</th>
<th>Oil [mL]</th>
<th>Detergent [mL]</th>
</tr>
</thead>
<tbody>
<tr>
<td>10.00%</td>
<td>85.50</td>
<td>15.30</td>
<td>10.00</td>
<td>0.56 / 5.60</td>
</tr>
<tr>
<td>15.00%</td>
<td>80.75</td>
<td>14.45</td>
<td>15.00</td>
<td>0.84 / 8.40</td>
</tr>
<tr>
<td>20.00%</td>
<td>76.00</td>
<td>13.60</td>
<td>20.00</td>
<td>1.12 / 11.20</td>
</tr>
<tr>
<td>25.00%</td>
<td>71.25</td>
<td>12.75</td>
<td>25.00</td>
<td>1.40 / 14.00</td>
</tr>
<tr>
<td>30.00%</td>
<td>66.50</td>
<td>11.90</td>
<td>30.00</td>
<td>1.68 / 16.80</td>
</tr>
<tr>
<td>50.00%</td>
<td>47.50</td>
<td>8.50</td>
<td>50.00</td>
<td>2.80 / 28.00</td>
</tr>
<tr>
<td>70.00%</td>
<td>28.50</td>
<td>5.10</td>
<td>70.00</td>
<td>3.92 / 39.20</td>
</tr>
<tr>
<td>75.00%</td>
<td>23.75</td>
<td>4.25</td>
<td>75.00</td>
<td>4.20 / 42.00</td>
</tr>
<tr>
<td>80.00%</td>
<td>19.00</td>
<td>3.40</td>
<td>80.00</td>
<td>4.48 / 44.80</td>
</tr>
</tbody>
</table>

homogeneous models. Mixture theory defines “apparent” (or “effective”) properties of heterogeneous materials based on their components [16], [18], [19]. In the context of wireless transcutaneous power transfer, the operating frequency determines the scale of interactions that can be simplified with the use of tissue layers and apparent properties, and whether homogeneous models can accurately model electromagnetic exposure metrics such as specific absorption rate (SAR) [20], [21].

Calculating and modeling apparent properties are applicable to tissue models in general, and particularly experimental validation of transcutaneous power transfer using tissue phantoms, such as those developed by Lazebnik et al. [17], [22], [23]. Although Lazebnik et al. and Porter et al. present measurements of their phantoms across frequency, and Lazebnik et al. compare the phantom properties to the empirical model by Gabriel et al. [11], these previous works have not addressed apparent dielectric properties using layered phantoms.

In this work, we first represent the relationship between tissue dielectric properties and smaller-scale structure utilizing tissue phantoms, measuring complex dielectric parameters over a range of frequencies. We then investigate apparent properties of layered phantoms with varying layer thicknesses in simulation, using measured phantom properties and two antenna topologies: a square single-turn loop and a meandered dipole. Finally, we propose a method of sensing changes in the tissue medium using the same antenna that would be used to transmit power to a miniature implanted device. The overall goals of this work are to examine tissue models in the context of tissue variability, and to show how robustness to variability can be integrated into transcutaneous system design.

The main contributions of this work are as follows:

- Modified tissue phantom formulations that provide more control over frequency-dependent dielectric properties
- A comparison of dipole and loop antenna UHF field distributions and SAR in proximity to homogeneous and heterogeneous tissues
- A demonstration of the importance of using wideband phantoms when analyzing antenna impedance
- A discussion of wideband and narrowband antenna characteristics with regard to changes in tissue properties (an extension of the wide- or narrowband concept with regard to frequency)

![Fig. 1. Summary of methodology in the current study.](#)

**II. METHODS**

This study was conducted in three parts, as summarized in Figure 1: first, we fabricated phantoms of various formulations and measured dielectric properties over frequency for each formulation; next, we used the measured phantom properties in simulations of layered and homogeneous tissues and simulated loop and dipole antennas in proximity to variable tissue, examining antenna impedance and SAR; finally, we measured differential input impedance of loop and dipole antennas in proximity to phantoms of various properties, to investigate the usefulness of the antennas as resonators for detecting changes in tissue properties.

Loop and dipole antennas were chosen for this investigation because their field patterns are associated with different dominant power transfer mechanisms. The results presented in this work can be generalized to other printed antenna topologies with regard to these field patterns and dimensions. More detail on power transfer mechanisms for the loop and dipole antenna topologies used here is provided in prior work [25].
A. Phantom Formulations and Properties

Gelatinous phantoms with various dielectric properties were fabricated using simplified formulations based on work by Lazebnik et al. [17], [22], [23]. The formulations are shown in Table I for different ratios of oil/gelatin mixture, where the amount of detergent was also varied to examine whether this provided further control over the phantom properties.

After formulating according to the procedure in [17] and allowing the phantoms to solidify for 24 hours, the dielectric properties of each phantom were measured with a dielectric probe (SPEAG DAK 3.5) across the frequency range of 400 MHz to 2 GHz to examine the UHF range for mid-field wireless powering [24].

The phantoms were fabricated in cylindrical “puck” shapes, of sufficient thickness to appear infinite to the dielectric probe, verified by measuring the dielectric properties with and without a short placed behind the phantom and verifying that the measured properties were the same [8].

B. Layered and Homogeneous Tissue Models

The measured phantom properties were used to investigate apparent dielectric properties in simulation, by comparing layered and homogeneous tissue models. Antennas and tissue models of similar geometry to the measured phantoms were simulated in ANSYS HFSS (ANSYS Electronics Desktop 2017.0). Homogeneous and layered tissue was modeled with dielectric properties of each tissue layer defined in terms of conductivity and permittivity. A dipole or loop transmit antenna was placed in contact with the external tissue surface, with antenna dimensions optimized for power gain, as described in prior work [25]. A 10 µm layer of parylene was included in the simulation to encapsulate each antenna [26], [27]. An illustration of the simulated antennas and tissue models is shown in Figure 2.

The dielectric properties of the homogeneous models and each layer of the layered tissue model were defined using measured properties of the phantoms most closely matching skin, fat, and muscle tissues, and in the layered model the thickness of skin and fat layers were varied according to values reported in [28], as listed in Table II. The tissue thicknesses were varied independently over all possible combinations. The overall thickness of the model was kept constant at 25 mm by setting the appropriate muscle layer thickness behind skin and fat, to match the fabricated phantom thicknesses.

The layered and homogeneous models were compared in terms of maximum 1-g average SAR [29] and transmit antenna input impedance [30], to investigate whether layering phantoms is a feasible way of verifying antenna system performance with variable tissue properties.

C. Antenna Impedance for Detecting Variable Properties

To investigate a method of sensing changes in dielectric properties, the input impedance of fabricated transmit antennas was measured using a vector network analyzer (Agilent 8753ES S-parameter VNA) and the fabricated phantoms of various formulations. A differential probe was used to measure differential input impedance of fabricated single-turn loop and meandered dipole antennas when in proximity to each phantom. The change in input impedance was examined as a possible method of sensing changes in properties, similar to using a dielectric resonator [30].

III. RESULTS

A. Phantom Formulations and Properties

Measured phantom dielectric properties are shown in Figure 3, alongside Cole-Cole empirical model values from [11], [31]. The conductivity here represents a combined loss term due to ionic mobility and polarization, and is therefore frequency dependent. Higher percent oil clearly decreases the phantom relative permittivity, and decreases the slope of conductivity versus frequency, consistent with lower polarization losses. The addition of more surfactant increases conductivity closer to the empirical model values for muscle and skin, while causing the permittivity and conductivity to vary over a smaller range across formulations.

Physically, the lower oil content phantoms were more solid than the higher oil content phantoms. The addition of more detergent makes the phantom consistency more gelatinous, such that the higher percentage oil and detergent phantoms did not hold their shape outside of a container. However, the high percentage oil and detergent phantom properties did not match any of the empirical model properties (being high conductivity but low permittivity); all of the phantoms matching the model tissue properties held their shape, making them suitable for fabricating layered phantoms.

The properties of phantoms that were closest to empirical model values for fat (80%, original formulation), muscle (15%, modified formulation), wet skin (30%, modified formulation), and dry skin (25%, modified formulation) were used in subsequent simulations of layered and homogeneous tissue.

B. Layered and Homogeneous Tissue Models

Comparisons of homogeneous and layered simulated tissue were first performed at a single frequency of 915 MHz, where the properties of each phantom (as used in the simulation models) are as listed in Table III. As previously indicated and consistent with Figure 3, permittivity decreases with greater percent oil, while conductivity is not proportional to percent oil due to the effect of the formulation on the slope of conductivity versus frequency.

Maximum 1-g average SAR at 915 MHz is shown in Figure 4 for homogeneous and layered simulation models. Fat thickness was varied over a wider range than skin thickness,
Fig. 2. Simulated antennas and tissue models used to investigate apparent dielectric properties: (A) single-turn loop antenna, (B) single-turn loop with layered tissue, (C) single-turn loop with homogeneous tissue, (D) meandered dipole antenna, (E) meandered dipole with layered tissue, and (F) meandered dipole with homogeneous tissue.

Fig. 3. Measured relative permittivity ($\varepsilon_r$), conductivity ($\sigma$), and loss tangent ($\tan \delta$) of each phantom formulation compared to Cole-Cole model values for fat, muscle, and skin from [11], [31]. Phantom formulations are labeled with the oil ratio [17], where an asterisk indicates the modified formulation with more detergent.

so the four clusters of points in each of the plots represent increasing fat thickness, and the four points within the cluster represent increasing skin thickness. SAR for the loop tends to increase with skin thickness and is relatively independent of fat thickness. SAR for the dipole is more dependent on fat thickness, decreasing with greater fat thickness and increasing
TABLE III
DIELECTRIC PROPERTIES OF PHANTOMS CLOSEST TO MUSCLE, DRY SKIN, WET SKIN, AND FAT AT 915 MHZ

<table>
<thead>
<tr>
<th>Oil Ratio</th>
<th>Detergent [mL]</th>
<th>Relative Permittivity</th>
<th>Conductivity [S/m]</th>
<th>Tissue Match</th>
</tr>
</thead>
<tbody>
<tr>
<td>15% *</td>
<td>8.40</td>
<td>53.83</td>
<td>0.701</td>
<td>Muscle</td>
</tr>
<tr>
<td>25% *</td>
<td>14.00</td>
<td>46.25</td>
<td>0.741</td>
<td>Skin - Dry</td>
</tr>
<tr>
<td>30% *</td>
<td>16.80</td>
<td>44.03</td>
<td>0.775</td>
<td>Skin - Wet</td>
</tr>
<tr>
<td>80%</td>
<td>4.48</td>
<td>8.84</td>
<td>0.147</td>
<td>Fat</td>
</tr>
</tbody>
</table>
* Modified formulation from [17] with more detergent

Fig. 4. SAR of the loop and dipole antennas in proximity to layered models and homogeneous models with properties of phantoms closest to skin, fat, and muscle at 915 MHz.

Fig. 5. Impedance of the loop and dipole antennas in proximity to layered models and homogeneous models with properties of phantoms closest to skin, fat, and muscle at 915 MHz.

The input impedances of the dipole and loop antennas in proximity to each layered model and homogeneous model at 915 MHz are shown in Figure 5. For the loop at this frequency, the impedance values of the layered model fall between the homogeneous models of fat and wet skin for both antennas. Lighter marker color for the layered models corresponds to greater skin thickness, therefore it appears that greater skin thickness increases real and imaginary impedance of the loop.

For the dipole, similar to the layered model SAR, the layered model impedances are clustered by fat thickness and impedance tends to increase with skin thickness. The impedance values of the dipole for the layered models do not appear to fall along the same trend as the homogeneous properties.

The trend behind the dipole layered model impedances becomes apparent when impedance is viewed across frequency, as in Figure 6a. Proximity to different tissue configurations shifts the frequency profile of the antenna impedance, as expected for changes in media near an antenna. Overall, the profiles of the layered models fall between the homogeneous fat model and the homogeneous models of the other tissues. The small size of the loop antenna causes the peak in impedance to fall beyond the highest frequency included in these simulations, but the behavior can be extrapolated from the simulated frequency range and the expected antenna behavior. Note that the loop input impedance is primarily inductive, while the dipole input impedance is mostly capacitive.

Because the trend of impedance with frequency is more informative than values at a single frequency, maximum SAR was also examined at two frequencies to investigate frequency-dependent behavior. Maximum 1-g average SAR values for the layered and homogeneous models are shown in Figure 6b at 915 MHz and 2 GHz. The layered models show good agreement with the homogeneous models of skin and muscle for the loop system, but for the dipole system SAR increases by a greater amount with frequency for the layered model than for the homogeneous models. At 2 GHz, the dipole layered model SAR is higher than that predicted by the homogeneous models.
Heatmaps of the spatial SAR distribution for each antenna and frequency are shown in Figure 7, for the layered model with nominal tissue thicknesses and the homogeneous model representing wet skin tissue. The wet skin homogeneous model was chosen because it has the highest conductivity of the homogeneous models, to compare to prior work that simplified heterogeneous structure to homogeneous models with high loss parameters to conservatively estimate SAR [32].

C. Antenna Impedance for Detecting Variable Properties

Measured differential input impedance is shown in Figure 8. The fabricated antennas were measured first with the antenna traces in contact with the phantom, and again with the substrate in contact with the phantom. Encapsulation has been shown to improve power transfer and decrease SAR, and the substrate-contact condition was intended to mimic a layer of encapsulant protecting the metal traces of the antenna from contact with lossy tissue [26], [33]. It was found that the antenna-contact configuration impedance changed more with the phantom properties, likely due to the combined effects of the antenna being closer to the tissue and the contact with lossy tissue. The loop antenna (with contact) showed smaller variation than the dipole for phantom oil ratios less than 50%, but greater impedance variations for oil ratios of greater than 50%. This is indicative of the loop antenna being more narrowband than the dipole antenna.

IV. DISCUSSION

Simulated and experimental tissue models are necessary for evaluating wireless electromagnetic powering of implantable medical devices, where antenna systems must operate reliably in a dynamic tissue environment. Biological tissue structure is complex, and the use of simplified tissue models can facilitate the design and evaluation of transcutaneous powering systems. However, the simplified tissue models must still provide an accurate representation of the environment, especially SAR and power efficiency. Appropriate choice and application of simplified models must be informed by knowledge of multiple fields, including biological dielectric material properties, electromagnetic dosimetry, and antenna theory.

In this work, apparent dielectric properties were investigated as a representation of how bulk properties change as a function of smaller-scale structure, and how various apparent tissue properties can be mimicked with different phantom formulations and layered phantoms [34]. Phantoms were fabricated and measured dielectric properties were then used in simulations of layered and homogeneous tissue.

The measured phantom properties were similar to those reported by Lazebnik et al. [17]. The slope of the conductivity is indicative of the water content, because this conductivity is calculated from imaginary permittivity. Losses increase to a greater extent with frequency for higher water content phantoms (steeper slope in conductivity versus frequency). The addition of more detergent increases phantom conductivity and compresses the range of relative permittivity across varying
While phantom permittivity can be matched to empirical model properties of muscle and skin, the phantom conductivity varies over a smaller range than actual tissue, and even the 80% oil phantom does reach the small model values of fat permittivity and conductivity. This contributes to a higher loss tangent for the fat phantom than Gabriel’s Cole-Cole model for fat [11], [31]. However, fat tissue has been shown to vary in dielectric properties among measured samples [7]. Perhaps more importantly, the controllable phantom formulation allows representation of various tissue properties that can be used to evaluate the sensitivity of a system to changes in tissue properties.

Maximum average SAR has been used in prior work to compare homogeneous and heterogeneous tissue models [29]. Homogeneous tissue with high dielectric properties has been used to obtain a conservative estimate of SAR [32], but it has also been suggested that SAR does not necessarily scale with higher dielectric properties [21], [33]. The primary concern with modeling heterogeneous tissue as homogeneous is the potential misrepresentation of SAR due to the lack of small-scale tissue structure [36].

In this work, the maximum SAR trend with frequency for the dipole antenna differed between the homogeneous and layered models: maximum SAR for the dipole was overestimated by homogeneous models at 915 MHz, but underestimated by homogeneous models at 2 GHz. The dipole antenna was larger than the loop, therefore the dipole system was presumably more dependent on deeper tissue and layer interfaces including fat and muscle. However, even though the maximum loop SAR corresponded to homogeneous models across frequencies, the SAR distribution was noticeably affected by the presence of layered tissue structure for both antennas. This suggests that maximum SAR may not be sufficient for quantifying the effects of tissue structure. Examining the results of SAR across frequencies in this work indicates that the smaller-scale tissue structure is essential for estimating SAR distribution. The layered model simulated in this work assumed homogeneity within tissue layers, but even this simplification will not be valid at high enough frequencies.

The dipole antenna showed generally higher values of maximum SAR than the loop antenna at 915 MHz, particularly for homogeneous models. However, at 2 GHz, the loop antenna showed higher maximum SAR than the dipole for all models except homogeneous fat. This further emphasizes the importance of analyzing field distributions and antenna size with specific regard to tissue structure in implantable applications, where the effect of tissue proximity and SAR distributions must complement the intuitions suggested by traditional antenna theory.

The results of the impedance analysis in this work show that a homogeneous model designed to match input impedance of heterogeneous tissue at a given frequency may not accurately represent the full frequency profile. Therefore, a simplified homogeneous tissue that has been verified by comparing impedance to a heterogeneous model is limited in utility to a specific tissue geometry, frequency, and antenna dimensions. While this may not seem to be an issue if the analysis is performed at a fixed frequency, the frequency behavior of tissue is directly related to changes in impedance due to environmental parameters at a fixed frequency. This is illustrated by the results of the current work, comparing changes in dipole impedance for tissue models at 915 MHz versus the impedance over a range of frequencies. This result emphasizes the importance of characterizing tissue models across a range of frequencies to ensure that they accurately model tissue behavior.

The choice of tissue models with realistic frequency profiles is still limited, as the properties of one model are not representative of the variability that will be encountered among individuals and different tissue structures. Evaluation of transcutaneous systems using variable tissue models is a more comprehensive approach to ensuring that the function of a system is robust to changes in tissue properties. The variable phantom formulations used in this work provide controllable dielectric properties, and the layered tissue models simulated in this work demonstrate additional structural representations that could be replicated with layered phantoms. Multiple models could then be used to simulate variable tissue and evaluate a system’s behavior in each case.

The frequency-dependent impedance profile of an antenna enables use of the antenna as a resonator for detecting changes in surrounding media [30]. Here, we focused on impedance variations due to properties of the tissue medium, however, variations in antenna impedance can also occur due to antenna positioning and other environmental changes. Impedance variations indicating environmental changes provide informative feedback in adaptive powering systems [1]. While robustness to environmental changes is desirable in a static system, in an adaptive system changes in the system parameters are not undesirable, provided they can be detected and compensated by tuning properties such as input power, impedance, or frequency. The results of this work are therefore generalizable to adaptive transcutaneous powering, using the transmit antenna as a resonator and impedance variations as feedback for an adaptive tuning scheme.

The sensing capability of an antenna used as a resonator is higher when in closer proximity to the sensed media. In the context of this work, this means moving the antenna closer or in contact with the tissue surface. However, contact with lossy tissue may then decrease the effectiveness of the antenna for power transfer [33]. Therefore, if using the same antenna as a sensing resonator and as a transmit antenna for wireless powering, using the antenna substrate as a protective dielectric layer on one side of the transmit antenna can provide both sensing capability and power efficiency. Placing the antenna traces in contact with the tissue provides greater sensing capability, whereas placing the dielectric substrate layer between the antenna and the tissue improves powering efficiency. This technique is valid for symmetric antennas such as the loop or dipole presented in this work.

Because the antennas in this work were optimized for power transfer, different antenna dimensions may improve the utility of the antennas for sensing. Other antenna dimensions or topologies could be investigated in future work, including patch antennas [37], or complex patterned transmit antennas...
Fig. 7. Simulated SAR of the loop and dipole antennas in proximity to layered and homogeneous tissue models at 915 MHz and 2 GHz.

Fig. 8. Measured differential input impedance of the loop and dipole antennas in proximity to phantoms of various formulations. Each antenna was either placed with the printed circuit board substrate or the antenna in contact with the phantom.
[38]. Each of these topologies is expected to have different SAR behavior and frequency profile, but the work presented here for a loop and dipole antenna can be generalized based on expected antenna field patterns and power transfer mechanisms.

In general, narrowband antennas such as the loop will provide large changes in impedance over a small range, while wideband antennas will provide smaller changes in impedance over a wider range. Here we extend the concept of wide- or narrowband with regard to frequency to wide- or narrowband with regard to tissue properties. As shown in the results of this work, changes in tissue properties affect the antenna impedance at a given frequency because they shift the frequency profile of the antenna - a phenomenon directly related to changes in antenna impedance with operating frequency.

With regard to sensing changes in tissue properties, the use of separate antennas for sensing and powering may be necessary, in order to design for the optimal properties in either case. Additionally, the properties of the substrate itself could be tuned to change the impedance profile of the antenna. A thicker substrate will result in a more stable impedance, but even a thinner substrate layer will provide a more controlled spacing between the antenna and the tissue, preserving the ability to detect changes in the tissue properties while ameliorating the more unpredictable effects of imperfect antenna contact with the tissue. The permittivity of the substrate is another variable, and could be further investigated for improving the antenna’s use as both a sensing resonator and power transmitter.

V. CONCLUSION

Tissue models are essential for evaluating the performance and safety of wireless transcutaneous powering systems. Simplified tissue models can represent the complex structure of biological tissue, but the structural simplification must not misrepresent the characteristics of the tissue medium. In this work, we investigated loop and dipole antenna topologies in proximity to tissue models (both simulated and experimental), and compared heterogeneous and homogeneous tissue models in terms of input impedance and SAR. The results indicate the importance of accurately representing frequency-dependent characteristics of tissue even when performing evaluations at a single frequency, in order to accurately model changes in impedance. As frequency increases, heterogeneous tissue structures become more important to SAR calculations, and using a representative worst-case set of properties in a homogeneous model will not accurately reflect the SAR distribution in heterogeneous tissue. The modified phantom formulations presented in this work are well-suited to representing heterogeneous tissue structure, with permittivity and conductivity of tissue layers representative of frequency-dependent tissue properties. Additionally, the same antenna can be used to transmit power and to detect changes in tissue properties via input impedance, with attention to wide- or narrowband characteristics.

The results of this study suggest that tissue models used to evaluate transcutaneous systems should be evaluated and designed across a wide frequency range, before choosing multiple tissue models to evaluate the robustness of a system to tissue variability, and potentially incorporating the sensed impedance changes of the antenna as feedback for adaptive tuning of transcutaneous powering.

REFERENCES


