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Ultrawideband temperature-dependent dielectric properties of animal liver tissue in the microwave frequency range

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Abstract
The development of ultrawideband (UWB) microwave diagnostic and therapeutic technologies, such as UWB microwave breast cancer detection and hyperthermia treatment, is facilitated by accurate knowledge of the temperature- and frequency-dependent dielectric properties of biological tissues. To this end, we characterize the temperature-dependent dielectric properties of a representative tissue type—animal liver—from 0.5 to 20 GHz. Since discrete-frequency linear temperature coefficients are impractical and inappropriate for applications spanning wide frequency and temperature ranges, we propose a novel and compact data representation technique. A single-pole Cole–Cole model is used to fit the dielectric properties data as a function of frequency, and a second-order polynomial is used to fit the Cole–Cole parameters as a function of temperature. This approach permits rapid estimation of tissue dielectric properties at any temperature and frequency.

1. Introduction

The dielectric properties of biological tissues have been the subject of active research since the beginning of the 20th century. Fundamentally, the dielectric properties determine how electromagnetic fields will interact with and propagate within biological tissues. More practically, accurate knowledge of the dielectric properties of biological materials has a host of applications in the consumer, food/agricultural and biomedical industries. Some of these applications include characterizing and limiting human exposure to electromagnetic fields (Rosen et al. 2002), determining how food products will heat in a microwave oven (Tran...

Often, dielectric spectroscopy of animal and human tissues is performed *ex vivo* for reasons of convenience or due to difficulties in establishing *in vivo* experiment protocols. In these cases, measurements often occur at room temperature. However, even during *in vivo* experiments, most measurement modalities (e.g., open-ended coaxial probes) require direct contact with the material under test, and measurement temperature is often reduced from normal body temperature. In addition, certain treatment technologies such as hyperthermia and ablation take place at temperatures far greater than room temperature, and the corresponding change in the tissue dielectric properties must be taken into account.

A number of researchers measured the dielectric properties of biological tissues at discrete frequencies and varying temperatures and presented their results in terms of linear temperature coefficients, which are defined as per cent change in either permittivity or conductivity per degree Celsius (Schwan and Li 1953). Although linear temperature coefficients can be expected to vary with microwave frequency and tissue temperature, previous studies only provide values at a limited number of specific discrete frequencies and temperatures. In addition, to the best of our knowledge, the validity of linear temperature coefficients over wide temperature ranges has not been investigated. A brief summary of the temperature coefficient data currently in the literature is presented in table 1. In addition, a large body of research pertinent to the agricultural industry has been conducted on the temperature dependence of the dielectric properties of meats (see, e.g., Tran and Stuchly (1987), Ryynanen (1995), Bircan and Barringer (2002), Sipahioglu et al (2003)). This is relevant to predicting how meat will thaw, heat, and cook in a microwave oven. These studies were carried out at discrete frequencies between 915 and 2450 MHz, but temperature coefficients were not calculated.

Linear discrete-frequency temperature coefficients are impractical for ultrawideband (UWB) applications since temperature coefficients must be presented at every frequency and temperature range of interest, requiring the compilation of cumbersome look-up tables, as well as data interpolation at intermediate frequencies. In addition, it is not clear that linear temperature coefficients are appropriate at all temperatures and frequencies. Therefore, the goals of this study are (1) to systematically measure the continuous temperature dependence of the dielectric properties of a representative tissue type—bovine and porcine liver—over a broad frequency range (0.5–20 GHz), and (2) to present a novel and compact data representation technique. We propose and demonstrate a method to characterize the change in tissue dielectric properties with temperature over wide bandwidths. Namely, we use a single-pole Cole–Cole model to fit the frequency dependence of the dielectric properties, and a second-order polynomial to model the temperature dependence of the Cole–Cole parameters. Using this method, the dielectric properties of liver tissue at temperatures ranging from room temperature to ~60 °C, and frequencies spanning 0.5–20 GHz (which cover the ISM bands near 900 MHz, 2500 MHz and 5800 MHz) can be easily calculated.

The remainder of this paper is organized as follows. Section 2 discusses the experimental methods as well as the data reduction techniques. Section 3 discusses the validations of the precision probe performance at elevated temperatures and the Cole–Cole fitting procedure. Section 4 presents the dielectric spectroscopy as well as the data fitting results. In addition, we calculate temperature coefficients at discrete frequencies to demonstrate that our results are consistent with the data reported in the literature. Section 5 summarizes the results of this study. The details of the data reduction techniques are presented in the appendix.
### Table 1. Chronological summary of the previously published temperature-dependent dielectric properties data.

<table>
<thead>
<tr>
<th>Tissue type</th>
<th>Frequency range</th>
<th>Temperature coefficient (% °C⁻¹)</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Various tissues</td>
<td>50, 200, 400, 900 MHz</td>
<td>See reference for temperature coefficients</td>
<td>Schwan (1957)</td>
</tr>
<tr>
<td>Biological tissues</td>
<td>Microwave</td>
<td>$\frac{\Delta \sigma}{\sigma} = 2$, $\frac{\Delta \varepsilon}{\varepsilon} = -0.5$</td>
<td>Johnson and Guy (1972)</td>
</tr>
<tr>
<td>Brain</td>
<td>&lt;0.1 GHz</td>
<td>$\frac{\Delta \sigma}{\sigma} = 2$, $\frac{\Delta \varepsilon}{\varepsilon} = 2$</td>
<td>Foster et al (1979)</td>
</tr>
<tr>
<td></td>
<td>&lt;2 GHz</td>
<td>$\frac{\Delta \sigma}{\sigma} = 2$, $\frac{\Delta \varepsilon}{\varepsilon}$: small</td>
<td></td>
</tr>
<tr>
<td></td>
<td>7 GHz</td>
<td>$\frac{\Delta \sigma}{\sigma} = -1$, $\frac{\Delta \varepsilon}{\varepsilon}$: small</td>
<td></td>
</tr>
<tr>
<td>Biological tissues</td>
<td>Microwave</td>
<td>$\frac{\Delta \sigma}{\sigma}$: varies from 1 to 2</td>
<td>Schwan and Foster (1980)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\frac{\Delta \varepsilon}{\varepsilon}$: varies from −0.3 to 2</td>
<td></td>
</tr>
<tr>
<td>Biological tissues</td>
<td>Microwave</td>
<td>∼2</td>
<td>Burdette et al (1985)</td>
</tr>
<tr>
<td>Barnacle</td>
<td>0.1, 2.2, 9.5 GHz</td>
<td>$\frac{\Delta \sigma}{\sigma}$: 2 to −2, cross-over point near 2 GHz</td>
<td>Foster and Schwan (1989)</td>
</tr>
<tr>
<td>muscle fibres</td>
<td></td>
<td>$\frac{\Delta \varepsilon}{\varepsilon}$: small</td>
<td></td>
</tr>
<tr>
<td>Myocardium</td>
<td>0.2, 1.1, 2.5, 3.0, 4.0, 5.0, 6.0 GHz</td>
<td>See reference for temperature coefficients</td>
<td>Semenov et al (2000)</td>
</tr>
<tr>
<td>Bovine liver</td>
<td>915 MHz</td>
<td>$\frac{\Delta \sigma}{\sigma} = 1.82 \pm 0.28$, $\frac{\Delta \varepsilon}{\varepsilon} = -0.130 \pm 0.059$</td>
<td>Chin and Sherar (2001)</td>
</tr>
<tr>
<td>Animal and human liver</td>
<td>0.3-3 GHz</td>
<td>Pig: $\frac{\Delta \sigma}{\sigma} = 1.1$, $\frac{\Delta \varepsilon}{\varepsilon} = -0.17$</td>
<td>Staufler et al (2003)</td>
</tr>
<tr>
<td>Rat prostate</td>
<td>915 MHz</td>
<td>Bovine: $\frac{\Delta \sigma}{\sigma} = 2.0$, $\frac{\Delta \varepsilon}{\varepsilon} = -0.04$</td>
<td>Chin and Sherar (2004)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$\frac{\Delta \sigma}{\sigma} = 1.10 \pm 0.11$, $\frac{\Delta \varepsilon}{\varepsilon} = -0.31 \pm 0.05$</td>
<td></td>
</tr>
</tbody>
</table>

* Tissues not specified in the reference.
* Frequencies not specified in the reference.

## 2. Experimental methods

In this study we used 13 samples from 7 livers (6 bovine and 1 porcine). Fresh bovine liver was obtained from a local slaughter house, while fresh porcine liver was obtained from the Department of Animal Sciences at the University of Wisconsin–Madison. We chose liver tissue as a representative high water-content tissue (Reinoso et al 1997). Since previous studies have shown that there is little inter-species difference in the tissue dielectric properties (see, e.g., Gabriel et al (1996a)), we used bovine and porcine liver tissue for convenience and ease of accessibility.

In all cases, experiments were started within 2 h of excision. Two samples, measuring approximately 5 × 10 × 3 cm, were cut from the liver. Each sample was wrapped in aluminium foil leaving only the top surface exposed. Approximately 75% of the top exposed surface was covered by an acrylic plate with holes for the measurement probes. This allowed most of the tissue surface to be covered to prevent excessive dessication. The samples were then placed in...
Figure 1. Diagram of the experimental setup that demonstrates the measurement setup for two samples simultaneously.

Measurements of tissue temperature and dielectric properties over the frequency range 0.5–20 GHz were taken every 5 min during both a heating and cooling phase. During the heating phase, the oven was turned on and the temperature was continuously adjusted, allowing the tissue to heat slowly. Then, during the cooling cycle, the oven was turned off and the tissue was allowed to cool back to room temperature. We heated approximately half of the samples to about 37 °C (corresponding to body temperature), and the other half of the samples to about 60 °C. Temperatures greater than about 60 °C were not attempted due to significant physiological changes that were expected to take place in the tissue above these temperatures (Bircan and Barringer 2002). Tissue temperature in the sensing region of the dielectric probe was assumed to be consistent with and equal to the average of the two surrounding temperature probes.

The dielectric measurement and data processing procedure used in these experiments was previously described in Popovic et al. (2005), and will be explained briefly here. The complex reflection coefficient at the calibration plane of the open-ended coaxial probe was recorded. A de-embedding model, based on probe-specific numerical simulations that accounted for the specific probe geometry, was used to translate the reflection coefficient from the calibration plane to the aperture plane. Subsequently, a rational function model (Stuchly et al. 1994, Anderson et al. 1994) was used to convert the de-embedded reflection coefficient into the complex permittivity of the tissue. The sensing depth of the probe is approximately 1.5–3 mm, depending on the dielectric properties of the material under test, as described in detail in Hagl et al. (2003).
In general, it is assumed that soon after excision, cell death is insignificant, but there is some fluid loss, as well as changes in oxygen tension, pH and temperature\textsuperscript{3}. Of these factors, only fluid loss and temperature are expected to affect the dielectric properties at microwave frequencies. In addition, it has been shown that tissue thermal conductivity is not altered within a few hours after animal death (Duck \textit{et al.} 1990), so we would expect the thermal properties of our tissue samples to be very similar to the thermal properties of \textit{in vivo} tissue. Finally, changes between \textit{in vivo} and \textit{ex vivo} properties of biological tissues can be minimized as long as the measurements are performed within several hours after excision so that minimal liquid loss takes place (Foster and Schwan \textit{et al.} 1989). Since the tissue was obtained within 2 h of excision, and experimental conditions were controlled to prevent excessive fluid loss, we would expect the temperature change to be the only dominant effect on the \textit{ex vivo} tissue dielectric properties.

The dielectric properties data set obtained from these measurements is multi-dimensional in the sense that it depends on both frequency and temperature. In our measurements of each tissue sample, we acquired data at 401 frequency points and approximately 30 different temperatures. Thus the complete raw data set for each tissue sample comprised over 10,000 points. We reduced the complexity of this data set in two steps. First, the wideband dielectric properties data at each temperature for each sample were fit to a single-pole Cole–Cole model. This reduced the number of data points at each temperature to four (the number of free parameters in the Cole–Cole model). Second, the temperature dependence of the Cole–Cole parameters for all tissue samples was fit to a second-order polynomial function. This further collapsed the data set down to a total of only 12 parameters, yielding an ultra-compact representation of the wideband temperature-dependent dielectric properties. A detailed description of the data reduction and fitting procedure is presented in the appendix.

3. Validations

We validated the performance of the precision probe at elevated temperatures by measuring the dielectric properties of ethanol and methanol at 30 °C, 40 °C and 50 °C, and comparing the results to data published by Buckley and Maryott (1958). The maximum difference between our measurements and the published methanol data was approximately 2 units and 0.2 S m\textsuperscript{−1} for dielectric constant and conductivity, respectively. The maximum difference between our measurements and the published ethanol data was approximately 1 unit and 0.1 S m\textsuperscript{−1} for dielectric constant and conductivity, respectively. Since these differences are small, we conclude that the probe performance is not degraded at high temperatures.

To investigate the validity of fitting the wideband dielectric properties data at each temperature to a single-pole Cole–Cole model, we calculated the difference between the measurement data and the single-pole Cole–Cole fit at each frequency and temperature:

\[ \delta \varepsilon_r, \text{fit} = \varepsilon_r, \text{measured} - \varepsilon_r, \text{fit} \]  
\[ \delta \sigma, \text{fit} = \sigma, \text{measured} - \sigma, \text{fit}. \]

3 Personal communication: Professor Patricia Keely, Department of Pharmacology, University of Wisconsin–Madison, September 2005.
figures 2(c) and (d) show the differences between the raw data and the two-pole Cole–Cole fit. The two-pole model does not result in increased accuracy in fitting the raw data and leads to a more complex representation due to the increased number of parameters. Therefore, we conclude that the single-pole model is sufficient to represent these data.

4. Results and discussion

Figure 3 shows an example of the tissue temperature variation as a function of time during a heating and cooling cycle. The temperatures of the two fluoroptic thermometers along with the average are shown. As can be seen, there is a slight temperature difference between the two probes (the largest difference observed between the two probes in any experiment was 3.7 °C). We adjusted the oven temperature continuously during the heating phase to ensure that the heating rate remained approximately constant. However, since there was no active cooling mechanism, the cooling rate was not constant, and decreased as a function of time.

Figure 4 shows example plots of the dielectric properties of liver tissue (symbols) as a function of frequency at three distinct temperatures along with the one-pole Cole–Cole fits to the experimental data (lines). As is evident from the plots, the complex permittivity trend with temperature differs over frequency. In addition, there are ‘cross-over’ points, which are the frequencies where the permittivity and conductivity do not change with temperature. The dielectric constant curve has a cross-over point at about 4 GHz. Below this point, the
Figure 3. Example of the tissue temperature variation as a function of time. The temperatures of both fluoroptic thermometers along with the average are shown.

Figure 4. Example of (a) dielectric constant and (b) conductivity of liver tissue as a function of frequency at three distinct temperatures. The symbols represent the experimental data while the lines represent the one-pole Cole–Cole fit to the experimental data.

Permittivity decreases slowly as temperature increases. This trend reverses (and increases in magnitude) above the cross-over point. The conductivity curve has two cross-over points: one near 2–3 GHz and one near 16 GHz. Below the first cross-over point, the conductivity increases slowly as temperature increases. In the frequency range of \(\sim 3–15\) GHz, the conductivity decreases as temperature increases, and above \(\sim 15\) GHz, the conductivity again increases as temperature increases. These trends are consistent with those for water (Komarov et al. 2005). This figure also demonstrates that the one-pole Cole–Cole fit is an excellent match to our experimental data over this frequency range. This supports the data shown in figures 2(a) and (b).

4.1. Wideband analysis

To quantify how well the global Cole–Cole model with temperature-dependent parameters fits the individual data sets, we used the quadratic coefficients defined in (A.2)–(A.5) to reconstruct
the Cole–Cole parameters at specific temperatures, and then used those parameters in (A.1) to calculate the complex permittivity as a function of frequency. The difference between the reconstructed curves and the measured curves was defined as

$$\delta \varepsilon_{\text{r,recon}} = \varepsilon_{\text{r,measured}} - \varepsilon_{\text{r,reconstructed}} \quad (2a)$$

$$\delta \sigma_{\text{recon}} = \sigma_{\text{measured}} - \sigma_{\text{reconstructed}} \quad (2b)$$

The above quantities are expressed as absolute differences instead of per cents due to the fact that the conductivity takes on very small values at low frequencies, which can result in very large, though meaningless, per cent errors.

The temperature-dependent Cole–Cole parameters for the cooling cycles of all experiments are shown in figures 5(a)–(d). The circles are the Cole–Cole coefficients of a fit of a measurement to a single-pole Cole–Cole function. The solid curves are the quadratic fits to the ‘super-set’ of the parameters from all experiments. As such, the quadratic curves represent an average fit to all the temperature-dependent Cole–Cole parameters. Table 2 lists the quadratic coefficients corresponding to the solid curves in figures 5(a)–(d).

In addition, figures 5(e) and (f) show the difference between the raw data and the reconstructed data obtained by using the second-order polynomial, as defined in (2a) and (2b).
The maximum difference due to the reconstruction is approximately 5.6 units and 5.4 S m$^{-1}$ for permittivity and conductivity, respectively. The difference in the permittivity increases at lower frequencies because the error in our processing scheme increases at low frequencies. This accuracy level is comparable to that achieved by Gabriel et al. (1996b) in fitting a Cole–Cole model to experimental data. The difference between the experimental data and the Cole–Cole fit presented in Gabriel et al. (1996b) for a variety of animal and human tissues, including liver, is on the order of 1–10 units and 1–10 S m$^{-1}$ for permittivity and conductivity, respectively, over our frequency range of interest.

The differences due to the reconstruction arise because there is a variation in the baseline (room temperature) liver dielectric properties, possibly due to slightly different fluid (blood) and fat contents in the different samples (Schwan and Li, 1953). Since the quadratic fit is an averaging process, the reconstruction works much better for samples whose properties are closer to the average than for samples whose properties deviate from the average.

We observed a distinct and repeatable difference between the dielectric properties measured during the heating cycle and those measured during the cooling cycle. In particular, the frequency-dependent permittivity and conductivity measured at different temperatures during the cooling cycle was always well behaved, easily fit to a Cole–Cole model, and consistent with trends expected both from our previous experience and the extensive literature database of microwave tissue properties. In contrast, measurements of the permittivity and conductivity at different temperatures during the heating cycle had unexpected but very repeatable anomalies that were not amenable to modelling by a Cole–Cole equation. The consistency of these anomalies over many measurements argues that they are a consequence of non-equilibrium effects in the tissue that occur during the heating cycle, but not during the cooling cycle. Chin and Sherar (2001) have previously claimed that irreversible structural changes take place during tissue heating but are negligible during tissue cooling. However, as discussed previously, irreversible structural changes that would affect the dielectric properties are not expected to occur below approximately 60 °C. Moreover, we have conducted experiments in which the tissue was heated to ~40 °C, cooled to ~25 °C, reheated to ~60 °C, and cooled back down to room temperature. In these experiments, we observed that the dielectric properties measured during both cooling and heating transients, including the same anomalies, are repeatable. This supports two conclusions: (1) the processes occurring during the heating cycle that produce the anomalous dielectric properties measurements are not due to irreversible structural changes of the tissue, and (2) those same processes are absent during the cooling cycle.

According to previously published studies (e.g., Chung et al. (1999), McDannold et al. (2000), Hazle et al. (2002)), whose primary objective is to establish the thermal dose guidelines for tissue necrosis, the temperature at which irreversible structural changes are expected to occur is somewhat ambiguous, since thermal dose is a time-temperature process. In contrast, we are only concerned with irreversible changes that affect the dielectric properties, rather than cell viability, blood coagulation, etc. To that end, while we used 60 °C as an approximate threshold below which irreversible structural changes that affect the dielectric properties are not expected to occur, the ability to cycle the temperature and obtain repeatable, physically sensible (e.g., consistent with Cole–Cole model) results was ultimately the basis upon which we concluded that there were no evident structural changes of significance to microwave dielectric spectroscopy below approximately 60 °C.

At this time, we are not able to explain the heating cycle data, but we find the cooling cycle data to be both repeatable and well behaved and accurately described by a physically intuitive Cole–Cole model. Thus, we only display the results for the cooling cycle. Since most dielectric properties measurements are performed on recently excised tissue after it experiences...
passive cooling, our analysis on the cooling cycle data remains applicable to many practical situations.

Figure 6 shows an example of a comparison between the experimental data (symbols, the same data as presented in figure 4) and the one-pole Cole–Cole curve obtained using the global temperature-dependent Cole–Cole parameters shown in table 2 (lines). As this figure demonstrates, the dielectric properties predicted by the proposed method agree favourably with the experimental data. This supports the data shown in figures 5(e) and (f).

The Cole–Cole parameters obtained using this approach were compared to the most relevant pole of the four-pole Cole–Cole fit presented by Gabriel et al. (1996b). The two data sets compare favourably, with two important differences: (1) unlike in this study, $\varepsilon_\infty$ is not used as a fitting parameter in the data presented in Gabriel et al. (1996b), but instead is set to 4.0, and (2) since Gabriel et al. (1996b) use a four-pole model to fit data over an extremely wide frequency range, the value for $\sigma_i$, which is an inherently dc quantity, is significantly different. Table 3 shows a comparison between the Cole–Cole parameters at 37 °C calculated using the quadratic coefficients in table 2 and the values for the first pole presented in Gabriel et al. (1996b).

4.2. Single-frequency analysis

To compare our data to those previously reported in the literature, we calculated linear temperature coefficients for our data at 915 MHz and 2.45 GHz separately for the heating and
cooling cycles. Figure 7 shows an example of the temperature dependence of the permittivity and conductivity at 915 MHz and 2.45 GHz. The average linear temperature coefficients for all experiments along with the standard deviations are shown in Table 4. We calculated the linear temperature coefficients by plotting the permittivity and conductivity for the cooling cycles at a specific frequency as a function of temperature and fitting a line to the data. The slope of the
Figure 8. Comparison between temperature coefficients from this study and previous studies.
Error bars correspond to the standard deviation in this study and the errors reported in Chin and Sherar (2001, 2004). The data given in Foster et al (1979) are for brain tissue, the data in Chin and Sherar (2004) are for prostate tissue, and the data from the other references are for liver tissue.

line, which is the change in permittivity or conductivity per degree Celsius, was divided by
the average value of the permittivity or conductivity, respectively, over the temperature range
under consideration, to obtain the temperature coefficients.

Although we calculated linear temperature coefficients to compare our data to those
previously reported, it is clear from figure 7 that linear temperature coefficients are
not appropriate over such large temperature ranges, particularly for the conductivity at
2.45 GHz. Since the conductivity curves have a cross-over point near 2.45 GHz, the behaviour
of conductivity as a function of temperature is not straightforward, as demonstrated in
figure 7. This quadratic behaviour was also noted by Sipahioglu et al (2003), who measured
the dielectric properties of a variety of meats as a function of temperature (over a similar
temperature range as this study) at 2.45 GHz. The explanation suggested for this trend is as
follows. The total dielectric loss is a sum of ionic and dipole losses, and since the dipole
loss decreases with temperature, and the ionic loss increases with temperature (as seen from
figure 5(d)), the total dielectric loss will first decrease, and then increase with temperature
(Sipahioglu et al 2003).

Figure 8 shows a graphical comparison between the temperature coefficients found in
this study and those reported previously. Since, to the best of our knowledge, temperature
coefficients at 2.45 GHz have not been previously reported, only the temperature coefficients
for our data are shown at 2.45 GHz. The temperature coefficients for this study are separated
according to the heating and cooling cycles, while, for clarity, only one set of temperature
coefficients are shown for the literature data. We note that the average values from our
experiments are consistent with data presented in the literature. The temperature coefficients
for both the heating and cooling cycles are within the ranges of the previously reported
temperature coefficients. Therefore, even though we were not able to obtain adequate Cole–
Cole fits to the UWB heating cycle data, we are confident that our data for both cycles are
consistent with previously reported measurements. In addition, the permittivity temperature
coefficients at 915 MHz and 2.45 GHz are of the same magnitude, while the conductivity
temperature coefficients are significantly smaller at 2.45 GHz than at 915 MHz. This is
consistent with the observation that the conductivity curves have a cross-over point near 2–3 GHz.

5. Conclusion

We have measured and characterized the UWB dielectric properties of animal liver tissue over the microwave frequency range from 0.5 to 20 GHz and the temperature range from room temperature to \( \sim 60 \degree C \). We demonstrated that linear temperature coefficients are impractical and inappropriate over wide frequency and temperature ranges. Instead, we presented a new compact data-cataloging technique, whereby the frequency dependence of the dielectric properties is modelled via a single-pole Cole–Cole model, and the temperature dependence of Cole–Cole parameters is modelled by a second-order polynomial. This simple and elegant technique enables researchers to conveniently estimate the liver dielectric properties at any frequencies and temperatures in the specified range.

Acknowledgments

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Appendix.

Using a MATLAB fitting function that performs the Nelder-Mead direct search optimization, we fitted the data at each temperature for each experiment to a single-pole Cole–Cole model over a frequency range of 0.5–20 GHz:

\[
\hat{\epsilon}(\omega) = \epsilon_\infty + \frac{\Delta\epsilon}{1 + (j\omega\tau)^{1/\alpha}} + \frac{\sigma_i}{j\omega\epsilon_0}.
\]  

(A.1)

In this equation, \( \hat{\epsilon}(\omega) \) is the complex relative permittivity expressed as a function of angular frequency \( \omega \), and \( \epsilon_\infty, \Delta\epsilon, \tau, \sigma_i \) and \( \alpha \) are parameters chosen to fit the model to experimental data. To limit the number of fitting parameters, we set \( \alpha \), which is an empirical parameter that accounts for the observed broad distribution of relaxation time constants in tissue, to 0.1. We assumed that \( \alpha \) does not change with temperature and is consistent with the value reported by Gabriel et al (1996b) for liver. Limits were set on the parameters so that they would remain within physical ranges (e.g., the lower bound on \( \epsilon_\infty \) was set to 1). A single-pole Cole–Cole model was used because only one pole with a ps time constant dominates the tissue dielectric response over the frequency range of interest.

We combined the temperature-dependent parameters from all experiments into a single ‘super-set,’ plotted them as a function of temperature, \( T \), and fitted them to a second-order polynomial using a MATLAB function that performs a least-squares polynomial fit:

\[
\epsilon_\infty(T) = A_1 T^2 + B_1 T + C_1
\]  

(A.2)

\[
\Delta\epsilon(T) = A_2 T^2 + B_2 T + C_2
\]  

(A.3)

\[
\tau(T) = A_3 T^2 + B_3 T + C_3
\]  

(A.4)

\[
\sigma_i(T) = A_4 T^2 + B_4 T + C_4.
\]  

(A.5)
We determined that a second-order polynomial was sufficient to represent the data while minimizing the number of coefficients from higher-order fitting functions. This procedure was repeated for all four Cole–Cole parameters.

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