Feasibility of water content-based dielectric characterisation of biological tissues using mixture models

Feasibility of Water Content-Based Dielectric Characterisation of Biological Tissues using Mixture Models

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ABSTRACT

Objective: **This study quantitatively examines the validity of mixture formulae as models to describe the microwave-range dielectric properties of biological tissue of varying water content.** ***Methods:*** **Mixture formulae, specifically the Maxwell Garnett and Bruggeman models, are used to predict the dielectric properties of** ***ex-vivo*** **bovine muscle and liver tissue samples varying in water content. The tissues are modelled as comprising of cell and macromolecule inclusions in a water matrix. The model predictions are compared to dielectric measurements performed with a network analyser and dielectric probe in the 0.5 – 8.5 GHz band.** ***Results:*** **There was a poor match between the properties predicted by the models and the measured results at most frequency points, for both tissue types. However, the overall predicted and measured trends over the measured band correlated well.** ***Conclusions:*** **The Maxwell Garnett and Bruggeman models may prove a valuable tool aiding in the characterisation of the dielectric properties of materials with different water contents, however, currently, direct application of the model with the assumption of solid inclusions in a water matrix is not feasible without substantial improvement to the models.** ***Significance:*** **The dielectric properties of biological tissue are of fundamental importance for many medical technologies ranging from diagnostic to therapeutic. There is a need for continuous improvements to be made to the techniques used to measure and characterise the dielectric properties of tissues. Mixture models are investigated in this study as potentially a valuable candidate modelling technique for the dielectric profiling of tissues based on water content.**

Index Terms — **Biodielectrics, Dielectric Properties, Mixture Models.**

1 INTRODUCTION

THE dielectric properties describe the interaction of electric fields with matter. These properties include the relative permittivity (ε_r) and conductivity (σ). The relative permittivity describes the extent to which charge is polarised under an applied field, and hence is a measure of energy storage. The conductivity describes the flow of charge under the applied field, and hence measures energy loss [1]. The dielectric properties of biological tissues have been an area of study extending back decades [2] with the comprehensive literature review of Gabriel *et al.* resulting in the creation of a large database covering many tissues over a wide frequency range [3], [4]. Databases such as Gabriel's, though an excellent resource, may not provide enough information for all applications. Specifically, measurement of dielectric properties of biological tissue involves many potential confounding factors such as the source of the tissue (animal or human), whether the tissue is measured *in-vivo* or *ex-vivo*, the state of hydration, natural heterogeneity in any given tissue, temperature, measurement technique, sample size, measurement accuracy and uncertainty, the presence of pathology, and animal age [5]. Therefore, most of the recent studies have focused on characterisation of sources of error in the dielectric measurements and properties of tissues affecting the dielectric [6]–[9].

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Dielectric properties depend on the structure and composition of tissues [10]. As a general trend, relative permittivity tends to decrease with increasing frequency while conductivity tends to increase [11]. The properties are of importance as they describe how the human body reacts to exposure to electromagnetic (EM) radiation. Directly linked are established and emerging medical sciences, diagnostic and therapeutic technologies such as dosimetry, microwave imaging and microwave ablation, that require an accurate knowledge of *in-vivo* dielectric properties of body tissue [12], [13]. As such, knowledge of the dielectric properties of tissues allow assessment and optimisation of such technologies in terms of safety and efficacy. As most of the tissue dielectric data in literature is based on *ex-vivo* dielectric measurements, there is a need of *in-vivo* tissue dielectric characterisation either via direct *in-vivo* measurements or by modelling the factors contributing to the properties. Direct *in-vivo* measurements of human body tissue may not be possible due to invasive nature of the measurement equipment, therefore, modelling the *in-vivo* dielectric properties of tissue as a function of measurable contributors can be a suitable alternative.

Mixing formulas are models that describe the macroscopic dielectric properties of materials in relation to the constituent components [14]. With regards to biological tissues, the extracellular fluid, cells and extracellular macromolecules (proteins) are the components of interest [10], [14]. Different mixing formulae exist, including the Maxwell Garnett (Rayleigh), Bruggeman, Looyenga and Birchak formulae [14]. These formulae model a tissue as comprising of inclusions in a homogenous background medium. All of these models have assumptions and limitations. For example, a requirement that the wavelength of the EM radiation be much larger than the size of the mixture inclusions is needed for accurate modelling [14]. Another finding is that accuracy is best for dilute mixtures [14]. Further, the different models can give different results, particularly in cases of larger volume fractions or the case of high contrast in dielectric properties between the inclusions and background. However, the models have been proven useful, if these assumptions and limitations are respected [14].

A good model should accurately predict results theoretically that match those derived experimentally. For example in [15], Smith *et al.* studied the dielectric properties of adipose and bone marrow tissues from 1 kHz – 1 GHz and found good correlation with those predicted from the Maxwell Garnett model. A good model may also be used to map the measured dielectric properties of a tissue sample of initially unknown makeup to an accurate profile of the histology. Such techniques have powerful diagnostic potential [16]–[20]. In [16], the relative permittivity of blood, fat, liver and brain were measured from 10 kHz – 10 MHz. Liver and brain were then modelled as mixtures of blood and fat, and various mixing formulae applied to attempt to predict the volume fraction of blood in these tissues, with a poor correlation found. In [17] and [18], mixture models were applied to dielectric measurements of cancerous tissue samples and showed the ability to correlate the values, using the models, to the volume fraction of cancerous cells in the tissue. In [19], dielectric measurements of plasma and erythrocyte cytoplasm were compared with the Maxwell Garnett model able to effectively model the dielectric properties of the two fluids. Finally in [20], mixture models were applied to dielectric spectroscopy of single cells to aid in characterisation with potential applications including cancer staging, monitoring of the health of cell lines and even intricate cell level biology such as the study of ion channels. As such, mixing formulas have clear value in correlating dielectric properties to accurate and important histological information about tissues.

This paper examines the validity of mixing formulae, applied to biological tissue, in response to EM radiation in the range 0.5 GHz – 8.5 GHz. This covers the frequency range of interest for many emerging medical applications [13]. At these frequencies, the dielectric properties are largely dependent on the free water content [11]. Hence, a model with water content as the main parameter of focus is of interest. The ability of two typical mixing formulae, the Maxwell Garnett (MG) and Bruggeman (BG) models, to accurately predict the dielectric properties of tissue samples based solely on knowledge of volume fractions of water and "not-water" solid is assessed with respect to experimentally measured values. This work provides a unique contribution, as previous studies have examined the validity of mixing models to predict the dielectric properties only in lower frequency bands [15], and have not specifically investigated water content [16]–[20].

The paper is laid out as follows. In the next section a brief theoretical description of mixture models is given, with emphasis on the MG and BG models. Following this, Section 3 discusses the sample preparation, dielectric measurements and then reports and discusses the experimental results to those predicted by the MG and BG models for *ex-vivo* bovine muscle and liver tissue samples of varying water content. Section 4 then concludes the paper discussing the suitability of the MG and BG models for modelling the dielectric properties in the microwave band for tissue of varying water content such as those measured, as well as examining study limitations and future work.

2 MIXTURE MODELS

Mixture models attempt to describe, using mathematical formulae, the dielectric properties of a mixture in terms of the dielectric properties of the components [21]. In these formulae a lossy medium, is described by the complex permittivity:

$$\varepsilon = \varepsilon' - j\varepsilon'', \tag{1}$$

where ε is the complex permittivity made up of a real part, ε' (also denoted as ε_r) and imaginary part ε''. The real part, ε_r is the relative permittivity, while the imaginary part, ε'', relates to conductivity, σ, according to:

$$\sigma = 2\pi f \varepsilon_0 \varepsilon'' + \sigma_i, \tag{2}$$

where f is the frequency in Hz, ε_0 is the permittivity of free space and σ_i is the conductivity under DC conditions.

A given mixture is modelled as being made up of inclusions in a background medium (matrix). Biological tissues, particularly when the electromagnetic stimulus of interest is in the microwave band, may be modelled as a mixture comprising of cellular and macromolecule solid inclusions in a water matrix. These inclusions are taken as being simple geometric shapes like spheres [21]. Overall, the mixture is assumed to have a macroscopic complex permittivity, which is valid if the mixture is homogenous. Even if it is not, this assumption holds if the wavelength of the EM radiation is much larger than the inclusions [21]. Microwave imaging systems for example usually use the part of the microwave band from 300 MHz – 30 GHz, equivalent to electromagnetic waves from 1 m to 1 cm in free space [13].

Consideration needs to be made as to whether the assumption that the wavelength is larger than the inclusions is valid for a given application. For biological tissues, the largest inclusions of interest are cells, with the largest cell in the body, the ovum being 0.1 mm in diameter while muscle fibers, for example, are of the order 50 µm in diameter [22]. Biological tissues, as a medium, will affect the speed of propagation of an electromagnetic wave and hence the wavelength. For instance, with an 8.5 GHz wave (the extreme point of the range considered in this study), in a medium of water, with a ε_r value of 72 at that frequency (at 25 °C) [4], the wavelength reduces from 35.3 mm in free space to 4.2 mm in water. Hence, for a study including cells the size of the ovum for example, lower frequency microwaves would be necessary to maintain the validity of this assumption.

The response of an inclusion to an electric field is measured as the polarizability of the inclusion. This polarizability can be readily calculated for a single sphere, with the complex permittivity of the entire medium than derived, assuming knowledge of the complex permittivity of the background and the concentration of spherical inclusions [21]. It is convenient to express the complex permittivity of the medium (ε_{eff}) in terms of complex permittivities of the matrix (ε_e), inclusions (ε_i) and the volume fraction of the inclusions (v_f). This expression is given as the Maxwell Garnett (MG) formula [14], [21]:

$$\varepsilon_{eff} = \varepsilon_e + 3\varepsilon_e v_f \frac{\varepsilon_i - \varepsilon_e}{\varepsilon_i + 2\varepsilon_e - f(\varepsilon_i - \varepsilon_e)}\ . \tag{3}$$

The MG formula describes ε_{eff} linearly if the contrast between the dielectric properties of the matrix and inclusions is close to unity. Further there is the requirement of a sparse mixture, as with large volume fractions assumptions such as negligible interaction between inclusions, and equivalency of the external and local EM fields, are not valid [14], [21]. A further drawback with the MG model is that it is not symmetric, meaning that if the properties and volume fractions of the matrix and inclusions are interchanged the resultant value of ε_{eff} is not the same [14]. The Bruggeman (BG) model improves on the MG by being symmetric and is expressed as [14]:

$$\frac{\varepsilon_i - \varepsilon_{eff}}{\varepsilon_i + 2\varepsilon_e} + \left(1 - v_f\right)\frac{\varepsilon_e - \varepsilon_{eff}}{\varepsilon_e + 2\varepsilon_{eff}} = 0 \tag{4}$$

A further point to note is that these mixing formulae were originally derived for lossless media. The application of these models to biological tissues, which are lossy media, introduces some additional assumptions and limitations. Key among these assumptions is the requirement that the inclusion diameter be smaller than the skin depth, which is satisfied when considering microwaves as the EM source and cells (or smaller macromolecules) as the tissue inclusions [14].

Although mixing models, and in particular the MG and BG models described here, rely a number of simplifications and assumptions, accurate modelling of tissues can be achieved if these are adhered to [14]. In the next section the performance of the model when applied to biological tissues is examined.

3 TISSUE SAMPLES AND MEASUREMENTS

In this section, the preparation and handling of the bovine *ex-vivo* muscle and liver samples is described as well as the calculation of the volume fraction of matrix and inclusion in each sample. Next, the basis of the experimental and theoretical measurements of the relative permittivity (ε_r) and conductivity (σ) of the samples is outlined. Finally, the experimental results, and the theoretical results predicted using the MG and BG models, are reported and discussed.

3.1 TISSUE SAMPLES AND WATER CONTENT

Bovine liver and skeletal muscle tissues were obtained post-mortem from a local abattoir. The tissues were transported to the laboratory in sealed Styrofoam containers within an hour of death. Both of these tissue samples were then equally divided by mass into 7 separate samples, referred to as L0 – L6 for the liver sample set and M0 – M6 for the muscle sample set.

As discussed in section 1, biological tissue is modelled in this study as comprising of cellular and macromolecular inclusions in a matrix of extracellular fluid, the latter being effectively water, and the former tissue solids. Further the dielectric properties in the microwave band are largely dependent on this water content [11]. Hence, a key part of this study was to vary the water content of the tissue samples, and to examine the performance of the MG model when applied to tissues of differing levels of dehydration.

Table 1. Water and Solid Mass Fractions of the L0 and M0 tissue samples

Tissue	Water Mass Fraction	Solid Mass Fraction
Liver	0.67	0.33
Muscle	0.74	0.26

The L0 and M0 samples were used solely to calculate the ratio of water to solid in the two tissue types. In order to empirically calculate the mass fractions of water and solid in the original tissues, the L0 and M0 samples are initially weighed on a mass balance and then allowed dry out. Periodically over the drying process the weights of the samples are taken until there is a final point reached where no change in mass is observed. At this point, it is assumed all the water has evaporated and the material remaining is purely the solid mass fraction. The difference in mass between this final reading and the first is then assumed to be the mass of water lost, giving the water mass fraction of the samples. It is assumed these water and solid mass fractions are the same for the other samples from the respective tissue. The water and solid mass fractions for both the liver and muscle tissues are given in Table 1.

Table 2. Water and Solid Volume Fractions of the L0 and M0 tissue samples

Tissue	Water Volume Fraction	Solid Volume Fraction
Liver	0.70	0.30
Muscle	0.78	0.22

As the MG model requires volume fractions, the mass fractions were converted using density values. The density of water was taken as 1 g/cm^3, while the densities of the bulk tissues were taken from the literature as 1.051 g/cm^3 for liver [23] and 1.0597 g/cm^3 for skeletal muscle [24]. These values, combined with the mass fractions from Table 1, were used to derive the water and solid volume fractions for the tissues, which are given in Table 2.

Table 3. Water and Solid Volume Fractions of the liver samples

Liver Sample Number	Water Volume Fraction	Solid Volume Fraction
L1	0.70	0.30
L2	0.55	0.45
L3	0.36	0.64
L4	0.23	0.77
L5	0.16	0.84
L6	0.04	0.96

Table 4. Water and Solid Volume Fractions of the muscle samples

Muscle Sample Number	Water Volume Fraction	Solid Volume Fraction
M1	0.78	0.22
M2	0.75	0.25
M3	0.65	0.35
M4	0.41	0.59
M5	0.21	0.79
M6	0.07	0.93

The remaining samples are then dehydrated to varying extents using a microwave oven, except for the L1 and M1 samples which hence are assumed to have water and solid volume fractions as quoted in Table 2. L1 and M1 are taken as being non-dehydrated samples and closest to the original *ex-vivo* samples taken. The other samples are used to model tissues of increasingly low water content and are heated for variable amounts of time (1 – 30 minutes) in the microwave oven, with the mass of the sample taken before and after the heating. The difference in mass is assumed to be lost water and this, combined with knowledge of the original water and solid mass fractions allows calculation of the new water and solid mass fractions of the dehydrated samples, which are then converted to volume fractions as described above. Tables 3 and 4 show the water and solid volume fractions of the samples for the two tissue types, liver and muscle respectively.

Tissue water loss is generally non-uniform with the greatest loss occurring on the tissue surface [6]. This non-uniformity of water content poses challenges in measuring the average dielectric properties of the sample. Further, the available techniques used to measure these properties in the microwave band are not optimised for solid or rigid samples. Dielectric measurements of the samples were performed using a slim-form probe, with more detail on this given in the next section. The probe is touched against the sample material in order to take a measurement. The probe measures the properties of the area of the material closest to the probe tip, and assumes the contact is uniform and smooth. The result would be measurements dominated by the properties of the material nearest the tip, effectively that of the surface and hence may not be representative of the whole sample. Further, the surface of the tissue was uneven with the possibility of air gaps causing measurement error.

Figure 1. Example of the Muscle tissue samples used in this study in both the non-homogenised (left) and homogenised (right) states. The dielectric properties of the homogenised samples were measured when placed in test-tubes as shown. The homogenised tissue is also shown in a petri dish for illustration.

In order to tackle the issue of uniformity and to obtain measurements representative of the whole, the samples were homogenised. The homogenisation technique employed was that of cryogenic grinding, where the samples are frozen using liquid nitrogen, rendering the tissue brittle, before being ground using a mortar and pestle [25]. Upon evaporation of the liquid nitrogen, the homogenised samples were put in sealed test tubes and allowed to warm up to ambient temperature. The slim form probe was then immersed to a depth of 5 mm into the samples and measurements taken. The temperature for these measurements was again 23°C ± 0.5°C.

Figure 1 shows a sample of muscle used, both in the non-homogenised and then homogenised state.

3.2 DIELECTRIC MEASUREMENTS AND THEORETICAL ANALYSIS

For each of the tissue samples, L1 – L6 and M1 – M6, experimental and theoretical values of the dielectric properties of relative permittivity and conductivity were obtained across the 0.5 GHz – 8.5 GHz band. The experimental values were measured directly, while the theoretical were calculated using both the MG and BG models.

The experimental protocol for each sample involved selection of 5 random sites, with 5 sets of measurements taken at each of these sites. The mean permittivity values for each sample was then calculated. Each dielectric measurement contained 101 frequency points, linearly spread across the 0.5 GHz – 8.5 GHz band. Measurements were conducted using the Keysight E5063A ENA Series Network Analyser with an attached slim-form probe from the Agilent 85070E Dielectric Probe Kit. The measured complex permittivity, measured as real and imaginary parts, is converted into the relative permittivity and conductivity as described in equations (1) and (2).

To validate the dielectric measurements of tissues, the measurement uncertainty of the system was evaluated using a 0.1 M aqueous solution of sodium chloride at 23°C. Repeated measurements were performed on this reference solution at each of the 101 measurement points in the band and averaged. The uncertainty was calculated as the combined uncertainty, a function of 3 separate parameters quantifying random and systematic error [5]. These parameters included measures of Repeatability, calculated as the percentage the standard deviation of measurements from the mean; Accuracy, calculated as the percentage by which the measurement deviated from reference [5]; and drift, calculated as the percentage of systematic drift in measurement over time. A fourth source of uncertainty, cable movement, was irrelevant to this setup as no cable was involved. The calculation of the combined uncertainty from these parameters in given by:

$$Combined\ Uncertainty = \sqrt{\sum_{n=1}^{n=3} p_n^2}\,, \quad (5)$$

where p_n is one of the parameters, with there being 3 in total.

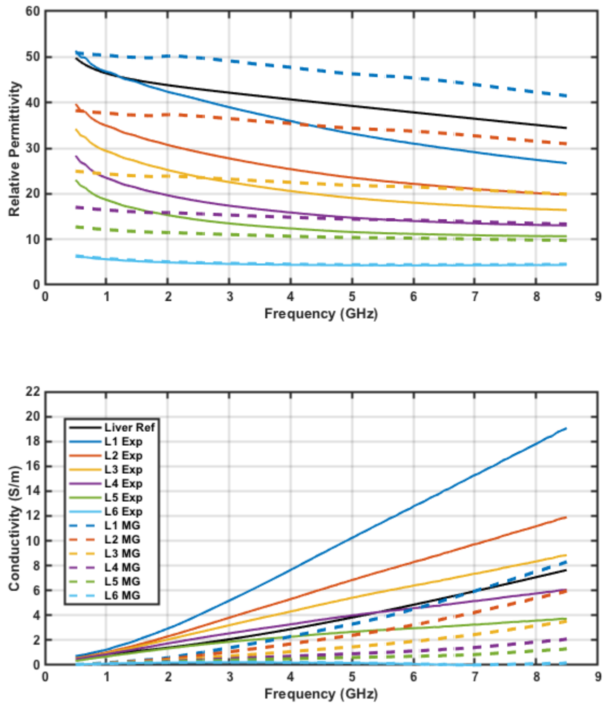


Figure 2. Relative permittivity (ε_r) and conductivity (σ) over the 0.5 GHz – 8.5 GHz band for liver tissue samples L1 – L6, numbered in order of increasing solid volume fraction. Both the experimental (solid lines) and theoretical values predicted by the MG model (dashed lines) are shown for each sample. Also shown are reference values for liver from the IT'IS database (solid black lines).

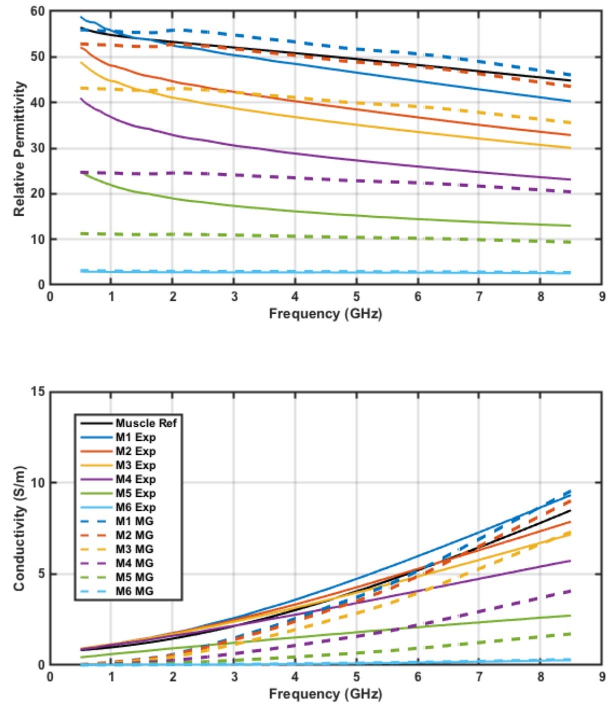


Figure 3. Relative permittivity (ε_r) and conductivity (σ) over the 0.5 GHz – 8.5 GHz band for muscle tissue samples M1 – M6, numbered in order of increasing solid volume fraction. Both the experimental (solid lines) and theoretical values predicted by the MG model (dashed lines) are shown for each sample. Also shown are reference values for muscle from the IT'IS database (solid black lines).

The values for these 3 parameters as well as the combined uncertainty is given in Table 5, for each the relative permittivity and the conductivity.

To calculate the theoretical values for the properties, predicted by the MG model, equation (3) is used to calculate the real and imaginary parts of ε_{eff}, at each frequency point, with the real part equating to the relative permittivity and the imaginary part converted into the conductivity using equation (2). Equation (3) requires prior knowledge of the complex permittivities of the matrix (ε_e), inclusions (ε_i) and the volume fraction of the inclusions (v_f). In this study, the matrix is taken as the water, with the inclusion the "not water" or solid part of the tissue. The values of ε_e across the band were experimentally measured using the dielectric probe with samples of deionised water (at 23°C ± 0.5°C). The values of ε_i across the band for the two tissues were extrapolated from the measured experimental results of L6 and M6, with these values adjusted to compensate for the residual water content leaving theoretical values for wholly dry tissue. The values of ε_i could have also been measured directly from the fully dehydrated L0 and M0 samples, however as L0 and M0 were not fully dehydrated at the time the other samples were measured, a decision was made to use the data from L6 and M6 to calculate ε_i. This ensured all experimental measurements were conducted at the same time, with the same handling protocol and hence error minimisation between samples. Finally, the value of v_f is taken as the solid mass fraction values from Tables 3 and 4 above. In an analogous fashion, the theoretical values for the properties as predicted by the BG model are derived from equation (4).

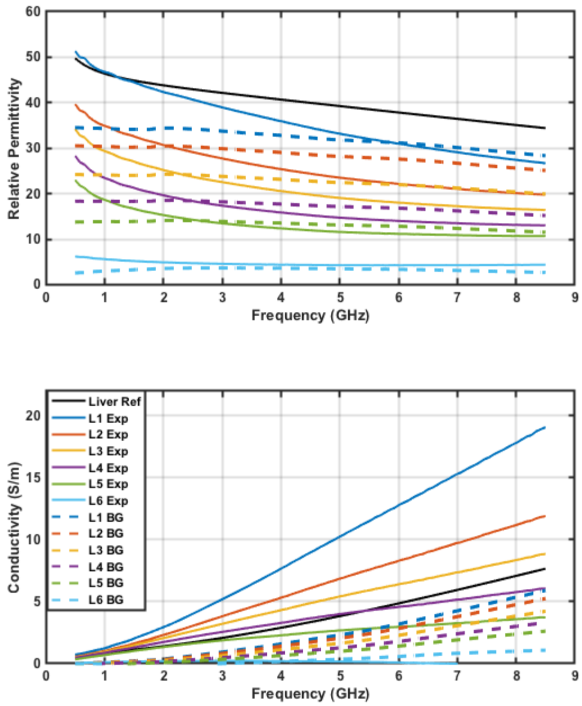
3.3 RESULTS AND DISCUSSION

The experimental and theoretical values as predicted by the MG model for both the relative permittivity and conductivity for the liver tissue samples, across the frequency band, are shown in Figure 2. Also included are the reference values for liver from the IT'IS database of dielectric values for biological tissues [4]. Similarly, the values for the muscle tissue sample set, with theoretical values derived from the MG model are shown in Figure 3.

Corresponding results for the theoretical values as derived from the BG model for liver and muscle tissues are shown in Figures 4 and 5 respectively.

Table 5. Measurement Uncertainty of the system for both relative permittivity (ε_r) and conductivity (σ)

	% Uncertainty in Relative Permittivity (ε_r)	% Uncertainty in Conductivity (σ)
Repeatability	0.45	0.04
Accuracy	0.66	5.00
Drift	0.04	0.06
Combined Uncertainty	**0.79**	**5.0005**



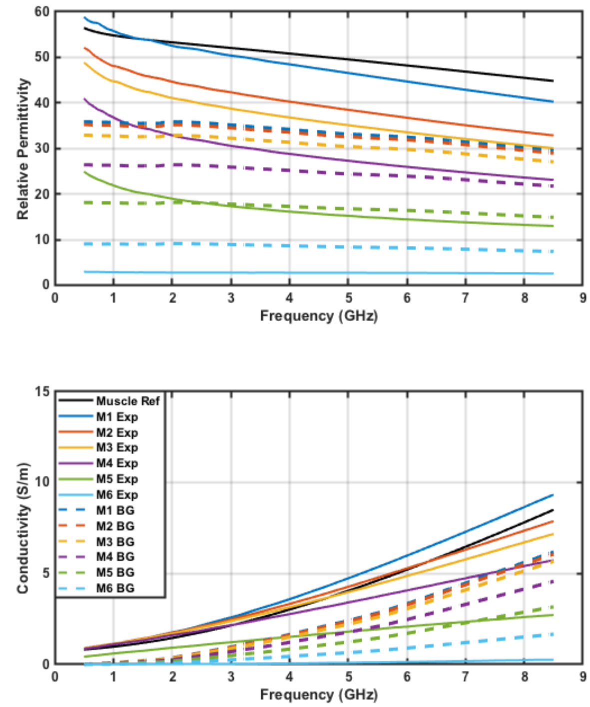


Figure 4. Relative permittivity (ε_r) and conductivity (σ) over the 0.5 GHz – 8.5 GHz band for liver tissue samples L1 – L6, numbered in order of increasing solid volume fraction. Both the experimental (solid lines) and theoretical values predicted by the BG model (dashed lines) are shown for each sample. Also shown are reference values for liver from the IT'IS database (solid black lines).

Figure 5. Relative permittivity (ε_r) and conductivity (σ) over the 0.5 GHz – 8.5 GHz band for muscle tissue samples M1 – M6, numbered in order of increasing solid volume fraction. Both the experimental (solid lines) and theoretical values predicted by the BG model (dashed lines) are shown for each sample. Also shown are reference values for muscle from the IT'IS database (solid black lines).

The results for the MG model (Figures 2 and 3) show that relative permittivity values are, generally, over-estimated by the MG model for high water volume fraction samples and under-estimated for low water volume fraction samples. The conductivity values are generally underestimated in all cases. However, for both relative permittivity and conductivity, the general trend of the theoretical curves match those of the experimental, especially for the conductivity values and in the mid-range of the band for the relative permittivity. Further, the order of the modelled curves matches that of the experimental; lower solid volume fraction (i.e. higher water content) samples having higher relative permittivity and conductivity in all cases.

Table 6. Average Fractional Error (AFE) between the measurement and the MG and BG model predicted values of relative permittivity (ε_r) and conductivity (σ) for each tissue sample across the 0.5 GHz – 8.5 GHz band

Sample	AFE (%) in relative permittivity (ε_r)		AFE (%) in conductivity (σ)	
	MG Model	BG Model	MG Model	BG Model
L1	34.3	10.4	70.0	79.4
L2	38.5	17.2	67.1	73.0
L3	14.0	15.8	73.9	72.8
L4	9.8	12.4	77.3	71.7
L5	16.1	12.4	76.5	67.1
L6	2.9	27.2	83.3	34.3
M1	9.9	29.7	33.3	56.8
M2	24.2	17.1	33.6	52.8
M3	12.0	15.6	38.9	52.6
M4	19.3	13.6	59.9	55.0
M5	34.9	10.9	67.1	42.4
M6	7.3	21.1	10.6	45.9

The BG model results (Figures 4 and 5) show predicted relative permittivity values that are, generally, under-estimated with respect to the measured values for the high water volume fraction samples and over-estimated for the low water volume fractions. The conductivity values are generally underestimated in all cases, as also seen in MG model. Also, in common with the MG model, the general trend of the theoretical curves match that of the experimental.

The error between the experimental and theoretically derived values for each of the dielectric properties for each sample was captured as the average fractional error (AFE), expressed as a percentage. Equation (6) shows the equation for the average fractional error, with Exp_n, MG_n being the experimental and theoretical values respectively at frequency point n, with the experimental value assumed the more accurate.

$$AFE\ (\%) = \frac{100\left(\sum_{n=1}^{n=101} \frac{|Exp_n - MG_n|}{Exp_n}\right)}{101}\ . \tag{6}$$

These average fractional error values are reported in Table 6. In general, for both the MG and BG models, the AFE is lower for the relative permittivity values than for the conductivity. For the MG model the AFE for the muscle tissue samples was noticeably smaller than for the liver tissue samples, with this being reflected in the graphs for that model; the experimental and theoretical samples match better in Figure 3 (muscle) than in Figures 2 (liver) for the MG model. For the BG model the AFE is not significantly different for either tissue type.

4 CONCLUSIONS

Characterisation of the dielectric properties of biological tissues is essential for the development of many emerging electromagnetic-based medical technologies. These properties depend largely on the water content of the tissue, particularly in the microwave band where many of these new technologies operate. This paper explores the ability of mixing formulae, specifically the Maxwell Garnett and Bruggeman models, with tissue structure modelled as that of solid inclusions in a water matrix, to accurately predict the dielectric properties of tissues at microwave frequencies.

Experimental results from liver and muscle tissue samples varying in water and solid volume fractions are compared to the results predicted by the two models. The theoretical results derived from both models follow the general trend of the experimental results over the band for a given sample, however values of the properties at most discrete frequency points show a large difference between the predicted and experimental values. These differences in the experimental and predicted values for the samples across the band ranging from 3 – 39% for relative permittivity and 11 – 84% for conductivity for the MG model, with a similar error profile seen in the BG model.

It is anticipated that future studies would include a more extensive range of tissues and apply different mixture models as well as the possibility of modifying existing models to increase the accuracy between theoretical and experiments values. Biological tissues are complex structures and difficult to characterise in terms of dielectric properties. This study shows mixing formulae, in particular the Maxwell Garnett and Bruggeman models, as applied in this study, are not capable of accurately predicting dielectric properties of tissues. Improvements in modelling may be possible, and should be explored, as mixing formulae have the advantage of being relatively simple models, yet potentially powerful tools in the profiling of dielectric properties of biological tissues.

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