**SENSOR DENSITY AND HEAD SURFACE COVERAGE IN EEG SOURCE LOCALIZATION**

*Jasmine Song* ⋆  Colin Davey⋆  Catherine Poulsen⋆  Sergei Turovets†  Phan Luu‡  Don M. Tucker‡

⋆Electrical Geodesics, Inc. 500 East 4th Ave., Suite 200, Eugene OR 97401, USA
†Neuroinformatics Center, University of Oregon, Eugene OR 97403, USA
‡Department of Psychology, University of Oregon, Eugene OR 97403, USA

**ABSTRACT**

In research with electroencephalographic (EEG) measures, it is useful to identify the sources underlying the potentials recorded at the head surface in order to relate the EEG potentials to brain function. The EEG recorded at the head surface is a function of how current at specific brain (primarily cortical) locations propagates through the conducting volume of head tissues. The accuracy of source localization depends on a sufficient sampling of the surface potential field, an accurate estimation of the conducting volume (head model), and the inverse technique. The present paper reports the effect of spatial sampling of the potential field at the head surface, in terms of both sensor density and coverage of the inferior (lower) as well as superior (upper) head regions. Several inverse methods are examined, using the four shells spherical head model and the finite difference model. Consistent with previous research, greater sensor density improves source localization accuracy. In addition, across all sampling density and inverse methods, sampling across the whole head surface improves the accuracy of source estimates.

**Index Terms**— EEG, Spatial Sampling, LORETA

1. **INTRODUCTION**

The EEG is recorded at the head surface. Traditionally it is recorded only at scalp regions overlying the brain with the International Ten-Twenty System. The EEG reflects activity generated at some distance away in the cortex, combined with some level of noise (including non-cephalic biological, environmental, and instrument noise). Although the scalp potential data are employed as measurement variables in experimental studies or clinical diagnosis, researchers and clinicians ultimately want to discern the cortical sources of relevant EEG features. It is very important to estimate the accurate sources of EEG sensor signals. The dipolar fields of each brain region propagate in three dimensions, and cannot be assumed to originate directly under the sensor. Activity recorded at any sensor reflects a summation of all active sources in the brain, and the 3D propagation can be counterintuitive, as in the case of a tangential source, for which the sensor lying immediately above will show no activity. Therefore to determine brain sources of EEG potentials it is necessary to analyze objective biophysical models based on the properties of head and brain anatomy.

In biophysical models, current sources in the brain are typically modeled by point dipoles that are equivalent to the summed post-synaptic potentials of all the pyramidal cells in a patch of cerebral cortex. The cortex can be parcellated into discrete source patches such that the activity of the entire cortex can be modeled by a finite set of dipoles, typically several thousands.

The relationship between the current generated by a single dipole (the net current generated by all synchronized post-synaptic potentials in the corresponding patch) and a single scalp sensor is linear. In other words, for a given source dipole and a given location on the scalp, there exists a scalar lead-field value, which is determined by the geometry and conductivities of the head tissues, the location of the dipole, and the location of the sensors. Together these several determining factors are collectively referred to as the head model. The forward problem is calculating the voltages at the scalp, given a configuration of currents at the brain sources. The inverse problem is the source localization of the scalp potentials, which estimates current sources given head surface-recorded voltage data.

As with sampling of time-series data, the Nyquist theorem states that the sampling rate must be twice as fast as the highest frequency to be characterized in order to avoid aliased signals. Whereas temporal sampling is conducted with an analog-to-digital converter for the EEG time series, spatial sampling of the head surface potential field is conducted with the 2D sensor array. Adequate spatial sampling with a dense sensor array is necessary to avoid aliasing of spatial frequency information [1]. An important question, therefore, is what spatial frequency is propagated from the brain to the head surface.

Due to the high resistivity of the skull, it is often thought that the spatial frequency content of the EEG is relatively low, and thus low-count channel montages can be employed. At the same time, it is widely repeated that the EEG and event-related potentials (ERPs) have poor spatial resolution. However, recent studies show that the skull is in fact more conductive than previously assumed, and therefore higher spa-
tial sampling density is required to prevent spatial aliasing. Recording from the adult head surface with a closely-spaced (3 mm) sensor array has shown considerable high spatial frequency content, requiring sensor spacing of 1 cm or less. With whole head coverage in an optimal geodesic pattern and accurate estimates of skull conductivity (skull:brain conductivity ratio 15:1 rather than 80:1), half-sensitivity volume estimation suggests that approximately 500 channels are required for the human EEG. Evidence from simulation as well as clinical data has shown that aliasing is severe and problematic with standard EEG recording (16- and 32-channel) arrays.

While spatial sampling density is important, coverage is also crucial. Often EEG data are obtained only from the top half of the head, due to the assumption that only electrodes adjacent to the brain are needed. This bias in coverage can lead to very poor estimates of activity from inferior and medial sources in the cortex and does not capture the full dipo lar topography of brain activity projected to the head surface. Simulation studies have shown how accurate source estimation can be compromised when the surface potential field is sparsely characterized. These simulation studies have been confirmed with clinical data [2].

This study has been conducted to systematically address the issue of head surface coverage, particularly the typical effect of inadequate coverage of the inferior or lower head, including the face and neck, even though first principles of volume propagation [3] suggest this factor should be critical. The goal of the present study is to examine how source solutions (obtained with several inverse methods such as Minimum Norm and LORETA) are affected by sampling density and coverage of the inferior head-surface potential.

2. METHODS

2.1. Simulated Data

An isotropic spherical model and FDM (finite difference model) were used to generate the simulated data. The model contains four shells to represent the surface, skull, cerebral spinal fluid (CSF), and brain. The conductivity values for the scalp (skin), skull, CSF, and brain were set to 0.44, 0.018, 1.79, and 0.25, respectively [4]. The skull to brain conductivity ratio in this model is 14:1. The locations of the dipoles were derived following the method of Pascual-Marqui [5] by discretizing the gray matter volume of the Montreal Neurological Institutes (MNI) 305-subject average MRI. This resulted in 2447 dipole locations, each with 3 orthogonal orientations (7341 dipoles). Each dipole location covered a 7mm$^3$ volume. The dipole locations were warped to the spherical model to approximate the position of the brain in that model. Because we want to emphasize the practical effects of sampling density and coverage, we employ sensor positions from the Hydrocel Geodesic Sensor Net (HCGSN). Sensor positions were defined by the average positions of the 256-, 128-, 64-, and 32-channel HCGSN. We divided the sensors in each 256-, 128-, 64-, 32-HCGSN in half, such that sensors on the upper-half of the surface represent the typical placement found in conventional, sparse-array montages. Figure 1 shows the positions of the sensors for each channel count, illustrating sampling density as well as coverage. To examine the effect of density and whole-head coverage on the accuracy of source estimates, we compared the source solutions obtained with sensors from the upper half only (scalp) to those obtained from whole-head coverage.

With this spherical head model and FDM, forward projections from each dipole to all sensors generated 7,341 unique surface potential fields for each channel count and coverage. No noise was added to the simulated data; this optimized the inverse recovery for each condition, and it avoided combinatorial expansion of the number of conditions examined.

All inverse solutions employed the same spherical model and FDM used for generation of the simulated data. We examined the Minimum Norm [6] and LORETA [5] inverse estimation methods.

With simulated data based on forward projections, the location of the source for each scalp tomography is known. Therefore, we can quantify the errors in source reconstruction with the localization error distance (LED), which is defined as the Euclidean distance (mm) between the location of the maximum current distribution from the inverse solution and the position of the true generating dipole. Small LED values represent small errors.

2.2. Real Data

The clinical data were from epileptic patients to investigate the spatial sampling effect on localizing the spike onset. The EEG recording was performed using 256 HCGSN. The data were recorded against average references at sampling rate of 250 Hz. Both sensor nets (upper 128 and upper 64) were generated from 256 HCGSN. In each patient, the MR data
Table 1: Clinical features for epileptic patients: L, left; R, right; temp, temporal; fron, frontal; mid, midline; infe, inferior.

<table>
<thead>
<tr>
<th>Patient</th>
<th>Spike Type</th>
<th>EEG spikes</th>
<th>No.</th>
<th>dEEG Source</th>
<th>icEEG</th>
<th>Surgery</th>
<th>Engel</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Lfron</td>
<td>Rtemp</td>
<td>6</td>
<td>Rtemp</td>
<td>Rtemp</td>
<td>Rtemp</td>
<td>I</td>
</tr>
<tr>
<td>2</td>
<td>Lfron</td>
<td>Lfron</td>
<td>76</td>
<td>Mid</td>
<td>Lfron</td>
<td>Lfron</td>
<td>II</td>
</tr>
<tr>
<td>3</td>
<td>Rinfe</td>
<td>Rfron</td>
<td>56</td>
<td>Rtemp</td>
<td>None</td>
<td>Rtemp</td>
<td>I</td>
</tr>
</tbody>
</table>

and solution space was restricted to the gray matter. Solution spaces for the distributed source space contains from 2240 to 2272 cortical patches uniformly distributed over gray matter of the brain and mapped onto the FDM head model. The peak of the spike was used as a center of segments before and after 250 ms. The number of spike segments are ranged from 6 to 76. The segments are averaged to get epoch for each patient. The global field power (GFP) was calculated over the epoch [7]. The time points were chosen to localize the spike onset, which the time point is at the middle of the rising slope of GFP from the beginning of the spike to the peak time point in Figure 3. The averaged EEG data were analyzed using LORETA, which is constrained to the individual cortical surface. Each patient’s clinical feature is summarized in Table 1. Engel is the classification of post-operative outcomes. Class I is the seizure-free. Class II is the almost seizure-free.

3. RESULTS

Figure 2 shows the summary of LEDs for each inverse method, head model, sensor counts and coverage. The results show that as sampling density increases, localization error decreases, regardless of the inverse method and head model. Moreover, the localizations obtained with LORETA show the lowest error across all sampling densities, compared to the MN technique. Whole coverage has less error than upper coverage given the same number of sensors. At the configuration with the greatest density and coverage (256-channels), MN technique shows an average localization error of two dipole positions away (14mm) from the simulated source. The LORETA technique exhibits an average localization error of one dipole position away (7mm) from the simulated source. We fit the linear models for four cases (spherical HM with MN, Spherical HM with LORETA, fdm HM with MN and fdm HM with LORETA).

\[ y = \beta_0 + \beta_1 x_1 + \beta_2 x_2 + \beta_3 x_3 + \epsilon, \]

where \( y \) is the LED, \( x_1 \) is the coverage, \( x_2 \) is the number of sensors and \( x_3 \) is the depth of dipole. For all four cases, there are significantly difference between the LEDs from whole and upper coverages.

Figure 3 shows the source distributions at spike onsets with FDM head model, LORETA and whole-256, upper-128, and upper-64 sensor nets for epileptic spikes. For Patient 1 and 3, right temporal lobe were localized at the spike onset zone using whole-256 head sensor net but not by the upper sampling sensor nets. The right temporal lobe were the surgical region and the post-operative results are free of seizure. For Patient 2, left frontal was not localized at the spike onset by all sensor types. Since the surgical outcome is almost free of seizure, we cannot confirm whether the spike onset zone of Patient 2 is the left frontal lobe.

This shows the benefit of more channel counts and sampling over whole head on the accurate sources.

4. DISCUSSION

Although the importance of adequate spatial sampling of the surface potential field for the accuracy of source estimate is now well understood, there is less evidence on the effect of inadequate sampling of the inferior surface of the head. This is partly due to the fact that the lower head is difficult to sample with electrode caps, which put pressure only on the top surface of the head. It also reflects the misconception that EEG cannot be recorded from more inferior sites, such as the face. Modern sensor placement schemes, such as the five-percent system or the geodesic placement system [1] now include the inferior surface, and practical experience suggests that source localization is improved when the inferior surface of the head is sampled [8]. The present paper reports the first systematic simulations of both sampling density and head surface cover-
There is no difficulty in building the 256-channel nets. EGI has built the high density EEG nets since 1992 and they are widely used in both clinical and research markets. The cost increase is roughly proportional to the number of recording channels.

The results showed that whole-head coverage improves source localization accuracy. Whole-head coverage reduces the superficiality bias associated with the MN. When the whole surface of the volume (head) is sampled, the surface integral potential of each dipole will be zero. Because the MN finds the minimum energy, they will distort solutions toward superficiality in the case of top head but not for whole head. When the inferior surface is sampled, the errors associated with very deep sources are attenuated. The benefit of whole-head coverage is also relevant to source solutions obtained with LORETA technique that addresses the superficial bias inherent to MN. The source solutions obtained using LORETA show that although the superficial bias is dramatically attenuated, the addition of data from the inferior surface also improves the accuracy of source estimates for very deep sources.

In this paper, we established a baseline of accuracy that can be obtained under ideal conditions (e.g., noise-free data and perfect match between forward model and data generation model) and applied it to the clinical data. We demonstrated that dense, whole-head sampling provides the most accurate description of the surface voltage field, leading to more accurate source localization. With this as a starting point, future investigations can now move on to systematically address the effects of noise and model mismatch on the accuracy of EEG source localization.

5. REFERENCES


