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Electrical conductivity of tissue at frequencies below 1 MHz

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Abstract

A two-pronged approach, review and measurement, has been adopted to characterize the conductivity of tissues at frequencies below 1 MHz. The review covers data published in the last decade and earlier data not included in recent reviews. The measurements were carried out on pig tissue, *in vivo*, and pig body fluids *in vitro*. Conductivity data have been obtained for skeletal and myocardial muscle, liver, skull, fat, lung and body fluids (blood, bile, CSF and urine). A critical analysis of the data highlights their usefulness and limitations and enables suggestions to be made for measuring the electrical properties of tissues.

(Some figures in this article are in colour only in the electronic version)

Introduction

Research into the dielectric properties of biological materials and their variation with frequency has been unfolding for more than a century. The features of the dielectric spectrum of tissue are well reported and understood; there are three main dispersion regions identified as α, β and γ dispersions, and their origins and characteristics have been widely reported (for example, Grant (1981) and elsewhere).

In recent times, the research has been driven by the need to establish a credible database of dielectric properties (relative permittivity ε' and conductivity σ) of all body tissues for use in electromagnetic dosimetry studies. A database published in 1996 (Gabriel *et al* 1996a, 1996b, 1996c and electronically Gabriel and Gabriel 1997) is a frequently used resource for dosimetric studies; it will be referred to as the 1996 database.

While useful, the 1996 database has several limitations pointed out by its authors (Gabriel *et al* 1996c) of which two are particularly relevant at frequencies below 1 MHz: (i) the measurements underpinning the database were carried out on excised tissue while data pertaining to live tissue would have been more relevant in bioelectromagnetics studies and (ii) at frequencies below 1 MHz, where the literature values are scarce and have larger

than average uncertainties, the database provides only a 'best estimate' based on available knowledge.

The purpose of this paper is to add to this knowledge in two ways: by reviewing data published in the last decade (and earlier data not included in the 1996 review) and by reporting and critically analysing new experimental conductivity data. The emphasis in this paper is on conductivity; this is in view of the new data and also because the papers reviewed, with the exception of those by Raicu *et al* (1998a, 1998b, 2000) and Gabriel and Gabriel (1997), report only conductivity data. Permittivity will only be reported where more than one set of values are available and in the context of future work.

Dielectric properties of tissues—a brief review

The review is carried out per tissue type or thematic underline.

Brain tissue: grey and white matter

Latikka *et al* (2001) reported conductivity values at 50 kHz for grey matter (0.28 S m^{-1}), white matter (0.25 S m^{-1}), cerebrospinal fluid (CSF) (1.25 S m^{-1}) and tumours (0.1–0.43 S m^{-1}). They used a monopolar needle electrode during brain surgery on nine patients who had deep brain tumours. According to the authors, the measurements were subjected to several sources of error: bleeding, saline used for washing and leakage of CSF. The authors noted that the measured values differ from those given by Schwan and Kay (1957), Geddes and Baker (1967) and Gabriel *et al* (1996a). They ascribe the differences to the fact that their measurements were done *in vivo*.

A series of papers published in the 1960s and 1970s (Nicholson and Freeman 1975, Nicholson 1965, Ranck and BeMent 1965, Yedlin *et al* 1974) provide conductivity data for cerebrospinal tissue *in vivo*. These papers were not included in the 1996 review because they did not meet the selection criterion—to provide permittivity and conductivity data as a function of frequency. They are however important for two reasons (1) because the measurements were carried out *in vivo* and (2) they used directional sampling probes that are capable of detecting anisotropy in the electrical properties of tissue if present.

Nicholson and Freeman (1975) used low-frequency current source mapping to determine the conductivity tensor for anuran (fogs and toads) cerebellum *in vivo* ($\sigma_x = 0.118 \pm 0.012$ S m^{-1}, $\sigma_y = 0.022 \pm 0.003$ S m^{-1} and $\sigma_z = 0.012 \pm 0.004$ S m^{-1}). The medium is assumed ohmic and unaffected by neuronal activity generated by cerebral activity. The conductivity component is highest in the direction parallel to the fibres (σ_x), and the components transverse to the fibres but parallel to the surface are much lower. Nicholson (1965) had previously reported similar anisotropy for white matter (cat, *in vivo*), that is, a factor of 9–10 between the conductivity along and across the fibres.

Ranck and BeMent (1965) pulsed low-frequency non-stimulating currents on the surface of the dorsal columns in the cervical cord of cats (*in vivo*) to obtain the electrical conductivity along (0.47 S m^{-1}) and across fibres (0.08 S m^{-1}) and reported the conductivity of CSF to be 1.67 S m^{-1}.

Yedlin *et al* (1974) obtained data for the conductivity of the molecular and granular layers of the cerebral cortex (cat, *in vivo*) in three mutually perpendicular directions. In the molecular layer, $\sigma_x = 0.33 \pm 0.07$ S m^{-1} and $\sigma_z = 0.28 \pm 0.07$ S m^{-1} and in the granular layer, $\sigma_x = 0.18 \pm 0.05$ S m^{-1} and $\sigma_z = 0.15 \pm 0.03$ S m^{-1} where the x-axis is along the parallel fibres, the z-axis is normal to the cortical surface and the y-axis is orthogonal to the defined axes. The granular layer is more isotropic than the molecular layer due to structural difference

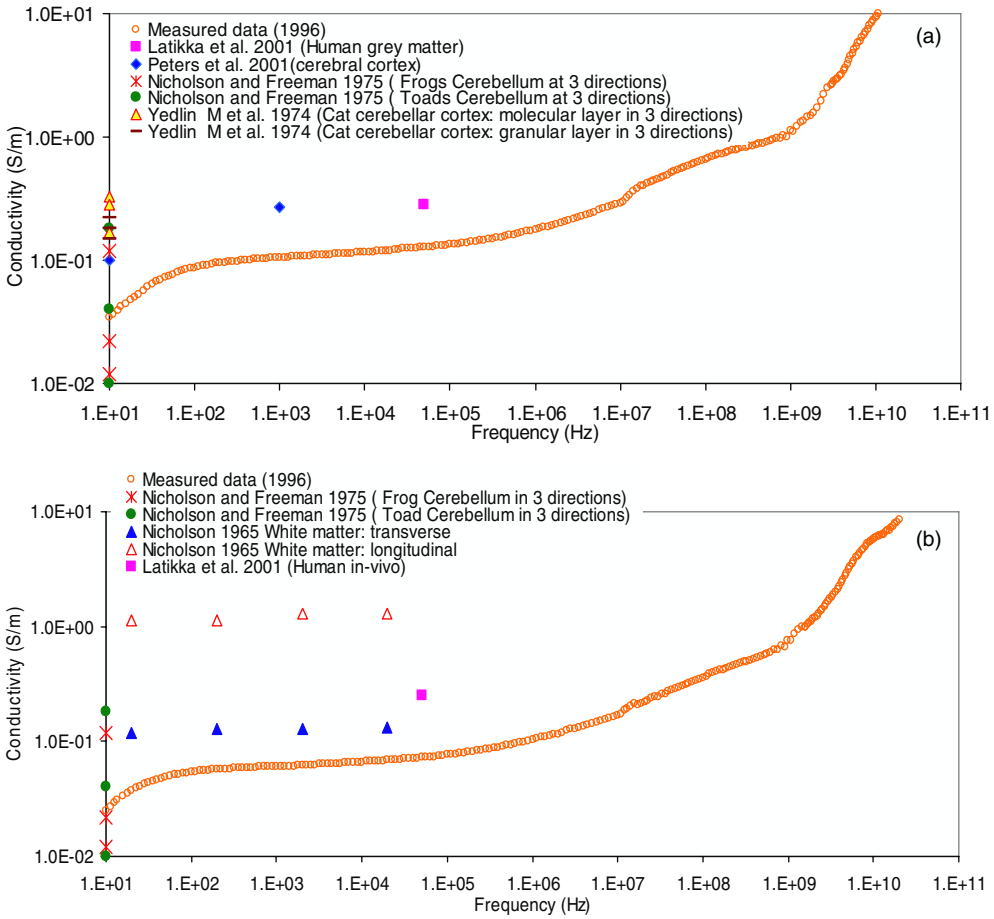

Figure 1. Conductivity of (a) grey and (b) white matter: data from the 1996 database and from the papers reviewed. Like the cerebrum, the cerebellum has a cerebellar cortex of grey matter and deep cerebellar white matter; where the region of the cerebellum is not specified, the data are included with the grey and white matter.

between the two; both are less isotropic than the white matter, be it in the cortex or in the dorsal column. Data pertaining to the grey and white matter are summarized in figures 1(a) and (b). Data by Peters *et al* (2001) were obtained by modelling the brain cortex as a suspension of randomly distributed elongated cells suspended in CSF; assuming the extracellular volume fraction to be between 0.18 and 0.28 and the conductivity of the CSF to range between 1.5 and 1.8 S m^{-1}, the effective conductivity of the mixture falls within the bounds 0.1 and 0.27.

In summary, the conductivity was obtained from impedance measurement, most studies assumed the tissue to be resistive (or, conversely, the capacitive element to be negligible). The most important finding is the anisotropy ratio of 10 for the conductivity of white matter. Grey matter is much more directionally uniform; ratios of between 2 and 5 observed in the cerebellum may well be due to its white matter content. The values in the 1996 database align closer to the lowest estimates; these data were obtained with coaxial probes where the field geometry does not allow directional sampling.

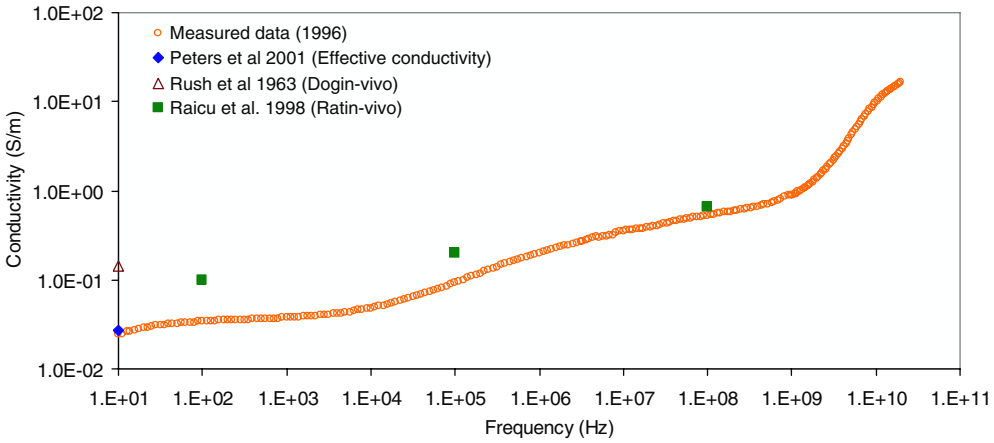

Figure 2. Conductivity of liver tissue: data from the 1996 database and from the papers reviewed.

Liver

Dielectric data for liver tissue were reported in several studies carried out under different conditions for a variety of reasons. For example, Riedel *et al* (2003) developed a contact-free inductive measurement procedure and demonstrated the system by carrying out conductivity measurements on liver tissue between 50 kHz and 400 kHz as a function of time after death. Hammerich *et al* (2002) reported changes in the electrical resistivity of liver tissue during induced ischaemia and postmortem. They observed increases in resistivity *in vivo* during occlusion. They analysed the data in terms of intra- and extracellular resistance and cell membrane capacitance.

Contributions by Raicu *et al* (1998a, 1998b) have provided data (permittivity and conductivity) for rat liver tissue, measured *in vivo*, in the frequency range of 10^2–10^8 Hz. The measured data were corrected for electrode polarization and found to be in reasonable agreement with some previous studies (Foster and Schwan 1996, Surowiec *et al* 1986).

Figure 2 shows Raicu *et al*'s (1998a, 1998b) data, data by Rush *et al* (1963) obtained by measurement, *in vivo*, on dogs and the effective conductivity of liver tissue calculated by Peters *et al* (2001). Peters assumed that the blood in the hepatic sinusoids of the liver has a conductivity of 0.65 S m^{-1} and that the hepatic cells are clustered forming plates organized radially around a central vein, and the cells were assumed non-conducting.

In summary, the values from the 1996 database align more closely with the lower estimates; however, we note that Raicu sample preparation involved flushing the surface with warm physiological saline, a practice that does not appreciably affect the spectral features except for the dc conductivity level of the conductivity spectrum (Raicu *et al* 1998b).

Muscle

Muscle tissues—skeletal, myocardial, lingual or other—exhibit large anisotropy in their electrical properties. This is due to the longitudinal muscle fibre structure and has been observed at frequencies below 1 MHz (Epstein and Foster 1983). Other aspects of the dielectric spectrum also depend on the relative direction of the fibre and the measuring field; the α dispersion is more prominent and the β dispersion is less defined in the longitudinal direction. This is also in accordance with the predictions of effective permittivity modelling

Table 1. Anisotropy of muscle tissue: conductivity in S m^{-1} along and across muscle fibres (Faes *et al* (1999) is a meta-analysis of review studies, the data are back-transformed mean values across studies and across frequencies in the range 100 Hz to 10 MHz; Rush *et al* (1963) and Epstein and Foster (1983) data are measured values obtained with tetrapolar array electrodes); Steendijk *et al* (1994) used a surface probe consisting of two perpendicular arrays of four platinum electrodes.

Reference	Perpendicular	Parallel	Ratio
Faes *et al* 1999	0.15	0.4	2.7
Rush *et al* 1963	0.04	0.6	15
Epstein and Foster 1983	0.079	0.52	6.6
Steendijk *et al* 1994	0.26	0.47	1.8
Peters *et al* 2001			10

of elongated structures predicted by Semenov *et al* (2002) and Peters *et al* (2001). The theory predicts an anisotropy ratio of the order of 10 for the static conductivity of the muscle tissue model. Practical difficulties in achieving field-fibre alignment means that anisotropy is not easy to observe experimentally (Fallert *et al* 1993, Tsai *et al* 2000); some of the problems associated with obtaining good data at frequencies below 1 MHz have been described by Tsai *et al* (2000, 2002) in the context of their *in vivo* measurement of swine myocardial resistivity. They did not observe anisotropy but report changes in the myocardial resistivity as a function of time after death.

Most recent studies on the electrical properties of muscle tissue focused on the differences between normal and ischaemic or hypoxic tissue because of their relevance to possible practical application in clinical diagnostics. There is also potential for non-invasive imaging, provided that the electrical characteristics of both normal and scar tissue are well defined. Schafer *et al* (1999) and Miyauchi *et al* (1999) observed changes in the α and β dispersions of normal and ischaemic skeletal muscle. Schwartzman *et al* (1999) investigated the region intermediate between healthy and infracted tissue in the case of chronically infracted ventricular myocardium. Data for normal myocardial tissue and muscle tissue anisotropy are summarized in figure 3 and table 1.

From recent data, we conclude that muscle tissue modelling is compatible with an anisotropy ratio of 10 for the effective conductivity of the medium; this, however, is not always observed experimentally. Figure 3 shows that the 1996 database align closer to the lower bound of more recent data; it is therefore reasonable to assume that the conductivity of muscle tissue at ELF falls within the bounds provided by the 1996 database and could be up to a factor of 10 or higher.

Skin

Skin is the interface of the body with environmental agents including electromagnetic fields; the knowledge of its dielectric properties is of importance in the assessment of human exposure and in numerous biomedical applications. Raicu *et al* (2000) carried out *in vivo* measurements on dry skin and on skin moistened with physiological saline, in the frequency range 100 Hz–100 MHz. They analysed the data using a suitable dispersion model. Comparing the parameters of the model for dry and saline-moistened skin, they note a fivefold increase in the dispersion magnitude, and one possible explanation they provide is that moistening increases the effective penetration depth such that contributions to the dispersion from the innermost skin layers become evident. This statement appears paradoxical; in fact, it is due to the reduction in the layering effect. When moistened, the skin appears more homogeneous

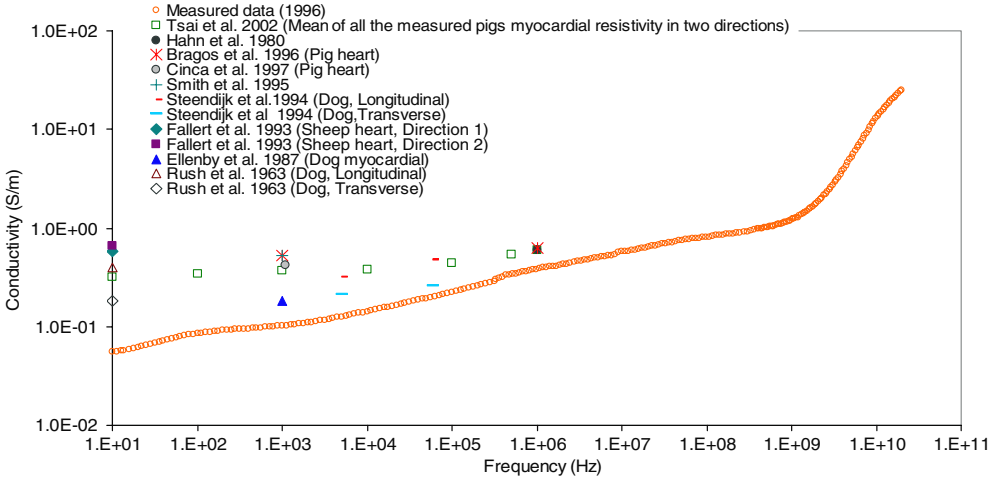

Figure 3. Conductivity of myocardial tissue: data from the 1996 database and from the papers reviewed. All measurements were carried out *in vivo* except for Ellenby *et al* (1987) that was measured *in vitro* at 37 °C.

and behaves like a high water content tissue. Raicu *et al* further speculate that a polarization phenomenon originating at the stratum corneum/epidermis interface occurs in dry skin, as suggested by Alanen *et al* (1999), but not when saline moistened. It therefore appears that topical measurement on dry (normal) skin *in vivo* may not be proportionately representative of the inner layers. On the other hand, the use of an aqueous coupling agent that is likely to hydrate the stratum corneum affects the results of the measurements. In practice, the use of a coupling agent gives more reproducible results and leads to better agreement between data from the recent literature (figures 4 and 5).

Marzec and Wachal (1999) measured the conductance and susceptance of soles and calves of leg skin in healthy controls and patients with ischaemia in the frequency range of 100 Hz to 100 kHz. Ischaemia was found to have no effect on the admittance at frequencies lower than 10 kHz where the effect of the stratum corneum is dominant. Observed differences at frequencies in excess of 10 kHz are ascribed to ischaemia in the underlying skin tissue.

Skin is the only tissue for which the dielectric data are available for both dry and moist conditions. Moistening provides good coupling to a topical measuring probe and reduces the effect of skin layers; however, it does affect the whole dielectric spectrum. The conductivity at ELF depends on the skin condition, the type of moistening agent—if used—and the site of the skin on the body.

Bone

Bone, cortical and cancellous, has anisotropic structure and, in consequence, anisotropic dielectric properties. For cortical bone, the ratio of the conductivity in the axial and radial dimensions is 3.2 (Reddy and Saha 1984). Cancellous bone is less anisotropic; the conductivity is higher in the longitudinal direction compared to the lateral and anterior–posterior directions (Saha and Williams 1989). More recent studies investigated the interrelationships of the electrical properties of cancellous bone with their mechanical characteristics, mass density and composition (Sierpowska *et al* 2003 and 2007). Sierpowska provides data for trabecular

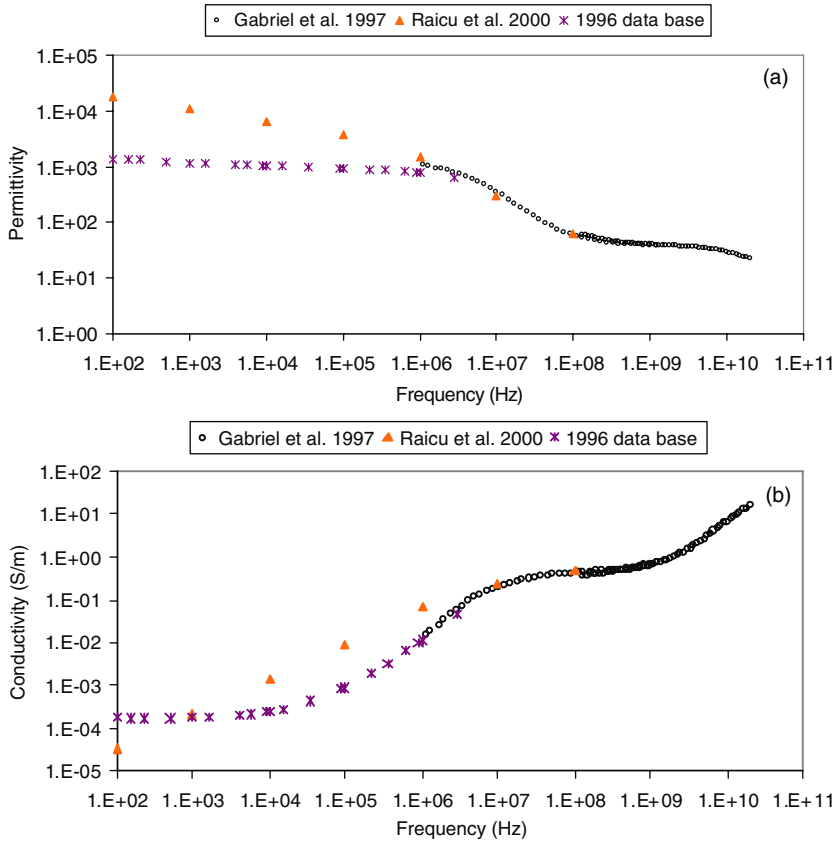

Figure 4. (a) Permittivity and (b) conductivity of skin (different parts of the body, excluding palms and soles). No moistening or contact gel was used.

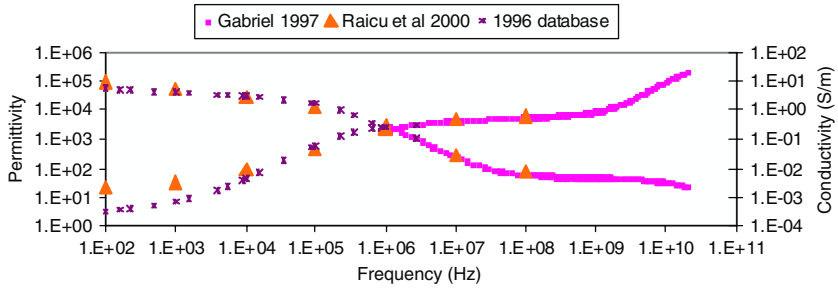

Figure 5. Permittivity and conductivity of skin (Raicu *et al* 2000: back of neck, moistened with physiological saline; Gabriel and database: ventral forearm, moistened with water). Open-ended coaxial probes of vastly different sizes were used.

(cancellous) bone from bovine femur ranging from 0.02 to 0.06 S m^{-1} depending on the site. This compares with 0.07 S m^{-1} for ovine cancellous bone in the database.

The structure and composition of bone vary with age, so do the dielectric properties (Peyman *et al* 2007a) in line with the observation of Sierpowska *et al* (2003, 2007) about the interrelationship of these properties.

Experimental study: dielectric properties of porcine tissue *in vivo*

This study was designed to obtain data at frequencies below 1 MHz to supplement the 1996 database with specific emphasis on avoidance of electrode polarization and detection of anisotropy in dielectric properties. The approach is to use linear four terminal probes to carry out impedance measurements on pig tissue, *in vivo*.

Material and methods

Pigs weighing around 50 kg were used in this study. The animals were sedated with 40 mg midazolan hydrochloride ('Hypnovel') (5 mg per 1 ml solution) delivered via intra-muscular injection and then anaesthetized using gaseous anaesthesia. This consists of halothane with nitrous oxide and oxygen as the carrier gases. The ratio of concentrations of nitrous oxide to oxygen was 2:1. Isofluorane would be used if the animal showed signs of cardiac arrhythmias. In all cases, the animals were being mechanically ventilated using a Manley Blease MP3 ventilator. The animal then had an arterial catheter surgically implanted. This was in the carotid artery or femoral artery depending on the position the animal would be in during the experiment. The arterial line allowed us to measure arterial pressure during the experiment and to take arterial blood for the analysis of blood gases. In addition, a venous catheter was inserted to allow infusion of 0.9% saline (2 ml kg^{-1} h^{-1}). After the arterial line was inserted, an additional intra-muscular injection of buprenorphine was administered. A terminal overdose of pentobarbital (a barbiturate) was administered to the animal after the *in vivo* measurements were completed. The treatment of the animals was in compliance with animal welfare regulations in the UK. The operating theatre was climate controlled for temperature and humidity. The core temperature of each pig was measured either rectal or oesophageal with a thermometer connected to a Propaq 106 (Protocol Systems, Inc.) medical data-recording instrument. It varied between animals and the mean value was about 38.14 °C (±0.56). The temperature of each tissue was recorded prior to the measurements; on average it was 36.76 °C (±0.88). This discrepancy is because the operators allowed cooling or warming with bags of sealed saline to get as close to 37 °C as possible and also some tissues cool more rapidly than others do.

The dielectric properties were obtained from complex admittance data (capacitance C and conductance G) measured using a four terminal probe and Solartron 1260 Frequency Response Analyser at eight frequencies in the range of 10 Hz to 1 MHz; no measurements were made at 50 Hz to avoid mains pickup effects. The electrodes were made from 1 mm thick, 4 mm wide titanium plates, the top 1 mm of each plate was wedged at 26° from the bisecting plane and all corners were rounded prior to platinum plating. The electrodes were mounted 1 mm apart, and thus, the sensing edges were 2 mm distant from each other. The outer electrodes were connected to the current output of the analyser and the inner electrodes to the voltage sensing ports via 1 kΩ resistors welded to the plates. All but the wedged ends of the plates were encased in a resin sleeve to obtain a sturdy probe suitable for *in vivo* sampling.

Fibrous tissue such as muscle could be cut along or across the fibre; in a longitudinal cut, the probe could be positioned such as the electrodes align with the fibre (0° position) or are at right angle to the fibres (90° position). In a cross-sectional direction, the cut fibres present radial symmetry and the relative fibre/electrodes' orientation does not depend on the positioning; this is referred to as the transverse direction.

Sampling requires the probe to be held firmly on the tissue being measured; body fluids were sampled *in vitro* immediately after extraction by immersing the probe in the fluid.

The cell constant, that is, the parameter that relates the measured capacitance and conductance to the permittivity and conductivity of the sample, was obtained experimentally

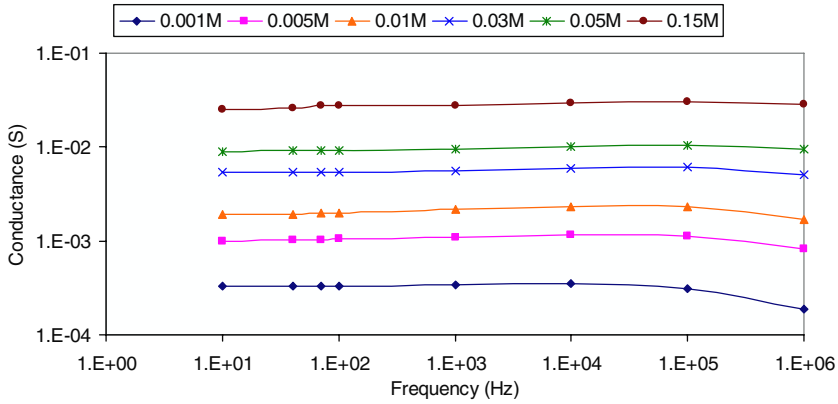

Figure 6. Conductance (S) of NaCl solutions used to calculate the cell constant.

using the conductance of five NaCl solutions ranging from 0.001 to 0.15 M (figure 6). Theoretically, ionic solutions such as aqueous NaCl exhibit no dielectric dispersion at frequencies below 1 MHz; the conductivity should not vary with frequency. In practice, electrode polarization and, if present, other experimental artefacts might introduce frequency-dependent variations. The data shown in figure 6 are fairly frequency independent such that, for each concentration, the value at any frequency (or the average value across the frequency range) varies with the ionic conductivity (and hence the concentration) of each solution. The ionic conductivities for the NaCl solutions at 20 °C (the measurement temperature) were obtained from Peyman *et al* (2007b). For the data shown in figure 6, the linear relationship between the measured conductances (at all frequencies and concentrations) and the corresponding conductivities has a correlation coefficient of 0.9932 and the factor relating the two parameters was found to be 0.020 m. The conductivity (S m^{-1}) is obtained by dividing the measured conductance (S) by the cell constant (m).

The corresponding capacitance spectra present a less coherent picture; the data will be reported in the following section.

Results

The capacitance spectra of the NaCl solutions were affected by electrode polarization, which, to some extent, is to be expected even with four terminal measurements. There was also an unexpected effect at the higher end of the measurement frequency range that gave rise to some negative capacitance values; the effect appeared systematic and concentration dependent. We investigated the possibility that it could be due to a stray inductive element but could not identify a source. Also investigated is the possibility that it could be a phase inversion effect, but this could not be clearly demonstrated from the impedance data. For this reason, it was thought prudent to limit the measurement to the conductance where the effects are small.

Measurement uncertainty

In the characterization of the electrical properties of biological tissue, heterogeneity and sample–sample variability are major contributors to the uncertainty. However, before confining the uncertainty to sample variability, it is necessary to quantify the systematic uncertainty to ensure that it is corrected or accounted for in the uncertainty budget. This was done using data for NaCl solutions (figure 6).

Table 2. G (S) is the average conductance from 10 Hz to 1 MHz; the conductivity (S m^{-1}) is calculated from the average conductance and the cell constant; the literature values are from Peyman *et al* (2007b).

NaCl concentration (M)	G (S) average	G (S) SD	G (S) % SD	σ (S m^{-1}) average	σ (S m^{-1}) Peyman *et al* (2007b)	σ (S m^{-1}) % difference
0.001	3.14×10^{-4}	5.30×10^{-5}	16.85	0.016	0.009	−67.15
0.005	1.03×10^{-3}	1.06×10^{-4}	10.23	0.052	0.047	−10.05
0.01	2.03×10^{-3}	2.05×10^{-4}	10.13	0.101	0.094	−7.88
0.03	5.52×10^{-3}	3.32×10^{-4}	6.01	0.276	0.281	1.62
0.05	9.53×10^{-3}	5.94×10^{-4}	6.23	0.477	0.466	−2.21
0.15	2.78×10^{-2}	1.74×10^{-3}	6.27	1.392	1.375	−1.23

Table 3. For NaCl solutions 0.005–0.15 M: at each frequency, the average percentage difference between measured and literature conductivity values.

Frequency (Hz)	σ (S m^{-1}) % difference
10 00 000	5.9
100 000	−15.1
10 000	−13.2
1000	−7.1
100	−2.8
70	−1.8
40	0.2
10	2.3

If free of systematic errors, there should be no variation in the measured conductance as a function of frequency. In theory, we can calculate the ionic conductivity of each of the NaCl solutions from the measured conductance and the cell constant. Table 2 shows the extent of the deviation of the measurement from this ideal behaviour. We have also looked at the error in conductivity as a function of frequency for all concentrations (table 3).

From tables 2 and 3, excluding the data for 0.001 M, which are not relevant to the range of conductivity of body tissue and fluids, we find systematic errors of up to 10%. As expected, the errors are not equal at all frequencies. These systematic errors are of the same order of magnitude as the random errors which, expressed as the standard deviation of all samples measured, range from 10% to 25% depending on tissue type, except in the case of lung (deflated) where variations of up to 60% were observed.

To eliminate or significantly reduce systematic errors, data for the biological samples were normalized to NaCl solutions of similar conductivity. Random variations are shown as error bars in figures 7 and 8.

Conductivity of tissue

Data for each tissue are average values from measurement on samples from at least three pigs. The conductivity spectra of body tissues and body fluids are shown graphically in figures 7 and 8, respectively; the conductivities at 40 and 70 Hz are given in table 4. Where comparison with the 1996 database and the data reviewed in this paper is not available, reference will be made to the conductivity of tissues in Duck (1990) and Grimmes and Martinsen (2008). Duck reports low-frequency conductivity data (dc to 10 kHz) gathered from the literature from both

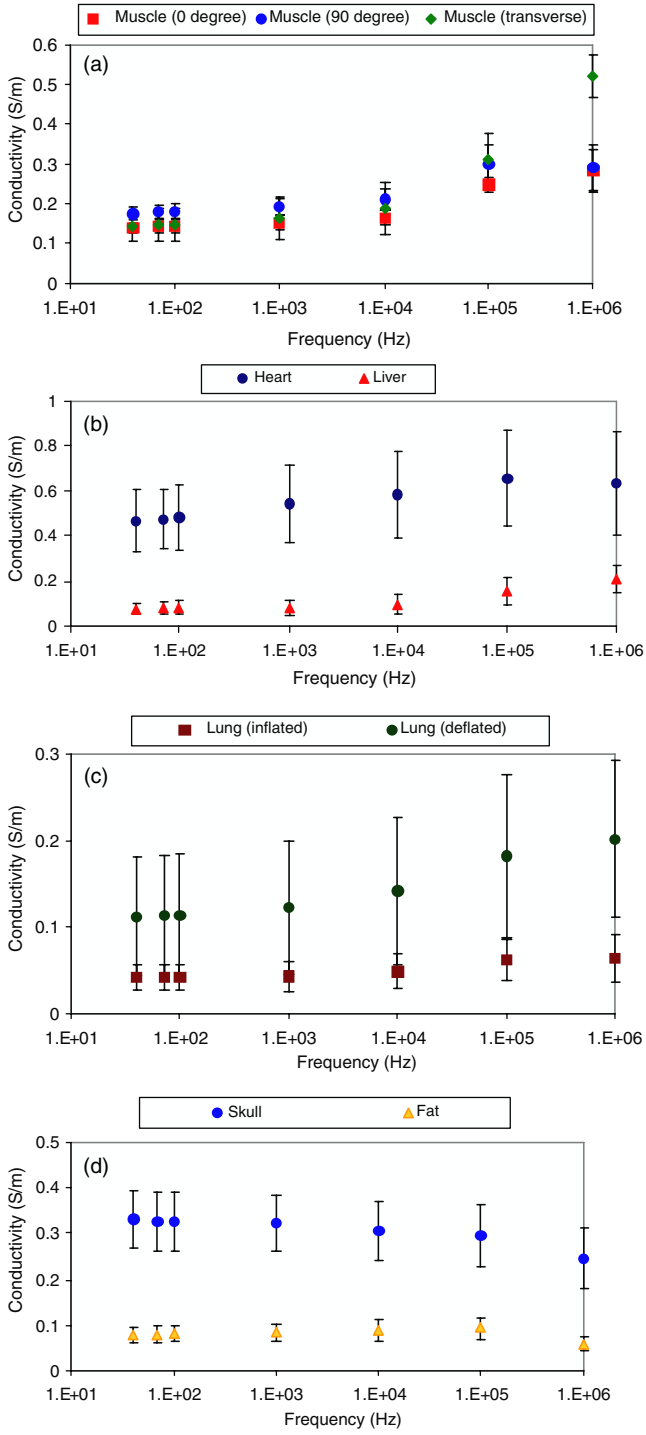

Figure 7. The conductivity spectra of pig (a) muscle at different directions, (b) heart and liver, (c) lung tissue at inflated and deflated states and (d) skull and fat tissues. The error bars represent the standard deviation of measurements on samples from different pigs.

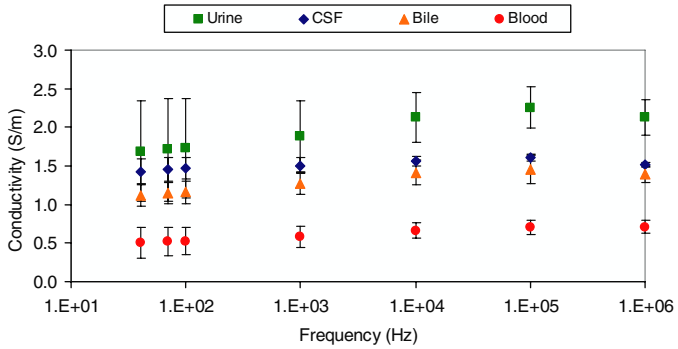

Figure 8. The conductivity spectra of pig body fluids. The error bars represent the standard deviation of measurements between three and six different pigs.

Table 4. Measured conductivity (S m^{-1}) of various tissues at 40 and 70 Hz.

Tissue	40–70 Hz
Muscle (0°)	0.15 ± 0.014
Muscle (90°)	0.19 ± 0.018
Muscle (transverse)	0.16 ± 0.037
Heart (atrium)	0.48 ± 0.13
Skull (mid-section)	0.32 ± 0.38
Fat	0.078 ± 0.019
Lung (inflated)	0.042 ± 0.014
Lung (deflated)	0.11 ± 0.069
Liver	0.091 ± 0.024
Urine	1.87 ± 0.69
CSF	1.59 ± 0.18
Bile	1.27 ± 0.15
Blood	0.60 ± 0.21

in vivo and *in vitro* measurements (mostly at 37 °C). Grimmes and Martinsen (2008) report conductivity data in the range 1 Hz to 10 kHz.

Skeletal and myocardial muscle, liver, lung, fat and bone were measured *in vivo* and blood, bile, cerebrospinal fluid and urine *in vitro* at 37 °C.

Skeletal muscle. Measurements have been carried out in three mutually perpendicular directions but we did not observe the expected anisotropy in electrical conductivity (figure 7(a)). One possible explanation is the lack of rigorous correspondence between the geometry of the probe and the structure of the tissue. Values are lower than the 1996 database but within the range of values (0.05–0.4 S m^{-1}) reported in Grimmes and Martinsen (2008).

Myocardial muscle. Data for myocardial tissue fall close to the literature data reviewed in figure 3.

Skull. The measurements were made on the mid-sectional part of the skull of 50 kg pigs: this part of the skull has flat areas which allow good contact with the electrodes. The bone exposed after stripping the periostium membrane appeared pitted and filled with red bone marrow. The

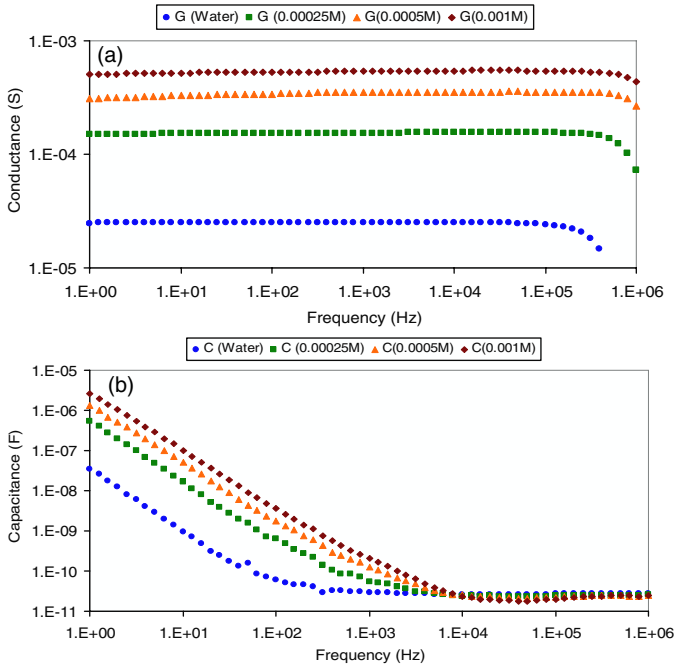

Figure 9. The measured (a) conductance and (b) capacitance values of water and low concentration NaCl.

measured conductivities (figure 7(d)) are higher than all previously reported data for bone. This property appears peculiar to the skull bone of young pigs (Peyman *et al* 2009). The skull bone of 250 kg pigs (mature animals) has a hard consistency and much lower conductivities (Peyman *et al* 2009).

Fat. The conductivity across the frequency range is 0.08 S m^{-1}; this value falls on the high side of the range of values reported in the 1996 database.

Liver. The standard deviation is of the order of 30%; the average aligns rather well with the Raicu *et al* (1998a, 1998b) data reviewed in figure 2.

Lung (inflated and deflated). Standard deviation of 30% and 60% respectively but the range of values falls well within the data reported by Duck (1990) and Grimmes and Martinsen (2008); the conductivity of lung deflated is about three times higher than when inflated, and this ratio is also in good agreement with data reviewed by Duck (1990).

Conductivity of body fluids. The measured conductivities of blood, bile and CSF are in good agreement with data reported in Duck (1990); there is a discrepancy in the case of urine, the measured conductivity is of the order of 2 S m^{-1}, and this compares with 3.3 S m^{-1} (Duck 1990) and 0.5–2.6 S m^{-1} (Grimmes and Martinsen 2008). The values are shown in table 4.

Finally, we have carried out measurements using a probe comprising a rectilinear array of four platinum-blacked platinum pin electrodes embedded in PTFE. This probe produced a coherent set of capacitance and conductance values, down to 1 Hz, when used to measure water and low concentration salt solutions (figure 9).

Electrode polarization effect manifests itself as an increase in capacitance at low frequencies; the polarization capacitance is proportional to the frequency raised to the power of −1.46. If we assume the relationship to be valid over a wide frequency range, the contribution of the electrode polarization can be subtracted from the measured capacitance, which is one way of accounting for this phenomenon. There are some high-frequency effects that have yet to be fully identified and resolved through further investigation; modelling of the impedance and its dependence on probe design is one way forward.

Conclusion

This study has produced new tissue conductivity data to complement the literature reviewed in this paper. Preliminary data for salt solutions obtained using a rectilinear array of platinum-blacked platinum electrodes show capacitance and conductance spectra consistent with a resistive system and a well-define electrode polarization element. Further work is needed to correct these shortcomings, but the regularity and reproducibility of the data augur well for the possibility of avoidance or correction of the unwanted effects and hence for the opportunity of making error-free permittivity and conductivity measurement in future.

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