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The Impact of Electrode Characteristics on Electrocorticography (ECoG)

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Abstract— **Used clinically since Penfield and Jasper's pioneering work in the 1950's, electrocorticography (ECoG) has recently been investigated as a promising technology for brain-computer interfacing. Many researchers have attempted to analyze the properties of ECoG recordings, including prediction of optimal electrode spacing and the improved resolution expected with smaller electrodes. This work applies an analytic model of the volume conductor to investigate the sensitivity field of electrodes of various sizes. The benefit to spatial resolution was minimal for electrodes smaller than ~1mm, while smaller electrodes caused a dramatic decrease in signal-to-noise ratio. The temporal correlation between electrode pairs is predicted over a range of spacings and compared to correlation values from a series of recordings in subjects undergoing monitoring for intractable epilepsy. The observed correlations are found to be much higher than predicted by the analytic model and suggest a more detailed model of cortical activity is needed to identify appropriate ECoG grid spacing.**

I. Introduction

The origin and time-frequency structure of bioelectrical signals critically informs the design of recording electrodes and subsequent signal processing and analysis. Electrocorticography (ECoG) records electrical activity from the cortical surface, either above or below the dura mater. This technique has proven useful for both clinical epilepsy seizure foci localization [1] and neuroscience research [2-4]. Recent work in ECoG-based Brain-Computer Interfaces (BCI) has triggered great interest in optimizing the design of ECoG electrodes used to record cortical activity [5]. ECoG signals can be recorded with electrodes of varying shape, size, spacing, material and placement, and they can be further processed with various signal analysis techniques. Models of the bioelectric generation and propagation of neuronal activity can help clinicians, researchers, and engineers make informed decisions on these parameters regarding recording electrode design and signal processing techniques. McIntyre et al. [6] is a good example of using bioelectrical field modeling to understand the volume of tissue directly affected by deep brain stimulation, and their model can help physicians to identify stimulation parameter settings that are most effective to individual patients. Several groups have proposed informative models in cortical surface field potentials. For example, Miller et al. [7] provided a simple *ad hoc* model explaining the spectral power modulation in ECoG signals observed during various behavioral conditions (e.g. movement). Taking a more biophysical approach, Slutzky et al. [8] calculated a lower bound for the spacing of ECoG electrodes based on the spatial frequencies of current sources within the volume conductor model of the cortex. The present work applies an analytical biophysical model to estimate the volume within cortex a given electrode will be sensitive to, in an attempt to identify the engineering factors and trade-offs in ECoG electrode design. The zero-lag correlations between electrodes at given separation distances predicted by the analytic model are compared to observed zero-lag correlations from human ECoG recordings in order to investigate the utility of this model for determining optimal electrode spacing.

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II. Methods

A. *Analytical Volume Conductor Model*

A model is needed to predict recorded potentials given electrical fields produced by neural sources. As has been done previously we begin with Maxwell's equations of electromagnetism, making the quasi-static assumption that capacitive effects may be neglected due to the low frequencies of interest (typically 0 to 200 Hz)[9]. Solving from the electrode surface to an infinitesimal sphere around the neural source leads to Laplace's equation. Note that although one neural source has been assumed, linear superposition applies, allowing arbitrary source configurations to be produced through simple addition.

$$\nabla^2 \phi = 0 \qquad (1)$$

Applying the method presented by Rattay for his analysis of surface stimulation of peripheral nerves by a disc electrode, the solution in the volume conductor [10] is

$$V(r,0) = V_0 \quad for\ r \le a \qquad (2a)$$

$$V(r,z) = \frac{2V_0}{\pi}\sin^{-1}\left(\frac{2a}{\sqrt{(r-a)^2+z^2}+\sqrt{(r+a)^2+z^2}}\right) \quad (2b)$$

for voltage V, radial distance from the center of a disc electrode r, with radius a, and depth (perpendicular to the plane of the electrode) z, assuming a disc electrode on a semi-infinite homogeneous medium, with perfectly insulating skull.

Electromagnetic reciprocity (which has previously been used to calculate lead-field matrices for similar problems [11]) states that source and field points may be interchanged, so that this same equation describes the potential on the electrode due to a source of strength V_o at position (r,z).

B. *Measuring Temporal Correlation*

If the cortical currents were independent at each point in space, one would expect the correlation between recording channels to simply reflect the cross-talk inherent in the volume conductor (i.e. the conductive cortical tissue and cerebrospinal fluid (CSF)). This assumption was tested by comparing the theoretical cross-talk between electrode pairs predicted by the analytic model with experimental data. Specifically, for the latter, we calculated the zero-lag cross-correlation of real ECoG signals between pairs of electrodes implanted subdurally in human subjects. The cross-correlation at zero time lag can be thought of as the size of the component of a in the direction of b, relative to the size of b itself (Eq. 3).

$$r = \frac{a \cdot b}{\|b\|^2} \quad (3)$$

C. *Data Collection*

Data was collected from 15 subjects (ages 8-45) undergoing ECoG grid placement for presurgical mapping of epileptogenic foci. Research procedures were approved by the local Institutional Review Board (IRB). During experiment sessions subjects were asked to relax with eyes open observing a blank screen for 60 seconds, which was essentially a baseline resting condition. ECoG signals were recorded from both clinical (3mm diameter, 10mm spacing) as well as custom electrodes (1.5 mm diameter, 4mm spacing) at 1200 Hz with g.USB amplifiers (Guger Technologies, Austria) and BCI2000 software [12]. An offline comb filter was applied to remove 60Hz line noise and harmonics before further offline processing.

III. Results

A. *Spatial Resolution as a Function of Electrode Size*

Intuitively, larger electrodes will record from larger volumes of brain tissue. However, it is often unclear the exact volume a cortical surface electrode with a particular size will record from (i.e. the volume within which neural activity will be detected by the electrode). Moreover, the shape of that volume of tissue will also depend on the cortical surface electrode shape and size. In this paper, we focus on electrodes with a round cross-sectional shape, as those are the most often used for cortical surface recording. In our model, we define sensitivity as the ratio of the voltage observed at the recording electrode to the source voltage at a particular spatial location. Figure 1 presents sensitivity vs. source depth and radial distance for several electrode sizes calculated theoretically assuming a homogeneous medium. The sensitivity of smaller electrodes decays more quickly with source depth, and also provides a smaller half-width at half-maximum (a measure of spatial resolution) at any particular depth. An electrode that is too small will not provide the necessary sensitivity to deeper cortical sources, while one that is too large will record unwanted deep sources (for example, from white matter) and a larger volume of tissue overall. Further, Figure 1c demonstrates that decreasing cortical surface electrode size beyond a certain point (in this case 1mm) provides little benefit to spatial resolution for sources at a given depth in cortex.

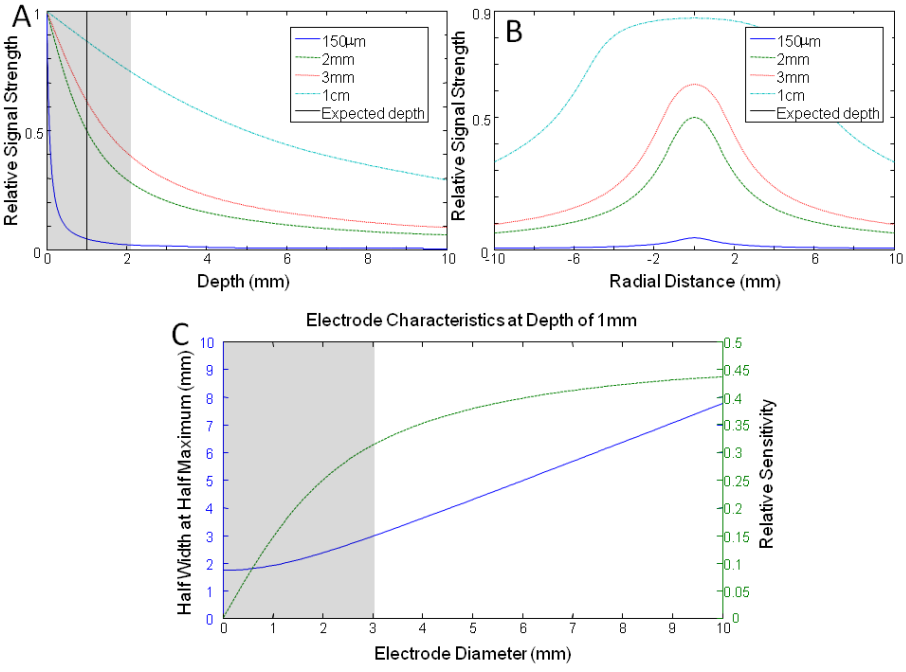

Figure 1 – Sensitivity of Disc Electrodes. (A) Sensitivity for electrodes of various diameters vs. depth in cortex. Notice the rate of decay changes dramatically based on size. The solid black vertical line at 1 mm is the anatomical depth of the cortical sources [13]. (B) Radial sensitivity for electrodes of commonly used diameters for sources that are 1mm deep. As expected, smaller electrodes provide smaller regions of sensitivity, however their peak sensitivity is also decreased, leading to overall lower signal-to-noise ratios. (C) Summary of both spatial resolution (half-width at half-max) and sensitivity for electrodes of various diameters for sources at 1mm depth in cortex. Precise optimization of electrode size will depend on the recording system noise floor and on any specific areas of interference. Shading denotes areas of typical interest for ECoG recording. The large scale of the plots is provided to demonstrate overall trends.

Given known anatomical information

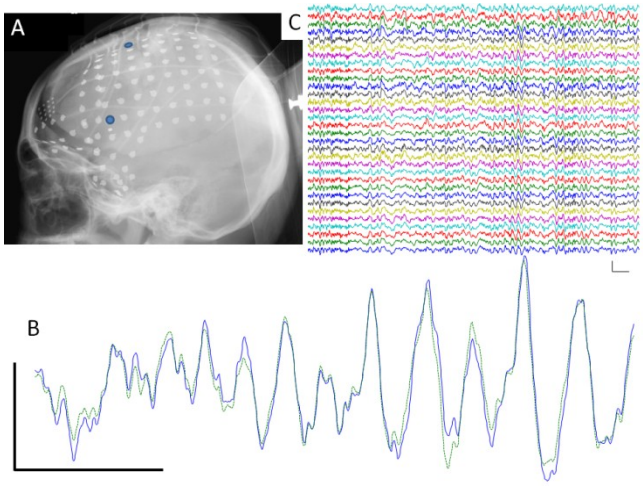

Figure 2 – Example of zero-lag time-correlated ECoG signals. (A) X-ray showing ECoG electrode positions, annotated to highlight two electrodes whose signals are shown in lower-right. (B) Signals recorded during baseline from electrodes 2 (dotted line) and 24 (solid line). Zero-lag cross-correlation between the full 50 second recordings on these channels was 0.91. (C) Channels 1 to 32 over 5 seconds, shown as an example of the large scale correlation across the grid. Scale bars show 250μV and 250ms. Electrodes (Ad-Tech Medical Instrument Corp.) are 3mm in diameter with center-to-center spacing of 10mm.

about the cortical structure, it is possible to determine the electrode size and spacing which balance sensitivity to the cortical volume of interest (i.e. amplitudes for neural signals of interest)with the size of that cortical volume (i.e. spatial resolution).

B. Correlation among Neural Sources

In addition to electrode size, electrode spacing is another important design parameter. Ideal electrode spacing will allow us to fully sample the cortical surface potential with a minimum number of electrodes. This is predominantly influenced by the spatial frequency of cortical surface potentials, which can be thought of as the degree to which ECoG signals recorded from two neighboring electrodes correlate. The technique used by Slutzky et al. [8] could be applied to electrodes of a given size to calculate a lower bound for spacing, i.e. the smallest spacing needed, assuming any correlation in ECoG signals is only due to the biophysics of volume conductor. If there is any correlated neural activity across the cortical surface, correlation in ECoG signals recorded from neighboring electrodes can increase, and an electrode spacing above the theoretical lower bound could be sufficient. For each subject in the baseline recordings introduced above, Equation 3 was applied to each pair of electrodes to estimate their correlation. In a simple situation where each volume contained an independent source, this metric would match

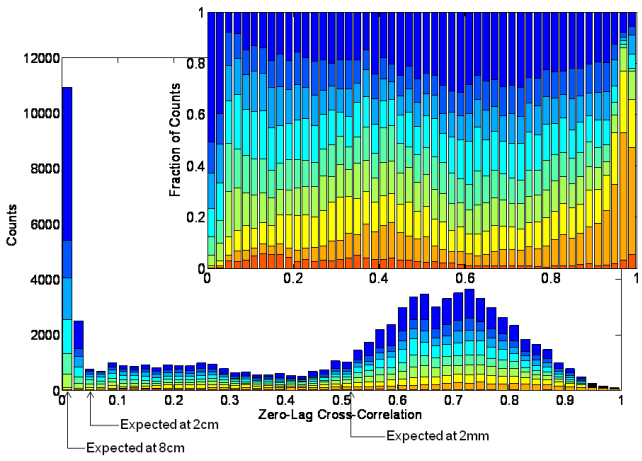

Figure 3 – Stacked histogram of correlation values (between electrodes pairs for each subject, aggregated, but not averaged, across all subjects) separated by distance between the electrodes. Colors signify 1cm increments of electrode spacing, from <1cm (red) to >8cm (blue). The peak near R=0.7 is remarkably unaffected by electrode spacing, while the highest and lowest correlations show clear electrode spacing related effects (see normalized inset), i.e. high correlation values for small electrode spacings (more counts with red and orange colors) and low correlation values for large electrode spacings (more counts with blue color). Arrows on the x-axis show values expected from the biophysical analytic model for electrode separation distances at 2mm, 2cm, and 8cm.

the spatial decay predicted by the analytical volume conductor model of Eq. 2b. Figure 2 demonstrates the high correlations often observed between even distant electrodes [14]. The lower portion of the figure shows two sample ECoG signals with high (>0.9) correlation. The electrodes from which these ECoG signals were recorded are highlighted in the upper left portion of the figure and are 63mm apart at opposite ends of a clinical ECoG grid. The upper right side of the figure presents recordings from 32 channels of the grid; high levels of correlation are visible across many channels. The transition from close to distant electrode pairs is presented more completely in Figure 3, which shows the stacked histogram of correlation values for 1cm distance increments. The observed correlations are much higher than anticipated from the simple volume conductor model. This suggests that even given electrodes with low biophysical correlation, as predicted by the volume conductor model, they are likely to still have high neural correlation as observed in recordings at least during a resting baseline conditions.

IV. Discussion

The spatial resolution of a cortical surface electrode array depends not only on the size and spacing of the electrodes, but also the volume of tissue to which each electrode is sensitive. An analytic model was used to predict the volume of sensitivity of cortical surface electrodes of various sizes. The resolution of an electrode at a given depth was primarily a function of electrode size; however changes in electrode size also produce dramatic changes in sensitivity to sources at various depths, making electrode size an important parameter for the signal–to-noise ratio. While having small electrodes provides favorable spatial resolution, it reduces the sensitivity of the electrodes at anatomically relevant depths. Furthermore, it is important to note that, for cortical surface recording, the gain in spatial resolution by reducing electrode size becomes almost negligible once electrode size

is below 1mm (Figure 1). Given the already low signal levels and prevalence of noise from irrelevant brain activity and from recording hardware, some trade-off is likely necessary to achieve robust and specific recordings. Further, the smallest electrodes (150μm in diameter) in Figure 1 show a very steep falloff in its sensitivity to signal sources within 1mm in depth, suggesting the potential for recording spurious (thermal, ionic, glial, etc.) non-neural activity, or only being sensitive to a small fraction of the dendritic arbor. The simple analytic homogeneous volume conductor model used here neglects many known anatomical features of the ECoG recording problem. Other groups, for example Slutzky et al. [8], have used more complex Finite Element Models to draw conclusions about the effect of sub- vs. epi-dural electrode placement and optimal electrode spacing. It is likely that the results presented here would be affected to some degree by a more realistic model, however the general trends should remain constant. Regardless of the effect of a non-zero conductivity skull, or a high conductivity CSF layer, electrode sensitivity will decay quickly with distance, and this decay will be slower for larger electrodes. A more realistic model, including the blurring effects of a CSF layer, a more realistic skull, and so on, will be necessary to facilitate optimization of electrode characteristics and will be investigated in future work.

It is well known that the biophysical correlation, or cross-talk, predicted by volume conductor models (analytic or FEM) is generally small compared to the neural correlation observed in recordings[13-16]. The observed recordings demonstrate this high level of correlation (in time, using the zero-lag cross-correlation). These data also demonstrate that the correlation is maintained within a wide range of electrode separation. The amplitude of the correlated signal makes it unlikely to be solely due to reference choice, and while common average referencing will remove it, this technique may complicate a rigorous analysis of the activity's source. Clearly, this high level of correlation will impact the optimal spacing and resolution of ECoG electrodes, since the goal is to record as much *independent* information as possible. While we observed high correlation in raw ECoG signals in the time domain, it is worth noting that previously researchers have found that high-frequency band activities are spatially more localized as compared to low-frequency band activities [14-17]. Powers of various frequency bands, e.g. sensorimotor rhythm at 10-30 Hz, high-gamma band at 40-200Hz, are found to be informative to underlying neural processes, and they have been used for BCI control. This suggests that a closer look at ECoG signal correlation in the frequency domain also needs to be considered when determining optimal electrode spacing [18], future work will focus on coherence in these BCI-relevant frequencies under resting and during various behavioral or cognitive tasks. Finally, a more thorough neurophysiological understanding of ECoG signals requires a more precise model of the origin and modulation of ECoG signals from cortical sources. This improved understanding will allow better electrode design for clinical brain mapping, neuroscience research, and BCI applications.

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