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A finite element model of needle electrode spatial sensitivity

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Abstract
We used the finite element (FE) method to estimate the spatial sensitivity of a needle electrode for bioimpedance measurements. This current conducting needle with an insulated shaft was inserted in a saline solution and the current was measured at the neutral electrode. Model resistance and reactance were calculated and successfully compared with measurements on a laboratory model. The sensitivity field was described graphically based on these FE simulations.

Keywords: FEM, needle electrode, electrical polarization impedance (EPI), bioimpedance, stainless steel, spatial sensitivity

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

1. Introduction

The main objective of this work was to describe the spatial sensitivity field of a needle electrode with an insulated shaft.

Spatial sensitivity is a frequently discussed topic in the field of impedance measurements. A common misconception is that the measurement reflects the properties of the sample situated between the measurement electrodes. In most cases this is not very accurate, and for small electrodes (relative to the tissue volume between them) this can be completely wrong. Earlier investigators have estimated the spatial sensitivity of monopolar needle electrodes in different ways⁴ (Cao et al 2002, Kinouchi et al 1997), but despite the high relevance we are not aware of any published models for estimating spatial sensitivity for needle electrodes.

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⁴ Cao et al and Kinouchi et al conducted a series of bioimpedance measurements and observed the sensitivity effect. No models for estimation of spatial sensitivity were made.
Needle electrodes are widely used in different clinical procedures and a high-quality electrical model for better descriptions of needle electrodes will hence be of great value. Typical examples are measurements and stimulations in anesthesiology and neurology (Tsui et al 2004, Sauter et al 2009, King et al 1998).

The aim of this study was to develop a method to describe the sensitivity field for needle electrodes in an ongoing project concerning a new method for anatomical needle positioning in the clinic (Kalvøy et al 2009). The description of the sensitivity field will furthermore be important in an ongoing resuscitation project using bioimpedance measurements to confirm the intravascular needle position. The objective of this project is to assure fast and accurate access to blood vessels for administration of drugs and liquid for induced hypothermia.

The sensitivity field $S$ can be expressed by the following equation (Geselowitz 1971):

$$ S = J'_\text{reci} \cdot J'_\text{cc} $$  \hspace{1cm} (1)

($J'_\text{cc}$ and $J'_\text{reci}$ are current density vectors where $J'_\text{cc}$ is the current density and $J'_\text{reci}$ is the reciprocal current density).

In the case of a two-electrode system, $S$ becomes

$$ S = |J'|^2 \left[ \text{1/m}^2 \right] $$  \hspace{1cm} (2)

To obtain the sensitivity field distribution for this system, a finite element (FE) model was used. This FE model was based on an actual lab model (Kalvøy et al 2009) with a needle in a saline solution. The saline solution was considered as being purely resistive due to our measuring frequency in the sub-megahertz range (Cooper 1946).

The needle was modeled with an insulated shaft. This insulation can be envisaged as a cylindrical capacitor with a capacitance given by the insulation thickness and the inserted needle length in the saline solution. When this non-insulated active electrode surface is brought into contact with biological tissue or an electrolytic solution, different electrochemical phenomena will occur resulting in an electrical polarization impedance (EPI) layer. The EPI was modeled as a constant phase element (CPE) in the equivalent circuit (Mcadams et al 1995).

A literature study showed that a description of electrode properties as a double layer, faradaic charge transfer and sorption as layers with certain thickness, permittivity and conductivity, was not easily obtained. However, a high-quality descriptive model should be feasible without an exact match on the molecular level, and in the present work we have defined all electrode interface properties as an EPI layer.

2. Materials and methods

2.1. The electrical model

To establish a FE model able to simulate the properties of the needle electrode setup, we first had to determine the main properties of the real setup used in the laboratory. From separate measurements of the different parts of the needle electrode (insulation and active electrode area), and analytical calculations, we found that the equivalent circuit in figure 1 would adequately describe our setup if the components were properly adjusted. This was verified by curve fitting to frequency spectra measured \textit{in vitro} using the setup shown in Kalvøy et al (2009). This model was used as a design basis for the FE model and also as a tool to interpret the solutions/results from the FE model.

As shown above, the electrical model consists of three main parts: the active electrode surface (EPI constrictional zone), the needle insulation and the bulk saline solution.
The resistor $R_B$ was used for modeling the bulk saline solution in the tank. This component was used to visualize a change in resistance caused by an increased distance between the bottom plate and the needle without changing the components modeling the needle and its vicinities. Because of the small current density in the large volume of the bulk saline (refer to section 2.4 showing the relation between impedance and current density), the resistance of $R_B$ is typically in the order of a few ohms. Thus, changes in this resistance will in most cases be negligible compared to the impedance of the other components, except that when the electrode is close to the neutral plate where the total impedance is small.

COMSOL Multiphysics, the FE application used in this work, has no option for implementing CPE directly into a FE model, but if the frequency is constant a CPE will be reduced to an ideal resistance and capacitance (Grimnes and Martinsen 2008). The simulations in this study were conducted with a sinusoidal single frequency of 100 kHz (in accordance with the lab measurements), and the equivalent conductivity and permittivity were inserted in the FE model.

The current leaving the active electrode surface will meet the impedance of the saline in the immediate vicinity of the needle tip. A higher resistance contribution is found in this constrictive zone (Grimnes and Martinsen 2008 as shown in equation (4)) caused by the relatively high current density in this volume compared to the bulk. The resistance ($R_E$) in series with the CPE was added to model the purely resistive properties of the saline in this constrictive zone. The CPE is constant regardless of the insertion depth, but with a value given by the geometry and electrical properties of the modeled layer (section 1).

The needle insulation can be seen as a cylindrical capacitor with the Teflon insulation as the dielectric. This constitutes a current path in parallel with the CPE which we modeled.
by $C_T$. The size of this capacitor will depend on the insertion depth according to the area of insulation exposed to the saline solution. Before entering the bulk the current through $C_T$ must pass the saline in the closest vicinities of the Teflon. Because of the relatively small cross section of this sub-volume, compared to the bulk, an increased resistance is found here. This is implemented in the model by the use of the resistance $R_T$.

This model has its limitations concerning current density. All measurements were done with 30 mV on the 0.3 mm$^2$ needle electrode which equals 20–25 mA cm$^{-2}$ for the impedance of the used needle at 100 kHz. A limit concerning current density and stainless steel electrodes was tested in Kalvøy et al. (2010); Geddes et al. (1971) described the frequency-dependent limiting current density for linear properties of stainless steel electrodes up to 10 kHz. An extrapolation of his curves will indicate a limiting current density at 100 kHz around 30 mA cm$^{-2}$.

### 2.2. The geometrical model for finite element simulations

In the following, the structure of the geometrical model template will be described.

This model template was modified for each simulation by repositioning the needle to the specific depth. This resulted in the separate FE models which are different in only one way: the distance between needle and the saline tank neutral bottom plate.

The resistances $R_E$, $R_T$ and $R_B$ shown in figure 1 model the properties of different volumes of the saline. In a 3D FE–model, the input conductivity and the defined geometry will replace these three components. This will enable a much more detailed model (quantified below) and will in most cases give a better match to the laboratory setup measurements.

The saline solution was modeled as a cylinder with a height of 35 mm and a radius of 52.5 mm. All elements between this cylindrical boundary and the needle were defined as saline (conductivity $\sigma = 1.3$ S m$^{-1}$ (Grimnes and Martinsen 2008)).

The needle ‘Disposable Monopolar Needle Electrode, 37 × 0.33 mm, Medtronic Inc., Minneapolis, MN, USA’ has an active electrode area of 0.3 mm$^2$. This needle has an insulated shaft. The thickness of this insulation is estimated to be 26 $\mu$m based on our own measurements and information from the manufacturer.

Metal parts were given a relative permittivity $\varepsilon_r = 1$ and conductivity $\sigma = 4.032 \times 10^6$ S m$^{-1}$ corresponding to stainless steel. Insulation was given a relative permittivity $\varepsilon_r = 12.1$ and conductivity $\sigma = 1 \times 10^{-12}$ S m$^{-1}$ corresponding to Teflon.

The needle was modeled with a simplification concerning the transition from Teflon insulation to the bare needle tip. This can clearly be seen when figures 3 and 5 are compared: the FE model has a defined setup where there should have been a smooth transition from Teflon to metal. A more accurate transition zone will be a subject for further work. Another
simplification was to model the needle tip curvature and length of the bare metal needle tip. These figures were not given from the manufacturer, and they had to be estimated from visual inspection using a microscope. These estimates, together with the known needle diameter, were tuned to make the correct electrode area (according to the manufacturer). A more accurate needle geometry model will be a subject for further work.

Figure 3. The needle (37 × 0.33 mm).

Figure 4. The needle electrode as has been modeled.

Figure 5. Illustration of the needle tip geometry dimensions (not to scale): B1 = 26 μm; B2 = 0.5 μm; B3 = 0.33 mm and B4 = 0.4 mm.
The main challenge in modeling the EPI was defining its corresponding geometry and electrical properties. To find the electrical properties, the electrical model was used and compared with the lab measurements conducted by Kalvøy et al. (2009). The EPI geometrical thickness in the FE model should of course have been as close to a real thickness as possible, but as described in Grimnes and Martinsen (2008); EPI thickness means small dimensions in the nanometer range giving an unnecessary complicated FE model. This resulted in a FE model with a ‘suitable’ layer thickness of 0.5 \( \mu \)m and with adapted conductivity/permittivity giving the correct values for the CPE, a trade-off between the realistic model and a model that could easily be made and still perform well without reducing accuracy.

EPI was modeled as a layer of 0.5 \( \mu \)m with a relative permittivity (\( \varepsilon_r \)) of 1000 and conductivity (\( \sigma \)) 0.01 S m\(^{-1}\).

2.3. The finite element model

To get a 3D FE model definition of the geometrical model, an axis-symmetrical approach was chosen. This simplified the model to a 2D definition, which is far easier to simulate and handle than a real 3D definition. The solutions from an axis-symmetrical 2D definition will still account for all spatial contributions, i.e. a 3D solution.

The following boundary conditions (types A–E defined in figure 6) were applied to the model. Boundaries being in the center of the model were treated as axis-symmetrical boundaries (type A). The 2D model is revolved 360\(^\circ\) around this axis shown as a dotted vertical line in figure 6 where the radius = 0. The model had its excitation at the upper needle end with a potential of 1.0 V applied as an electric potential (type B) \( V = V_0 \) which specifies the voltage at the boundary. Since the wanted solution is a potential, it is necessary to define its value in the geometry for the solution to be valid. Ground (type C) was simply applied as \( V = 0 \).

Electric insulation (type D) was applied as insulation giving \( \mathbf{n} \cdot \mathbf{J} = 0 \) specifying that electric displacement is zero outside the boundary. All other interior boundaries were defined as continuity (type E), where \( \mathbf{n} \cdot (\mathbf{J}_1 - \mathbf{J}_2) = 0 \). This specifies that the normal component of the electric current is continuous across the interior boundary.
A built-in application mode was used to define the partial differential equations (PDEs); Meriodonal Electrical Currents for Quasi-Static Fields (COMSOL Multiphysics Manual (COMSOL 2008)):

\[ -\nabla \cdot d((\sigma + j\omega\varepsilon_0)\nabla V - (J^e + j\omega P)) = dQ_j. \]

This has its origin in the Maxwell equations based on the following assumptions being valid:

\[ \frac{\partial D}{\partial t} = 0. \]

As to the Maxwell equations, changes in time of currents and charges are not synchronized with the changes of the electromagnetic fields. If this effect can be ignored, electromagnetic fields can be described by only considering stationary currents. This approximation is valid when time variations are small and the studied objects are far smaller than the wavelength.

The needle model frequency is 100 kHz and the largest model part measures 50 mm. This shows that this FE model can be defined and solved as a quasi-static field. The skin depth in all domains is much larger than the geometry, resulting in an approximation that neglects the coupling between the electric and the magnetic fields, giving

\[ \nabla \times E = 0. \]

Meshing of the model has been done by using the straightforward method given by the meshing routines in COMSOL Multiphysics (CM), only based on model geometry. The mesh could have been optimized with a lesser number of elements and probably become a more efficient model concerning the usage of computer force, but due to the choice of using an axis-symmetric approach, all solving/handling of the model has been made easier.
To verify the FE model element minimum quality \( q \), we used the following simple equation:

\[
q = \frac{4\sqrt{3}A}{h_1^2 + h_2^2 + h_3^2},
\]

(3)

where \( A \) is the triangle area and \( h_1 \), \( h_2 \) and \( h_3 \) are the side lengths of the triangular element. If \( q > 0.3 \), the mesh quality should not affect the solution quality (COMSOL Multiphysics Manual (COMSOL 2008)). We found that \( q \) was equal to 0.46 and hence, an acceptable element quality.

The application (CM) will normally select the appropriate solver for the FE model. This was also true for this model and a thorough discussion of the choice of the solver is therefore left out. A linear system solver called Direct (SPOOLES) was used. For further information about CM solvers, see COMSOL Multiphysics Manual (COMSOL 2008).

2.4. Sensitivity

The measured potential between the needle tip and the neutral plate can be described by

\[
u = I_{\text{reci}}^{-1} \iiint \rho J_{\text{reci}} \cdot J_{\text{cc}} \, dV.
\]

If we define

\[
J' = \frac{J}{I_{\text{reci}}},
\]

where \( I_{\text{reci}} \) is the reciprocal current which in a two-electrode system is the same as the current through the needle. Then, we get \( J' \) as a relative surface density of the injected unity current. We can express the system transfer impedance with

\[
Z = \iiint \rho J'_{\text{reci}} \cdot J'_{\text{cc}} \, dV.
\]

(4)

The spatial distribution of \( S \) (equations (1) and (2)) gives the sensitivity field, and by multiplying with the local resistivity \( \rho \), the impedance contribution or impedance volume density of each voxel can be found. The impedance contribution from a sub-volume is found from equation (4) if the integration is done over that specific volume. If the current density and the impedance volume density are known for each voxel, the potential drop caused by a voxel (or all voxels in a volume) can be found. It follows from this that the ratio of the potential drop between two equipotential lines to the total potential also gives the ratio of impedance contribution of this sub-volume to the total impedance. In a two-electrode setup, a potential plot is an easy way to display the sensitivity field.

The system impedance is expressed as equation (4) where \( I_{\text{reci}} \) and \( \rho \) are constants. \( I_{\text{reci}} \) and \( \rho \) are used as ‘common’ or generic entities representing saline and double layer to simplify this expression; the sensitivity field outer limit could be expressed by just plotting the electric potential distribution. The sensitivity field outer limit will be given within 97% of the accessible range of the results. This 97% criterion was chosen based on observations in the lab; an impedance change corresponding to 3% of the impedance measured at 17 mm was observed over the mostly entire needle work length when moved from top to bottom of the saline solution except in the remaining few millimeters where the needle was either close to the saline surface or close to the tank bottom.
3. Results

In figure 8, the impedance $Z$ ($Z'$ and $Z''$) is plotted as a function of the distance between the needle tip and the neutral plate. The results from the lab measurements have been compared as a quick test for the FE model validity. $Z'$ is the real part and $Z''$ is the imaginary part of $Z$.

To calculate and display the sensitivity field, the following approach was chosen: find the saline solution volume (with EPI included) corresponding to 97% of the measured impedance. Then, make a graphical illustration of the potentials between 100% and 3% (from 1.0 to 0.03 V). This is shown in figure 9.

All needle positions, measured as the distance from the neutral plate, in the range from 4 to 30 mm showed a sensitivity field as in figure 9. For distances outside this range, the spherical shape in figure 9 was lost. For the distances less than 4 mm or larger than 30 mm, the needle detects the object that has protruded its sensitivity zone. In these cases, the objects are the bottom and the top of the saline solution.

4. Discussion

Figure 8 shows that the properties of the lab model are fairly well described by the FE model concerning the description of sensitivity as the overall curve shapes are similar. For the values of $Z'$ and $Z''$, the accuracy of the FE model compared to the lab model is relatively poor and indicates an offset of approximately 100 $\Omega$ for $Z'$ and 20 $\Omega$ for $Z''$. The FE model could have been better tuned to the lab model concerning these values, but these deviances should have a minor impact on the main issue with this work, i.e. description of spatial sensitivity.

Figure 9 shows a 97% sensitivity radius of 3.75 mm in saline. The inner equipotential line showed that the volume within 1 mm radius corresponds to 77.3% of the measured impedance. In other words, we could have replaced the neutral electrode in the bottom of the tank with a spherical electrode of radius 3.75 mm, centered in the middle of the needle tip. The total measured impedance in the setup would only be reduced by 3%. In Kalvøy et al (2009), we
Figure 9. ‘97% zone’, i.e. electrical potential distribution and sensitivity field. The needle tip is at this point 15 mm from the neutral plate. This figure shows that the sensitivity field is a sphere with an approximate radius of 3.75 mm. The vertical left scale has millimeter units. The scale to the right shows a voltage factor, i.e. the value 1.0 is the maximum voltage and found at the needle electrode.

had to move the needle one-third into this sensitivity zone, approximately 2.6 mm from the neutral plate, to obtain a 97% reduction of the impedance. This was what we could expect since the neutral plate only intercepted the sensitivity zone from one side. The correlation between simulations and measurements was also confirmed in a comparison in figure 8; in both figures we found a similar change in resistance ($Z'$) and reactance ($Z''$) as the needle tip approached the boundaries at 0 mm and 35 mm.

The simplifications of the needle geometry, described in section 2.2, are done to reduce the use of computer resources. In our opinion, these simplifications were beneficial, as the results above were obtained with reasonable computation time on a regular computer and are still in accordance with the required FE model quality as described in equation (3).

Still, some deviances can be found. Lab model results show more smooth curves for needle distances between 5 mm and 30 mm than the FE model due to only a sparse number of models and probably some errors. Nevertheless, figure 8 was just made as a test to visualize the FE model performance compared with the lab model. Each value displayed in figure 8 is based on a separate FE model. To reduce the total number of simulations, only a few models have been made for the needle distances between 5 mm and 30 mm. The simulated resistances are higher than the measured and the simulated reactances are lower. These deviances between the measured impedance ($Z$) in the lab model and calculated impedance in the FE model
can probably be reduced by further fine-tuning of the material constants in the EPI as well as tuning of the other material constants. Such refinements to adjust the relative level are continuously in focus as we carry on our work on this modeling, but similarities of the resulting curve shapes from the simulation and the measurements verify a reasonable quality of our model.

It is desirable to be able to do early-stage-FE modeling without the close connections to a lab model, but as shown through this work, the FE modeling process needs supportive lab measurements.

5. Conclusion

It has been shown that the sensitivity field in a measuring system involving a needle electrode can be modeled, solved and described by a finite element model. In this system, the sensitivity field has been estimated to be a spherically shaped zone with a radius of approximately 3.75 mm from the needle metal surface. This result corresponds well with the observations from the needle-lab model.

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