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Estimation method for the anisotropic electrical conductivity of in vivo human muscles and fat between 10 kHz and 1 MHz

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Keywords: tissue conductivity, tissue electrical properties, anisotropic muscle, fat, bioimpedance measurement, low frequency, electromagnetic field exposure

Abstract

Objective. In low frequency dosimetry the variability in the electrical conductivity values assigned to body model tissues represents a major source of uncertainty. The aim of this study is to propose a method for estimating the conductivity of human anisotropic skeletal muscle and fat in vivo in the frequency range from 10 kHz to 1 MHz. Approach. A method based on bounded electrical impedance tomography was used. Bioimpedance measurements were performed on the legs of ten subjects. Anatomically realistic models of the legs were then created using magnetic resonance images. The inverse problem of the tissue conductivities was solved using the finite element method. The results were validated using resampling techniques. These findings were also used to study the effects of muscle anisotropy on magnetic field exposure. Main results. The estimated conductivities for anisotropic muscle were found to be in good agreement with values found in existing literature and the anisotropy was shown to decrease with increasing frequency, with the ratio of lateral to longitudinal conductivity increasing from 37% to 64%. The conductivity of fat was found to be almost a constant 0.07 S m−1 in the frequency range considered. Significance. The proposed method was shown to be a viable option when estimating in vivo conductivity of human tissue. The results can be used in numerical dosimetry calculations or as limits in future investigations studying conductivity with bioimpedance measurements.

1. Introduction

In the low frequency (LF) range, from 1 Hz to 10 MHz, time varying electromagnetic fields can excite nerve and muscle cells. The International Commission on Non-Ionizing Radiation Protection (ICNIRP) has published guidelines to avoid these adverse health effects (ICNIRP 2010). In these guidelines frequency dependent basic restrictions have been defined to limit the internal electric field strength in the exposed tissue. Because the internal electric field strength cannot be measured directly, ICNIRP has defined reference levels for a more practical form of exposure assessment. These reference levels have been derived using numerical dosimetry calculations on anatomically realistic models (Nagaoka et al 2004, Dimbylow 2005, Bahr et al 2007, Hirata et al 2009, ICNIRP 2010, 2020b). A problem arises since the electrical conductivity values assigned to the model tissues represent a significant source of uncertainty (Soldati and Laakso 2020). This has been highlighted in a gap in knowledge document (ICNIRP 2020a) which aims to assist ICNIRP in further developing their LF guidelines. In the document, need for the investigation of anisotropic tissue properties was also accentuated. The same concerns were also noted in a research agenda of the Institute of Electrical and Electronics Engineers International Committee on Electromagnetic Safety (IEEE ICES) (Reilly and Hirata 2016).

Arguably the most widely used work on tissue dielectric properties is the investigation reported by Gabriel et al (1996a, 1996b, 1996c). Their work characterizes over 30 types of tissue according to their dielectric
properties from 10 Hz to 20 GHz. The authors note that at low frequencies below 1 MHz, their data provides only a ‘best estimate’ of the permittivity and conductivity of the tissues. In order to address this lack of knowledge Gabriel et al (2009) reported a review and measurements of conductivities for 10 tissues at frequencies below 1 MHz.

Skeletal muscle is electrically anisotropic, with higher conductivity along the muscle fibres than across them (Epstein and Foster 1983, Dawson and Stuchly 1998, Hart et al 1999). The effects of muscle anisotropy are most prominent at lower frequencies and disappear when approaching the MHz range. This has also been shown in more recent works (Ahad et al 2009, Sanchez et al 2014, Nagy et al 2019), which have focused on linking the dielectric properties of muscle to the assessment of various neuromuscular diseases. In the LF range, the dielectric properties of fat and subcutaneous tissue have also been studied (Smith and Foster 1985, Wake et al 2016, Hershkovich et al 2019). Results show large variability between studies, which may be due to different origins of the samples. Also, the variability in sample water content has been shown to influence the conductivity of low water content tissues (Smith and Foster 1985, Wake et al 2016). Moreover, the majority of the aforementioned studies have conducted their measurements on excised tissue samples of animal origin. Excised samples might not resemble the normal physiological state of the tissue, since necrotic tissues have been shown to vary in dielectric properties compared to healthy in vivo tissues. This is likely due to changes in cell membrane integrity and extracellular water content which change as a function of time (Pethig 1987, Martinsen et al 2000, Grimnes and Martinsen 2010). In addition, the lack of standardized measurement methods and sample handling protocols, may also have an effect on the large variability in reported results.

To measure the conductivity of human tissue in its normal functioning state, and to avoid invasive measurement techniques, methods utilizing electrical impedance tomography (EIT) have been developed. In EIT, an electric current is applied on the boundary of a volume and the resulting potentials are measured from the surface. These measurements can then be used to estimate the electrical conductivity distribution in the interior of the volume. EIT struggles from poor spatial resolution (Bayford 2006), and thus, these methods usually use different kinds of volume conductor models, ranging from simple three-dimensional geometries to anatomically realistic, in order to simplify the reconstruction process. The following simplification of the inverse problem is known as bounded EIT (bEIT). bEIT has been mainly used in biophysics research regarding the human brain, specifically to study the brain-to-skull conductivity ratio (Oostendorp et al 2000, Gonçalves et al 2003, Clerc et al 2005, Lai et al 2005, Zhang et al 2006, Fernández-Corazza et al 2013, Ouypornkochagorn et al 2014, Dabek et al 2015, Fernández-Corazza et al 2016, Essaki Arumugam et al 2017). These studies rely roughly on the same approach: the conductivities, of the tissues of interest, are solved by varying the conductivities in the models, so that the difference between the measurements and the model estimates are minimized. The potential distribution on the model surface can be solved with various methods including: the boundary element method (BEM) (Oostendorp et al 2000, Gonçalves et al 2003, Clerc et al 2005, Dabek et al 2015), the finite element method (FEM) (Ohmine et al 2004, Zhang et al 2006, Fernández-Corazza et al 2013, 2016) or the finite difference method (FDM) (Aaron et al 1997, Hart et al 1999, Essaki Arumugam et al 2017).

The aim of this work is to propose a method which can be used to estimate human in vivo tissue conductivities non-invasively. The proposed method also allows for the characterization of anisotropic tissues. The results of this work can be used in low frequency numerical dosimetry to better model the electrical properties of intact human tissues. The human leg was chosen as the area of interest due to its relatively simple structure and low number of different tissues. The impedance was measured from different areas of the leg. Magnetic resonance imaging (MRI) was used to produce accurate images of the subjects’ legs which were then used to create anatomically realistic 3D-models of the legs. An in-house solver based on FEM was used to model the impedance. By minimizing the error between modelled and measured impedances the conductivity of fat and anisotropic muscle could be estimated. The newly derived conductivity estimates were also used to study the effect of muscle anisotropy on exposure to homogeneous magnetic fields. The strengths of induced electric fields were compared to the basic restrictions set by ICNIRP.

2. Materials and methods

2.1. Study participants

Ten healthy subjects, six male and four female, volunteered in this study. The ages ranged from 24 to 56, with a mean age of 36 and standard deviation of 12. This study was approved by the Aalto University Research Ethics Committee. The experimental procedure was described in an accessible document given to the subjects and written consent was obtained from each subject.
2.2. Impedance measurements

The electrical impedance of the test subject’s right leg was measured with a Keysight E4990A impedance analyzer (Keysight Technologies, Santa Rosa, CA). The impedance analyzer was set to apply a current stimulus of 1 mA and the impedance was measured with a tetrapolar electrode configuration from 5 kHz to 1 MHz, at 128 logarithmically spaced points. To interface the impedance analyzer with the subject, standard electrocardiography (ECG) electrodes, Ambu blue sensor 2300 (25 mm × 23 mm) (Ambu A/S, Ballerup, Denmark), were used. Prior to applying the electrodes, the skin was lightly abraded and cleaned with skin-cleaning swabs. The measurements were performed at room temperature in two perpendicular directions: along the leg, longitudinal direction, and across the leg, lateral direction. In the longitudinal direction the current injection electrodes were placed on the right wrist and ankle and the impedance was measured from five sections along the leg, each approximately 5 cm long, with the lower part of the patella approximately at the middle of the measured area. In the lateral direction the impedance was measured from three sections; thigh, knee and calf, with the electrode pairs on opposite sides of the leg. An example longitudinal measurement can be seen in figure 1 (1), with the subject lying on the MRI table.

2.3. Hook artefact correction

Unfortunately, when measuring bioimpedance from large volume conductors, parasitic capacitances from the body to ground can provide an additional path for the injected current used to calculate the impedance. This results in a measurement error and can be observed as ‘hook’ like artefact on an impedance plot (Buendía et al 2010a, 2010b, Aliau-Bonet and Pallas-Areny 2012). In addition to parasitic capacitance, the Keysight E4990A impedance analyzer cannot measure bioimpedance in the presence of large additional impedances exhibited by the skin-electrode interface due to electrode polarization (Gómez-Sánchez and Felice 2012). Fu and Freeborn (2018) showed that when measuring bioimpedance with the E4990A, these additional impedances and parasitic components limit the measurement frequency band from 10 to 100 kHz. The lower end of the frequency band is limited by the Keysight E4990A’s auto-balancing bridge, which is unable to balance, resulting in unstable or undefined measurements. The upper range of the frequency band was extended to 1 MHz with a hook artefact correction procedure described by Freeborn et al (2018), which is specifically developed to extend the measuring capabilities of the Keysight E4990A without the addition of a custom front-end amplifier.
The majority of the measurements exhibited a significant hook artefact in the high frequency data, with slight deviations in magnitude. The hook artefact correction procedure was applied to all bioimpedance measurements, 50 in the longitudinal and 30 in the lateral direction, resulting in impedance data at 11 logarithmically spaced frequencies from 10 kHz to 1 MHz. As the reader might note, the measurements from 5 to 10 kHz were discarded as an additional precautionary measure. Figure 2(a) shows an example measurement corrected with the method. It is easy to see that the errors induced by the measurement setup are most prominent at frequencies above 100 kHz. This is also illustrated in figure 2(b) where the real part of the complex impedance, resistance, is plotted.

2.4. Imaging and tissue segmentation
Immediately after the impedance measurements, the ECG electrodes were replaced with fish oil capsules, so the electrodes themselves could be modelled. This can be seen from figure 1(2). Magnetic resonance images (MRI) of the leg were acquired using a 3 T scanner (Magnetom Skyr; Siemens, Ltd Erlangen, Germany). To obtain good contrast between different tissues both T1- and T2-weighted images (turbo spin-echo sequences and a spatial resolution of 2.1 mm × 2.1 mm × 2.1 mm in both cases) were acquired. The resulting images were then segmented into eight different types: blood, cancellous bone, cortical bone, cartilage, fat, muscle, skin, and tendon. The segmentation was performed with the MRI spatial resolution. In the following, muscle and tendon were combined into a single anisotropic tissue, having two distinct conductivity values along and across the longitudinal axis of the leg models. They are referred to in this work as longitudinal and lateral muscle conductivities. Bone marrow was segmented as cancellous bone, since white and red marrow could not be told apart in the MR-images. The skin was added to the models as a single voxel layer after importing the models to MATLAB.

2.5. Volume conductor models
The segmented images were imported into MATLAB where they were voxelized using cubical elements, with a spatial resolution of 2.1 mm × 2.1 mm × 2.1 mm. There were three unknown frequency-dependent tissue conductivities: longitudinal and lateral muscle, and fat. Other tissues were assigned fixed conductivities as listed in table 1. The mean tissue conductivity values were derived from the literature at 100 kHz and were assumed frequency independent and isotropic. All conductivity values were considered to be ‘effective’, meaning that separate tissue compartments were regarded as homogeneous (Peters et al 2005).

2.6. Electric field modelling
Under the quasi-static approximation (Wang and Eisenberg 1994), valid in the frequency range considered (below 1 MHz), the electric scalar potential \( \phi \) satisfies:

\[
\nabla \cdot \mathbf{\varepsilon} \cdot \nabla \phi = 0, \quad \text{in } \Omega,
\]

with boundary conditions

\[
\begin{align*}
\hat{n} \cdot \nabla \phi &= 0, & \text{on } \delta \Omega_N, \\
\phi &= V_i, & \text{on } \delta \Omega_i, \quad i = 1, 2
\end{align*}
\]
where $\Omega$ is the domain (the leg models) and $\delta\Omega_N$ is the boundary of the domain (the surface of the models excluding the electrodes), $\delta\Omega_i$ are the electrodes, $V_i$ are the potential values of each electrode, and $\bar{\sigma}$ is the conductivity.

In the longitudinal direction, with the current injection electrodes placed on the wrist and ankle, the source and sink of the electric current were modelled as orthogonal planes with respect to the longitudinal axis of the leg models. In the lateral direction the current injection electrode geometry was modelled. An example of current flow from both cases is illustrated in figure 1 (5).

The scalar potential equation (1) with boundary conditions (2) was solved numerically with an in-house solver based on FEM (Laakso and Hirata 2012) with first-order cubical elements.

### 2.7. Conductivity estimation

The impedance between two voltage sensing electrodes can be modelled if the scalar potential inside the volume conductors is known. For example, the impedance between points $r_p$ and $r_m$, can be calculated by dividing the potential difference with the current:

$$Z_{\text{model}} = \frac{|\phi(r_p) - \phi(r_m)|}{I},$$

(3)

where the electric current, $I$, can be derived by integrating the current density ($f = -\bar{\sigma} \cdot \nabla \phi$) over a surface between the current injection electrodes.

To estimate the unknown conductivities of fat and longitudinal and lateral muscle, the least-squares method was used to find the optimal values for the conductivities that minimize the sum of squared residuals between the modelled and measured impedances:

$$\sigma_{\text{est}}(f) = \arg\min_{\sigma} \sum_i (Z_{\text{model}}(\sigma, f, l) - Z_{\text{meas}}(f, l))^2,$$

(4)

where the sum is taken over all measurements (eight measurements in ten subjects), $f$ is the frequency and $\sigma_{\text{est}}$ is a three component vector containing the estimated conductivity values of fat and longitudinal and lateral muscle.

To solve the optimization problem (4), we employed a ‘brute force’ approach. The impedances $Z_{\text{model}}$ were calculated for all possible combinations of the unknown conductivities. For both longitudinal and lateral directions, the muscle conductivity was varied from 0.1 to 1 $\text{S m}^{-1}$. The conductivity of fat was assumed isotropic and varied from 0.01 to 0.13 $\text{S m}^{-1}$. In all cases a resolution of 0.02 $\text{S/m}$ was used, resulting in 14 812 sets of impedances modelled with different conductivities. The modelled impedances were then interpolated to a finer grid, to attain a resolution of 0.01 $\text{S m}^{-1}$.

### Table 1. Constant tissue conductivities assigned to the leg models.

<table>
<thead>
<tr>
<th>Tissue</th>
<th>Source</th>
<th>Conductivity ($\text{S m}^{-1}$)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Value at 100 kHz</td>
</tr>
<tr>
<td>Blood</td>
<td>Schwan (1956)</td>
<td>0.68</td>
</tr>
<tr>
<td></td>
<td>Gabriel et al (2009)</td>
<td>0.7</td>
</tr>
<tr>
<td>Bone (Cancellous)</td>
<td>Smith and Foster (1985)</td>
<td>0.21</td>
</tr>
<tr>
<td></td>
<td>Williams and Saha (1996)</td>
<td>0.2</td>
</tr>
<tr>
<td></td>
<td>Gabriel et al (1996b)</td>
<td>0.004</td>
</tr>
<tr>
<td>Bone (Cortical)</td>
<td>Reddy and Saha (1984)</td>
<td>0.002</td>
</tr>
<tr>
<td></td>
<td>Smith and Foster (1985)</td>
<td>0.01</td>
</tr>
<tr>
<td></td>
<td>Kosterich et al (1983)</td>
<td>0.0173</td>
</tr>
<tr>
<td></td>
<td>Williams and Saha (1996)</td>
<td>0.069</td>
</tr>
<tr>
<td></td>
<td>Unal et al (2018)</td>
<td>0.0003</td>
</tr>
<tr>
<td>Cartilage</td>
<td>Binette et al (2004)</td>
<td>1.14</td>
</tr>
<tr>
<td></td>
<td></td>
<td>0.88</td>
</tr>
<tr>
<td>Skin</td>
<td>Wake et al (2016)</td>
<td>0.434</td>
</tr>
<tr>
<td>Electrode gel (0.9% saline)</td>
<td></td>
<td>1.6</td>
</tr>
</tbody>
</table>

* Values were linearly extrapolated from 1 to 100 kHz.

* Only the dermis was considered.
In order to study the effects of muscle anisotropy, this method was also used to calculate an isotropic muscle conductivity for comparison. This was done by choosing only the impedances with the same muscle conductivity in both directions. For additional comparison, a simplified model having a homogeneous isotropic conductivity was also considered. In this case, the solution of (4) was trivial.

The statistics and robustness of these estimates was evaluated using the Bootstrap resampling method. The original dataset was resampled 100 000 times with replacement to reveal a precise estimate of the Bootstrap distribution (Putth et al 2015). The 95% confidence intervals for the estimated conductivities could then be obtained from the Bootstrap distribution. A schematic diagram of the entire conductivity estimation process is shown in figure 1.

To evaluate how well the modelled impedance corresponds to the measured impedance, i.e. goodness of fit, the mean relative error (MRE) was calculated:

\[
MRE(f) = \frac{1}{n} \sum_{i=1}^{n} \left| 1 - \frac{Z_{\text{model}}(\sigma_{\text{est}}(f), l)}{Z_{\text{meas}}(f, l)} \right|
\]

where again the sum is taken over all measurements and \( n \) is the number of measurements.

2.8. Computational dosimetry of magnetic field exposure

In order to study the effects of muscle anisotropy and frequency dependent conductivity values on the induced electric fields, the leg models were exposed to a spatially uniform magnetic field, the frequency of which was varied from 10 kHz and 1 MHz at 11 logarithmically spaced frequencies. The magnitude of the magnetic field was kept constant at 0.1 mT, which is the reference level for occupational exposure defined by the ICNIRP (2010) guidelines. At each distinct frequency the magnetic field direction was varied in AP (front-to-back), LAT (side to side) and TOP (top-to-bottom) orientations. Due to the spatial resolution of the models (2.1 mm × 2.1 mm × 2.1 mm), no spatial averaging of the induced electric field was performed. However, the effect of numerical artefacts was reduced by using the 99th percentile value of the induced electric field strength calculated in each tissue compartment (ICNIRP 2010). The calculation of the induced electric field used the same FEM model as the impedance calculation, except that the source term was calculated from the magnetic vector potential (Wang and Eisenberg 1994).

3. Results

3.1. Leg models

The leg models represent the human legs as simplified volumes of eight different types: blood, cancellous bone, cortical bone, cartilage, fat, muscle, skin and tendon. Figure 3 shows example sagittal slices from thigh, knee and calf of how the T1- and T2-weighted MR-images were segmented into different tissues. Table 2 provides the mean volume fraction and standard deviation of each tissue segmented from the MR-images.

3.2. Estimated conductivities

Figure 4(a) shows the conductivity values estimated for anisotropic muscle and fat. In figure 4(b) the estimates for isotropic muscle and fat are shown. With the muscle modelled isotropic the corresponding fat conductivities are slightly lower of the anisotropic case, but still lie within the same range. A homogeneous conductivity was also calculated from the dataset, shown in figure 4(c). All line plots show the minimum estimates using the original dataset and the coloured areas show the 95% Bootstrap confidence intervals.

The goodness of fit was evaluated by calculating the MRE between the modelled and measured impedances (equation (5)) with the different estimated conductivity sets. The results are shown in figure 4(d), where the coloured area shows the standard error of the mean. By modelling muscle as anisotropic the MRE is almost a constant 14%. When using isotropic muscle conductivities the MRE is significantly larger, reducing from 23% at low frequencies to 15% when approaching 1 MHz. Modelling the legs as homogeneous volumes results in a large MRE of approximately 47%.

3.3. Effects of muscle anisotropy on induced electric fields

The results from the homogeneous 0.1 mT magnetic field exposure scenarios are shown in figures 5(a) and (b). The ratio between the 99th percentile values of the induced electric fields and ICNIRP (2010, 2020b) basic restrictions for occupational exposure to time-varying electric and magnetic fields is shown in figure 5(a). The basic restriction for the internal electric field, in all tissues of head and body, is defined as a function of frequency: 0.27 mV m\(^{-1}\) × \( f \), where \( f \) is the frequency in Hz. With both the anisotropic and isotropic conductivity sets the ICNIRP basic restrictions are easily satisfied. Figure 5(b) shows the relative error between anisotropic and isotropic electric field calculations. In both figures 5(a) and (b) the mean is plotted with coloured areas showing
the standard error of the mean. At 10 kHz, using isotropic conductivities can lead to a 38% overestimation of the electric field or an underestimation of roughly 20% depending on the direction of the magnetic field. These errors diminish with increasing frequency, reducing to less than ±10% when approaching 1 MHz.

4. Discussion

In this research, the conductivity of human anisotropic skeletal muscle and fat was estimated in vivo by combining bioimpedance measurements with individual MRI-based volume conductor models and solving the inverse problem with computational methods. In addition, these newly derived conductivity values were used to study the effect of muscle anisotropy on the induced electric fields from homogeneous magnetic field exposure.

The anisotropic conductivity values derived for muscle are compared to relevant literature in figures 6(a) and (b) for longitudinal and lateral conductivities, respectively. It is to note that values are extracted from the graphical results reported in the papers considered (excluding Gabriel et al (1996a, 1996b, 1996c) who have published their measurement data online). The works by Ahad et al (2009), Epstein and Foster (1983), Nagy et al
and Sanchez et al (2014) performed their measurements on excised animal tissue samples: rats, dogs, mice and rats, respectively. In these studies, the measurement setup was roughly the same: a dielectric cell was used to measure the permittivity and the conductivity of the sample, with variations mainly in sample sizes and measurement devices. With these kinds of measurements the muscle fibres can be carefully aligned to the electric field direction, thus the ‘true’ directional dependence can be observed. The measurements performed by Gabriel et al (1996b) were done by using an open-ended co-axial probe on ovine muscle. Their co-axial probe measurements mainly show the effect of fibre direction and not the dielectric properties with the field along and across the fibre, as noted by the authors. The most similar method, compared to the method proposed in this work, was conducted by Hart et al (1999), who used impedance measurements combined with an FDM to approximate the longitudinal and lateral conductivities of frog muscle. As can be seen from figures 6(a) and (b),
Hart et al (1999) reported considerably lower values compared to the other studies. This may be due to temperature dependence, since their measurements were performed on room temperature samples. All the other studies were conducted at body temperature. The anisotropic muscle conductivities found in this work fall in the high range of the literature values. Figure 6(c) shows the ratio of lateral to longitudinal muscle conductivity. Even though the longitudinal and lateral conductivities vary from study to study, the anisotropy follows a clear pattern. It is quite clear that, apart from this work and Gabriel et al (1996a, 1996b, 1996c), the measurement direction has been carefully aligned to the sample fibre direction since. As noted by Epstein and Foster (1983) misalignment from true perpendicular or parallel orientations to muscle fibres can lead to large measurement errors, which may be the cause of the variation. By observing the measurements by Hart et al (1999) the effects of sample temperature seem to disappear since the observed ratio, of lateral conductivity to longitudinal conductivity, is in line with the other studies.

The conductivity of fat was also compared to values found in literature. From figure 6(d) it can be seen that the values roughly split into two categories of high and low conductivity. These differences are likely due to different origins of the fat samples with varying water content in the samples. The values reported by Gabriel et al (1996a, 1996b, 1996c) are measured from human breast fat at body temperature. The follow up study by Gabriel et al (2009) reported values for porcine fat in vivo. Hershkovich et al (2019) measured and modelled the dielectric properties of human skin and subcutaneous adipose tissue using skin folds. Smith and Foster (1985) measured the conductivity from equine and canine samples with varying water content. In figure 6(d) the solid violet line is of high (21%) and the dashed line of low (8%) water content. Wake et al (2016) measured the properties of porcine skin and subcutaneous tissue. The discrepancies between these results may be due to the varying water content of the tissue samples, it being higher in subcutaneous tissue than in fat found deeper under the skin. The results of this work seem to represent the latter, since the skin was modelled quite thick as well as conductive.

The validity of the estimated homogeneous values can also be assessed. Gabriel (2006) reported the low frequency conductivities of the whole and parts of the body which were obtained by volume averaging. The conductivity for the whole body was found to be 0.28 S m$^{-1}$ at 10 kHz and 0.29 S m$^{-1}$ at 100 kHz and for the leg 0.24 S m$^{-1}$ at both 10 and 100 kHz. These values are in the same vicinity of the homogeneous tissue conductivities, shown in figure 4(c), of 0.22 S m$^{-1}$ at 10 kHz and 0.29 S m$^{-1}$ at 100 kHz.

The study participants can be roughly divided into two age groups: young- and middle-aged adults. Age dependence on the tissue conductivities was investigated by performing the conductivity estimation process to the subdatasets. The conductivity estimates and corresponding Bootstrapped confidence intervals showed no
statistical significance for neither anisotropic muscle nor fat between the age groups. The same process was also performed by dividing the dataset according to sex. Again no statistical significance could be found. The resulting conductivity estimates were of course more uncertain since the size of the dataset was halved in both aforementioned cases. The latter finding differs from the results published by Ohmine et al (2004), who estimated the conductivities of muscle (as an isotropic tissue) and fat with a similar method based on bEIT. They found the frequency independent LF conductivities of 0.40 S m$^{-1}$ and 0.35 S m$^{-1}$ for male and female muscles, respectively, and 0.15 S m$^{-1}$ and 0.11 S m$^{-1}$ for male and female fat tissues, respectively. In their work, four model patterns of human forearms were combined with measurements from 18 subjects. It is difficult to say if their forearm patterns accurately model how subcutaneous fat is distributed in males and females, which may be the cause of the observed differences.

For the homogeneous magnetic field exposure, with both anisotropic and isotropic muscle conductivities, at the occupational reference level of 0.1 mT (ICNIRP 2010, 2020b), the 99th percentile electric field strength was below the basic restrictions in all subjects in all three exposure scenarios. The highest exposure over all scenarios was 37.3 ± 5.5% and 36.7 ± 4.2% (mean ± standard deviation) of the corresponding basic restriction, for anisotropic and isotropic muscle conductivities, respectively. Otherwise the results are rather ambiguous, since depending on the direction of the exposure, anisotropic modelling of muscle may lead to an over- or underestimation of the electric field compared to using isotropic estimates. It is still clear that the effects of muscle anisotropy are the most prominent at lower frequencies. Unfortunately, since the frequency range was limited to 10 kHz it is still unclear whether the effects of muscle anisotropy further intensify in the extremely low frequency range.

This study has its limitations. Firstly, the accuracy of impedance measurements performed with the Keysight E4990A impedance analyzer are significantly degraded when parasitic capacitances are present in the measurement configuration. Unfortunately in this work there was no way to compare the results of the hook artefact correction, suggested by Freeborn et al (2018), to a known reference. Thus, it is recognized that above 100 kHz additional uncertainties might be present. Also in future investigations, emphasis should be placed on obtaining artefact free bioimpedance measurements which extend to the extremely low frequency range, where reliable data of tissue dielectric properties are most scarce and uncertain. Possible solutions include: fabricating a custom front end amplifier that increases the input impedance of the measurement device (Kassanos et al 2021), exploring other measurement instruments or applying correction methods to separate the electrode polarization impedance from the tissue impedance (Kalvoy et al 2011, Gómez-Sánchez and Felice 2012). Secondly, the anatomically realistic volume conductor models represent a simplified structure of seven different tissue types since muscle and tendon were modelled as a single tissue. The rather large spatial resolution of the MR-images (2.1 mm × 2.1 mm × 2.1 mm) was also a limiting factor when separating fine structures. Third, the brute force approximation required the use of frequency independent constant conductivities of tissues. However, without these assumptions the procedure would be computationally unfeasible. It is recognized that a significant source of uncertainty associated with the use of anatomical models in numerical dosimetry calculations are the electrical conductivity values used in the models. Lastly, the direction of cells in anisotropic tissues should also be more accurately modelled.

5. Conclusion

In this paper the electrical conductivity of human anisotropic skeletal muscle and fat was studied using a method similar to bounded electrical impedance tomography. The results provide new information on the electrical conductivity of human tissues in vivo. In addition, these newly derived values were also used to study the effects of muscle anisotropy on the induced electric fields from magnetic field exposure and it was shown that the effects of muscle anisotropy diminish when approaching 1 MHz.

Ethical statement

Subjects gave their written informed consent in accordance with the Declaration of Helsinki. None of the participants reported any discomfort or pain during the experiment. The study protocol was approved by the Aalto University Research Ethics Committee (approval ID: D/1076/03.04.2020).

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