When to include ECoG electrode properties in volume conduction models

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Abstract

Objective. Implantable electrodes, such as electrocorticography (ECoG) grids, are used to record brain activity in applications like brain computer interfaces. To improve the spatial sensitivity of ECoG grid recordings, electrode properties need to be better understood. Therefore, the goal of this study is to analyze the importance of including electrodes explicitly in volume conduction calculations. Approach. We investigated the influence of ECoG electrode properties on potentials in three geometries with three different electrode models. We performed our simulations with FEMfun, a volume conduction modeling software toolbox based on the finite element method. Main results. The presence of the electrode alters the potential distribution by an amount that depends on its surface impedance, its distance from the source and the strength of the source. Our modeling results show that when ECoG electrodes are near the sources the potentials in the underlying tissue are more uniform than without electrodes. We show that the recorded potential can change up to a factor of 3, if no extended electrode model is used. In conclusion, when the distance between an electrode and the source is equal to or smaller than the size of the electrode, electrode effects cannot be disregarded. Furthermore, the potential distribution of the tissue under the electrode is affected up to depths equal to the radius of the electrode. Significance. This paper shows the importance of explicitly including electrode properties in volume conduction models for accurately interpreting ECoG measurements.

1. Introduction

Brain–computer interfaces (BCIs) translate recorded neuronal signals into input for a computer system, e.g. to control communication software [1]. To improve upon BCIs, recordings of relevant neuronal population activity need to be acquired with high resolution, to capture the detail of the cortical topography. In order to achieve that, both the precise placement of the electrode grid on the cortex and the electrode properties, such as the electrode size and material, need to be optimized, for instance, with regard to the subject’s head anatomy [2]. Designing an optimal electrode configuration in a sophisticated manner requires a description of the relation between the spatial distribution of neural current sources and the recorded electrophysiological data, such as electroencephalography (EEG) and electrocorticography (ECoG). In addition, accurate electrode models might increase the accuracy of source localization results [3, 4]. Volume conduction models are used to solve the forward problem, i.e. compute the electric potential in the brain given by a known source. Many factors can be included in the forward computations, such as skull anisotropy [5], tissue inhomogeneities...
and anisotropies [6], and dispersive tissue and electrode properties [7]. Modeling of recording electrodes is commonly approached by incorporating the electrodes as voltage measurement devices with infinite input impedance and a surface area of zero, i.e. the so-called point electrode model [4, 8, 9].

However, if the electrodes are relatively large or the contact impedance is relatively low (e.g. in EEG [10]), shunting effects due to the electrode-electrolyte interface occur [4, 9]. The potential within the electrode is homogeneous and thus the potential under the interface becomes more similar to the one of the electrode. In a multi-electrode array (MEA) simulation [11], it was found that the effects of not including the electrode in the model are negligible when the distance between the electrode and the source is at least 2 times the electrode radius. The effects of the electrodes are negligible for most MEAs, considering their small surface area. However, these effects are expected to be especially relevant in ECoG because of the large electrode surfaces close to the current source covering a relative large area. Since a low contact impedance is desired to achieve a high signal-to-noise ratio [12], not including electrode shunting in the model can lead to erroneous results of forward simulations.

In this work, we investigated the influence that ECoG electrode properties have on the simulated recorded potentials in volume conduction models, as well as on the distribution of the extracellular potential under the electrode surface. The Finite Element Method for useful neuroscience simulations [13] (FEMfuns4) was used as the forward simulation method, because of its flexibility in including sub-domains and interior and exterior boundary conditions representing the electrodes. This study aims to discover what type of electrode model should be used when recording from large electrodes relatively close to the source. Several parts of the recording device were explicitly modeled, for example, the metal part of the electrode, which generally is neglected because of the high input impedance of the amplifier. The importance of a detailed electrode model was tested using various source configurations and geometries and a general recommendation is provided of when an electrode model should be used.

2. Methods

2.1. Forward model

The electric potential \( \varphi \) generated in the brain can be computed through the quasistatic approximation of Maxwell’s equations [14]:

\[
- \nabla \cdot (\sigma \nabla \varphi) = \nabla \cdot J_p, \quad \text{in } \Omega, \tag{1}
\]

with a given primary current density \( J_p \) (current produced by neuronal activity or from stimulating electrodes) in a medium \( \Omega \) with conductivity \( \sigma \).

A homogeneous Neumann boundary condition (BC) is applied on the exterior (non-electrode) boundary \( \partial \Omega \):

\[
\sigma \nabla \varphi \cdot n = 0, \quad \text{on } \partial \Omega, \tag{2}
\]

where \( n \) is the unit outer normal vector on \( \partial \Omega \). The insulating condition (2) ensures that no currents flow out of the boundary.

Electrode-electrolyte interface

In this work we focused on modeling the recording electrodes as surfaces or volumes and incorporated a real-valued contact impedance representing the interface. The interface currents are described as Faradaic reactions in a thin-layer approximation with a real-valued surface admittivity \( y_k \) (S m\(^{-2}\)) expressed by the Robin BC applied at the \( k \)-th electrode:

\[
- \sigma \frac{\partial \varphi}{\partial n} = y_k (\varphi - \varphi_{metal_k}) \text{ on } \Gamma_k, \quad k = 0, 1, \ldots \tag{3}
\]

where \( \varphi_{metal_k} \) is the voltage of the \( k \)-th electrode and \( \Gamma_k \) is the boundary of the \( k \)-th electrode. In the case of recording electrodes, equation (3) needs to be used self-consistently, since for each electrode \( \varphi_{metal_k} \) is unknown. Using a Lagrange multiplier, \( \varphi_{metal_k} \) with electrode surface \( S \) can be found in a standard saddle-point problem by requiring:

\[
\varphi_{metal_k} = \frac{1}{S} \int_S \varphi \, dS. \tag{4}
\]

2.2. Three geometries and model parametrizations

Three studies were performed to investigate the effect of the presence of the electrode surface on the recorded potential and in the tissue under the electrode. In all the studies Lagrangian FEM [15] was applied to simulate the electric potential generated in a volume conductor by a known source. The FEM simulations were performed using the open-source forward modeling implementation in FEMfuns5, which is built upon the open-source software FEniCS [16, 17].

2.2.1. Validation with in-vitro data

In the first study, a validation was performed on an in-vitro MEA set-up [18]. The geometric model (M1, figure 1(A)) that corresponds to the MEA set-up consists of a cylinder (diameter: 19 mm, height: 5 mm) filled with Ringer’s solution (figure 1(A), blue). The MEA was positioned at the bottom of the cylinder, with 60 conical recording electrodes (base diameter:

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4https://github.com/meronvermaas/FEMfuns.

5Code from FEMfuns allows neuroscientists to solve the forward problem in a variety of different geometrical domains, including various types of source models and electrode properties, such as resistive and capacitive materials, and can be found at https://github.com/meronvermaas/FEMfuns.
recording electrodes
stimulating electrode
ground electrode

Figure 1. Geometrical models M1-M3. (A) Bottom view of an experimental MEA set-up [18] cylinder filled with Ringer’s solution (in blue). In the bottom right corner of the cylinder, the external cylindrical ground electrode (in purple) is positioned. On the cylinder bottom surface 60 distributed (in a 4 by 15 rectangular grid) conical recording electrodes (in black) and one square stimulating electrode (in red) are placed. The top panel displays a close-up containing 8 recording sites and the stimulating electrode. (B) 3D schematic representation (not to scale) of the semi-infinite cylinder with a perpendicular bipole. (C) Realistic head model [thielscher2015], with scalp (red), skull (yellow), CSF (green), grey (purple) and white matter (blue) compartments.

80 µm, height: 80 µm). The stimulating electrode was modeled as a rectangular surface (width: 60 µm, length: 250 µm) on the same MEA (figure 1(A), red rectangle in the close-up top panel). An external cylindrical ground electrode (diameter: 2 mm, height: 4.3 mm) was represented by a cavity in the Ringer’s solution subdomain (figure 1(A), purple cylinder positioned in the lower right region). We used the mesh that was generated by [18] using FEMLAB 3.1a (COMSOL AB, Stockholm, Sweden), which consisted of 63,214 tetrahedra.

With this MEA set-up (M1), we replicated the simulation performed by [18], comparing modeled electric potentials to experimentally recorded potentials. In addition to [18], the range of the nodal potential values on each recording electrode surface was computed. The whole domain was uniformly filled with Ringer’s solution ($\sigma_{\text{Ringer}} = 1.65 \text{ S m}^{-1}$), where the purely resistive version of Poisson’s equation was solved, i.e. equation (1), with the homogeneous Neumann BC (2). The stimulating and ground metal voltages ($\phi_{\text{metal}}$ in (3)) were set to 754.4 mV and 0 mV, respectively. As reported in [18], surface conductance ($y_k$ in (3)) was 338 S m$^{-2}$ for the stimulating electrode and 975 S m$^{-2}$ for the ground electrode. The experimental data, provided by [18], and our simulated data were compared using a regression analysis.

2.2.2. Electrode shunting

In the second study we investigated the shunting effect of the electrode. We used a geometry (M2, figure 1(B)) that is composed of two cylinders: a first cylinder (C1) (height and radius: 30 cm) representing the volume conductor and, a second cylinder (C2) (diameter: 4 mm, height: 0.5 mm) placed at the center of the bottom surface of C1, representing the electrode. The surface area of electrodes is thus 12 mm$^2$, which is comparable with Resume II, Medtronic, ECoG electrodes [1]. C1 is a good representation of a semi-infinite halfspace, i.e. volume conductor with homogeneous Neumann BC at the lower surface and homogeneous Dirichlet BC elsewhere. The mesh was generated with gmsh [19] and contained approximately $17 \cdot 10^6$ tetrahedra, with the tetrahedron inradius ranging between 3 µm and 0.4 cm, being more refined close to the electrode.

The boundary value problem in the semi-infinite halfspace (M2) was governed by equation (1) with insulating BCs on the bottom equation (2) of cylinder C1. Homogeneous Dirichlet BCs were applied to the sides and top of C1. In this study we considered three different configurations (for a schematic overview, see figure 3) for the electrode C2:

(a) insulating BCs (2) at the interface between C1 and C2;
(b) explicit metal subdomain with insulating BCs (2) on the outer facets;
(c) surface conductance (3) at the internal boundary between the tissue and metal region, i.e.
between C1 and C2, and the metal subdomain with insulating BCs (2) on the outer facets.

Conductivities of the tissue (C1) and metal were 0.3 S m\(^{-1}\) and 10\(^7\) S m\(^{-1}\), respectively. Three surface conductances representing the interface were used. A parallel combination of a pseudo-capacitance and charge transfer resistance yielded interface impedances of 372, 46 and 5.6 \(\text{Ω}\) at 100, 1000, and 10 000 Hz, respectively [20]. As a source model, a perpendicular and parallel oriented bipole source (i.e. a dipole approximated by two monopoles with a magnitude of 1 \(\mu\text{A}\) separated by 0.5 mm, with a dipole moment of \(5 \times 10^{-10} \text{ Am}\)) was positioned 1 mm above the center of the electrode surface.

In addition, we reported the proportion between the simulated electrode potentials of electrode configuration (a) and the other configurations, (b) and (c), for the perpendicular bipole. The electrode potentials of electrode configuration (a) were calculated using either the point-electrode approximation or the disc-electrode approximation. The point electrode approximation is the simplest and most commonly used electrode approximation [8, ], where the electrode potential is described as the value assumed at the midpoint of the electrode, disregarding both its size and interface impedance. In the disc-electrode approximation, the average potential on the electrode surface is calculated. The reported proportion between these two electrode potentials for configuration (a) and the other configurations, (b) and (c), indicates by which factor the recorded potential differs if a more complete electrode model is used.

Electrode implementations (a) and (b) were calculated several times with a bipole source positioned at increasing distances from the electrode. Furthermore, potentials at horizontal cross sections at increasing distance from the electrode surface were calculated with electrode implementations (a) and (b) using one bipole at 1 mm distance from the electrode.

Finally, the Root Mean Squared Errors (RMSE):

\[
\text{RMSE} = \sqrt{\frac{\sum_{i=1}^{N} (\phi_i - \psi_i)^2}{N}}, \tag{5}
\]

and the Relative Differences (RD):

\[
\text{RD} = \frac{1}{N} \sum_{i=1}^{N} \frac{|\phi_i - \psi_i|}{\max|\psi_i|}, \tag{6}
\]

where \(\phi_i\) is the numerical solution at location \(i\), \(\psi_i\) the analytical solution and \(N\) the number of locations, were calculated in a 1 cm cube centered around the electrode position, using electrode implementation a), to compare the analytical [21] and numerical solutions. When the comparison between the solutions of two models is calculated, the RMSE will be called the Root Mean Squared Difference (RMSD).

2.2.3. Realistic head model

In the third study, as a proof of principle, we used a realistic head model (M3, figure 1(C)) with scalp, skull, cerebrospinal fluid (CSF), grey and white matter compartments (available in SimNIBS, [thielischer2015]). Two cylindrical electrode surfaces (\(\approx 12 \text{ mm}^2\), 1 cm distance between centers) were manually added using Blender [22] and all head surfaces were subsequently tessellated using TetGen [23]. This resulted in approximately 4.8 \(\times 10^6\) tetrahedra, where the inradius of all tetrahedra was below 0.5 mm. The tetrahedra were comparable to or smaller than the semi-infinite halfspace tetrahedra in M2, being more refined close to the electrode.

The boundary value problem (1) with homogeneous Neumann BCs (2) on the outer surface was solved in the realistic head model (M3). Tissue properties were resistive with conductivities of the grey and white matter [24], CSF [25], skull [26] and scalp [27] being 0.28, 0.25, 1.59, 3.5 \(\times 10^{-3}\) and 0.17 S m\(^{-1}\), respectively. Two electrode models were considered, one perfectly insulating and one consisting of metal with insulating BCs. A bipole radially oriented with regard to the average normal of the surface of one of the electrodes was positioned at a 2 mm distance from the electrode. To make our results specific, we have chosen a bipole consisting of two monopoles of 1 \(\mu\text{A}\) separated by 1 mm (i.e. a dipole moment of \(10^{-9} \text{ Am}\)). As results directly scale with bipole size the calculated values can be easily adapted to the particular situation at hand by scaling.

The difference between the potentials given the two electrode models was computed for tetrahedra in the grey matter within distance of 20 mm from the electrode surface. Box plots of these differences were binned into a range of 1 mm distance from the electrode. The root mean square (RMS) of the potentials was also calculated in bins to show the effect of the electrode relative to the magnitude of the potential.

3. Results

3.1. Study 1: validation with in-vitro data

The results of the simulation of this monopolar stimulation experiment [18] are visualized in figure 2. The electrical potential in this stimulation set-up was recorded at 60 electrodes and compared to experimental recorded values. Using Robin BCs (3), as reported by [18], results in an excellent fit between experimental and modeled potentials (figure 2, \(R^2 = 0.999, p < 0.0001\)).

In addition to [18], we displayed error bars in figure 2 to indicate the range of the nodal potential values on each recording electrode surface. The recording electrodes closest to the stimulating electrode (i.e. with the highest potential) show the largest
range of values along its surface. These results motivated us to further investigate effects of more complete recording electrode models.

3.2. Study 2: electrode shunting

The focus of Study 2 was on the effect of the three different electrode configurations, i.e. (a), (b) and (c), and of the point- versus disc-electrode approximation on the electric potential computed with FEM-funs in a purely resistive medium (M2) according to equation (1).

We analyzed and compared the potentials on the midline along the surface of the electrode (in green, figure 3, left panels) of the geometrical model (M2), see figure 1(B).

The results are displayed in figure 3, where, on the one hand, in the vertical cross section (figure 3(A)), we notice that the potential distributions of the five electrode implementations are overlapping in the tissue (the main cylinder, C1). On the other hand, the difference between the insulating (a) and more elaborate electrode implementations (b)-(c) in the horizontal cross section is remarkable. While the insulating electrode implementation (a) leads to a parabolic potential distribution peaking at the center of the cylinder C1 (see figure 3(B)), the potential distributions computed with (b) have a constant value within the electrode. As the interface impedance increases, the shape of the potential distribution returns from constant value (b) to the parabolic one in the point electrode model (a). Indeed, with higher impedances, the Robin BC (equation (3)) reduces to a homogeneous Neumann BC (2). Note that the potential magnitude is omitted in figure 3, since the magnitude of the potential scales linearly with the magnitude of the dipole. Only the potential distribution along the electrode surface is of interest here.

Regarding the point- versus disc-electrode approximation, we inspected the effects of the electrode model on the recorded potential for the perpendicular bipole of figure 3(B). The proportions between the point- and disc-electrode approximation of the insulating electrode (a) and the more complete (configuration (b) and (c)) are reported in table 1. Since the bipole is centered exactly at the midpoint of the electrode, the point-electrode approximation overestimates the recorded potential by at least factor 3. The disc-electrode approximation is less sensitive to the position of the source, which reduces the effect of the electrode model on the recorded potential by at most 33% (factor 1.33).

The influence of the electrode on the forward solution

The difference between an insulating and a metal electrode in relation to its distance to the source was examined. In figure 4(A), the potentials on the midline along the surface of the electrode (in green, figure 3, left panels) are depicted for four source-electrode distances for a perpendicular and a parallel bipole. Potential values are displayed for a single-point electrode (configuration (a), figure 4 in orange) and for an explicit representation of the electrode (configuration (b), figure 4 in green). The RMSD (equation (5)) between the potentials using electrode configuration (a) and (b) was calculated for source distances between 0 and 10.5 mm from the electrode (figure 4(B)). As sources are positioned increasingly further, the voltage becomes more homogeneous in both electrode implementations. Thus, the difference between (a)−(b) is smaller for sources at a distance further than the diameter of the electrode (dashed vertical line in figure 4(B)). From figure 4(B), we can indeed see that the RMSD curves decrease almost exponentially for increasing distances (top right insets display the

**Figure 2.** Replication of the validation study in [18]. The red points correspond to the modeled and experimental values. A linear regression (blue line) with $R^2 = 0.999$ and p value $< 0.0001$ is shown. Furthermore, the total range of nodal potential values on the surface of each electrode is depicted by the red error bars.
Figure 3. Vertical (A) and horizontal (B), (C) cross sections of potential distributions in a semi-infinite cylinder, with in the top panel an overview of the different electrode configurations indicated in the legend (a)–(c). For a complete overview and description of the domain, see figure 1(B). The potential on the midline along the surface of the electrode is plotted from top to bottom describing: (a) only tissue (blue) with no electrode (i.e. insulating BCs), (b) tissue (blue) and explicit electrode (grey) domain (insulating BCs at the boundary of the electrode), (c) similar to (b) but including a thin-layer approximation of the interface (red), Robin BC with conductivities indicated in the legend). Note the different scaling on the x-axis of the vertical and horizontal cross section. (A) and (B) are potential distributions due to a perpendicular bipole, (C) is the potential when using a parallel bipole at a distance of 1 mm from the electrode surface. Since the magnitude of the potential scales linearly with the magnitude of the bipole, we are interested in the distribution of the potential and not the magnitude, therefore the axis labels are omitted here.

Table 1. Ratio between recorded electrode potentials of different electrode models with as the source distribution a perpendicular bipole. The values represent the ratio between the recorded potentials of the insulating electrode model (a) with a disc- and point-electrode approximation (rows) and the more complete electrode models (columns).

| Interface 372 Ω (c) | Interface 46 Ω (c) | Interface 5.6 Ω (c) | Metal (b) |
|---------------------|---------------------|----------------------|-----------|
| Disc-electrode      | 1.06                | 1.2                  | 1.29      | 1.33      |
| Point-electrode     | 3                   | 3.4                  | 3.67      | 3.77      |

RMSD values on a logarithmic y-axis). The largest RMSD values can be observed for bipole distances within the diameter of the electrode, i.e. 4 mm, for both perpendicular and parallel bipoles (figure 4(B) left and right, respectively). From RMSDs of 42.5 µV and 49.2 µV for a bipole at 1 mm from the surface of the electrode, we observe RMSD values of 1.1 µV and 2.7 µV for a bipole at 4 mm from the electrode surface, for the perpendicular and parallel bipoles, respectively. The pace with which the RMSD decreases with the distance of the source depends on the conductivity of the medium, in addition, the magnitude of the potentials scales with the magnitude of the bipole. Therefore, it is essential to consider the magnitude of the RMSD (figure 4(B)) with respect to the magnitude of the potentials themselves (figure 4(A)).

In addition, the potentials at a horizontal cross section at increasing distances from the electrode surface using one source location was examined using the insulating (a) and metal electrode (b) implementations (figure 5). The potentials on the midline along
the surface of the electrode are depicted at four distances from the electrode for a perpendicular and parallel bipole (figure 5(A)). As the distance from the electrode surface increases, the difference between the two implementations decreases. In the bottom panel (5(B)), the RMSD (equation (5)) between the insulating (a) and metal electrode (b) implementations is illustrated, showing that only close (i.e. at distances of the radius of the electrode) to the electrode, there is an effect in the potential of the tissue underneath the electrode.

Specifically, from RMSD values of 42.5 µV and 49.2 µV on the electrode surface, to RMSD values of 2.9 µV and 6.6 µV at 2 mm away from the electrode, with a drop of 39.6 and 42.6, for perpendicular and parallel bipoles, respectively. The pace at which the RMSD decreases with the distance from the electrode is dependent on the conductivity of the medium and the magnitude of the potentials scales with the magnitude of the bipole. Therefore it is essential to consider the magnitude of the RMSD (figure 5(B)) with respect to the magnitude of the potentials themselves (figure 5(A)).

The numerical accuracy was checked by calculating the RMSE (equation (5)) and RD (equation (6)) for the insulating electrode implementation, since there was an analytical solution available [21]. The RMSE was 0.007 and 0.009, and the RD was $3.6 \cdot 10^{-7}$ and $2.7 \cdot 10^{-7}$ for the perpendicular and parallel bipole, respectively.

3.3. Study 3: realistic head model
As a proof of principle, in the last study, we simulated the effect of electrodes in a realistic head model with resistive material properties (1). A radial bipole was positioned at 2 mm below the surface of one of the electrodes. The resulting potential distribution is shown in figure 6(A)–(C). Insulating electrodes (figure 6(A)) lead to an inhomogeneous potential distribution along the surface of the electrode. In contrast, electrodes consisting of a highly conductive metal have a homogeneous potential distribution over the surface of the electrode (figure 6(B)). This is comparable to the findings in figure 4, except that the radial bipole here is not centered on the middle of the electrode. The difference between the simulated results for the insulating and the ones for the metal electrode is between $-7 \mu V$ and $80 \mu V$ (clipped to $7 \mu V$ in figure 6(C)). Note that the difference is non-zero not only under the surface of the electrode, but also in the nearby gyri (figure 6(C)). This is depicted in more detail in figure 6(D). The difference between the two solutions is shown in the box plots as a function of binned distance from the electrode. The root mean
Figure 5. A) Horizontal cross-section of a semi-infinite cylinder displaying potentials on the midline along the surface of the electrode at increasing distances from the electrode surface with a perpendicular (top panels) and parallel (bottom panels) bipole. The bipole is located 1 mm above the electrode. The potential distribution at the metal electrode is shown in blue, the potential distribution at the insulating electrode is shown in pink. On the x-axis the electrode is shown as a grey bar. The y-scales of the panels are different. B) Root mean squared differences (RMSD) between the potentials of the two electrode configurations on the midline along the surface of the electrode are shown as the distance from the electrode increases. In the top right corner of the plots, the RMSD is displayed on a logarithmic y-axis.

The purpose of this study was to determine whether the presence of ECoG electrodes has a significant influence on the electric potentials on the electrode surface and underlying tissue. Three studies with three different geometrical models were performed using the open-source forward modeling implementation in FEMfuns.

In Study 1, we replicated a stimulation study [18] and in addition illustrated that the potential distribution along the surface of MEA conical electrodes close to the source is inhomogeneous (figure 2). Applying a surface conductance via Robin BCs (3) to the recording electrodes did not affect the fit of the model with the experimental data. Thus, the investigators of study [18] used homogeneous Neumann BCs and disregarded an effect of the interface impedance. However, depending on the magnitude of the surface impedance and the electrode size, the potential distribution under the electrode could be altered.

The effect of the electrode on the potential distribution was further examined in Study 2 using a semi-infinite halfspace and several electrode implementations. The shunting effect of the electrode alters the potential. A low interface impedance and the high metal conductivity ensure an approximately homogeneous potential distribution under the electrode. When the interface impedance goes to infinity the potential distribution over the electrode surface becomes inhomogeneous and varies with distance from the source. This is clear from equation (3), where a high surface impedance \( \gamma_k \) reduces (3) to a homogeneous Neumann BC (2).

A number of EEG modeling studies [4, 9] conclude that assuming a surface instead of a point is necessary when the electrode surface impedance is very low or the electrodes are large compared to the head. Reference [4] reports that a more extensive electrode model can improve EEG forward model accuracy. However, this will be mostly prominent in neonatal EEG, where the electrode diameter is large relative to the head. Few studies have considered the effect of recording electrodes close to the source (e.g. ECoG, MEA) [11].
We show that large electrodes relatively close to the source (i.e. distances equal or smaller than the size of the electrode itself) require a surface rather than a point electrode modeling approach (figure 4). This result is comparable to previous findings, which conclude that an insulating point electrode is only sufficiently accurate within 4 times the electrode radius [11]. The set-ups that were used here represent commonly used ECoG grids [1], where the electrodes are cylinders with a diameter of 4 mm. Considering that the average thickness of grey matter is around 3 mm [28], the distance from the source to ECoG electrodes is in many cases smaller than the diameter of the electrode and therefore it is necessary to model the electrode explicitly.

Thus, when simulating large electrodes near the source, choosing the appropriate electrode model is essential. In table 1, we show that using a point-electrode approximation largely overestimates the recorded potentials, as compared to any of the more elaborate electrode models. The difference between the disc-electrode approximation and the elaborate electrode models is less pronounced. Therefore the disc-electrode approximation could be considered as a minimally required electrode model when recording close to the source; especially because of its straightforward implementation. A more complete electrode model should be considered if the electrode interface impedance is low to prevent an overestimation of the recorded potential by up to 33%.

In general, we have to consider that no clear consensus has been established yet on exactly how local the signal recorded by ECoG grids is [29, 30]. However, due to the decrease of the potential with distance, the majority of the signal will represent mainly local sources.

To better understand the electrode effect, one should consider that the value recorded by an electrode can be described as the integral over its surface (4), or more completely with Robin BCs (3) [4, 20, 31]. Thus, the potential given by larger electrodes is the average over an increasingly inhomogeneous potential distribution, resulting in a loss in spatial resolution. When recording close to the source,
the potential distribution over the electrode surface is likely to be more inhomogeneous. In contrast, when recording a faraway source, the potential distribution over the electrode surface will be homogeneous.

When including an amplifier input impedance to the electrode model (results not shown), values around 10 MΩ are already sufficiently high, so that there is no difference from the case with a perfectly insulating (infinite impedance) amplifier [32] considered in our calculations.

The shunting effects that cause the potential averaging over the electrode surface only affect the tissue in its proximity (figure 5). When recording close to the source with a low contact impedance, the potential distribution under the electrode is affected up to the radius of the electrode. Depending on the electrode size and the area that needs to be recorded from, it is thus important to ensure a sufficiently high surface impedance, so that current flow in the underlying external cortical layers is not affected.

We should note that comparing electrode configuration (a) and (b) depict a worst case scenario. Depending on the size of the surface impedance, electrode effects could be less pronounced. However, the surface impedance of the electrodes is not generally measured, and also can require extensive simulations [18]. An optimal surface impedance for a specific electrode set-up should thus be computed in order to assure that one can record without affecting the underlying tissue, and also in order to keep the impedance as small as possible to increase the signal-to-noise ratio [12].

As a proof of principle (Study 3), the difference between point electrode and metal electrode recorded potentials generated by a dipole in a realistic head model is visualized on a realistically shaped cortex (figure 6). The difference between the two electrode types relative to the RMS of the potentials is small for sources at a location further away (a distance of the diameter of the electrode). However, when the source is placed closer the electrode effect becomes more pronounced in the nearby tissue. The potential distribution on the electrode surfaces (figures 6(A) and (B)) demonstrates the need for a disc-electrode approximation. The recorded potentials using a point-electrode approximation are highly dependent on the position of the electrode midpoint.

Using a more complete electrode model can be important in determining the ideal spatial resolution of electrode grids recording close to the source. Adopting more elaborate source models could also assist in determining optimal grid design for these electrodes. Furthermore, both study 2 and 3 suggest that using the inhomogeneous potential distribution over the electrode surface using a point electrode could lead to significant errors in applications such as source reconstruction. Thus including a more complete electrode model could be necessary to improve source localization errors.

Including an explicit region for the electrode in the volume conductor can also shed light on the effect of a non-bending electrode with respect to the curvature of the brain, which affects the distance between the source and the electrode. With recent advances in high-density EcoG grids, the shunting effects of the electrodes might be aggravated, if, for example, the combined contact surface of the high-density EcoG electrodes covers a relatively large area of the underlying tissue. Which is in accordance with findings in EEG simulation studies [3, 4]. In future work we plan to use the electrode models to create sensitivity maps and receiver operating characteristic for high-density EcoG grids.

5. Conclusion

In conclusion, if the distance between an electrode and the source is equal to or smaller than the size of the electrode itself, electrode effects should not be disregarded in simulation studies. In the examples that were presented, typical EcoG electrode sizes of 2 mm radius [1] were used, indicating that modeling studies for these types of grids require a more extensive electrode model. Furthermore, it was shown that with a low electrode contact impedance and nearby sources, the potential distribution of the tissue lying directly under the electrode is affected by the presence of the electrode.

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