VIBROEEG: IMPROVED EEG SOURCE RECONSTRUCTION BY COMBINED ACOUSTIC-ELECTRIC IMAGING

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ABSTRACT

Electroencephalography (EEG) is the electrical neural activity recording modality with high temporal and low spatial resolution. Here we propose a novel technique that we call vibroEEG improving significantly the source localization accuracy of EEG. Our method combines electric potential acquisition in concert with acoustic excitation of the vibrational modes of the electrically active cerebral cortex which displace periodically the sources of the low frequency neural electrical activity. The sources residing on the maxima of the induced modes will be maximally weighted in the corresponding spectral components of the broadband signals measured on the noninvasive electrodes. In vibroEEG, for the first time the rich internal geometry of the cerebral cortex can be utilized to separate sources of neural activity lying close in the sense of the Euclidean metric. When the modes are excited locally using phased arrays the neural activity can essentially be probed at any cortical location. When a single transducer is used to induce the excitations, the EEG gain matrix is still being enriched with numerous independent gain vectors increasing its rank. We show theoretically and on numerical simulation that in both cases the source localization accuracy improves substantially.

***Index Terms*—** EEG, source localization, vibrations

1. INTRODUCTION

Electroencephalography (EEG) is a noninvasive measurement technique of the electrical potentials over the scalp aiming at reconstructing the underlying primal electrical activity of the brain cortex. Due to its high temporal resolution, EEG is a valuable for the diagnosis of neural conditions (epilepsy being a bold example) as well as a research tool [1]. Its major downside is the low spatial resolution, caused in part by the poorly known electrical conductivity distribution within the head (the forward model uncertainty) and by the sharp discontinuity in the conductivity on the brain/skull and skull/scalp interfaces which in turn causes spatial smearing of the electrical signals over the surface of the scalp. This effectively limits the number of independent electrodes to around 100. Yet, since about 1 cm^2 of cortex can generate electrical signal visible on the scalp [1], the number of possible spatially distinct electrical sources exceeds 10,000. This enormous difference makes the EEG source reconstruction problem severely ill-posed. Different regularization techniques stabilize the process at the expense of unreasonable simplification of complex real world neural activity patterns, for example, assuming sparse or minimal energy activations [2]. Recently, a number of techniques have emerged which intentionally perturb the electrical signal resulting from the cortical activity in order to improve the reconstruction process. In one of such methods, the so-called acousto-electric technique (AET) for current density imaging, focused ultrasound is used to perturb locally the conductivity of the medium providing localized information on the current flow [3]. The focused ultrasound though possible is cumbersome to generate within the brain [4]. In another method known as magneto-acoustic imaging (MAI) of bioelectric currents, the goal is to measure neural activity directly employing the Lorentz force originating from electrical activity in a strong static magnetic field [5]. However, the resulting signals are feeble relative to the noise for the realistic parameters of the system. Here, we propose vibroEEG - an alternative brain imaging technique relying on the combination of electric and acoustic effects and aiming at significantly improving the reconstruction process of primal cortical activity. We propose to excite the vibrational modes (not necessarily with focused ultrasound) of the electrically active brain cortex and record the resulting modulated voltage signal arising on the scalp due to the oscillating position of the neural sources. The cerebral cortex has resonant modes of the order of hundreds of Hz [6] that can be generated locally and controllably [7, 8] while the EEG signal itself is band-limited at about 100Hz. This arrangement permits probing the neural activity locally both in space and in time. Moreover, it allows to substantially increase the signal-to-noise-ratio (SNR) of the system due to spectral and spatial localization of the vibroEEG signal.

The rest of the paper is organized as follows: in Section 2 we present the theory of vibroEEG. In Section 3 the theory is validated by numerical examples based on finite element modeling of the vibrational modes of the cerebral cortex (CC) and the boundary elements modeling of the resulting vibroEEG gain matrix. The example of employing the rich internal geometry of the CC by vibroEEG for close sources separation is followed by a general spatial resolution analysis of the regular EEG and vibroEEG methods. Section 4 concludes the paper with a discussion of possible applications, limitations and extensions of the method.

2. THEORY

2.1. EEG forward and inverse problem

Electric potentials measured by EEG is produced chiefly by the electrical activity of the pyramidal neurons within the CC [1]. These are modeled as current dipoles (a closely positioned current source - sink pair of equal amplitudes) normal to the CC surface. On the macro scale, the head is modeled as consisting of regions of constant electrical conductivity σ. It can be shown that the primary current distribution $\mathbf{j}(\mathbf{r})$ gives rise to the electrical potential $v(\mathbf{r})$ according to the Poissons equation:

$$\mathrm{div}\ (\sigma \nabla v) = \mathrm{div}\, \mathbf{j}. \tag{1}$$

Consider now the discretized setting with S possible source locations $\mathbf{r}_1, \ldots, \mathbf{r}_S$ and N possible electrode locations $\mathbf{r}'_1, \ldots, \mathbf{r}'_N$ on which the resulting electrical potential is measured. We assume the current dipoles are perpendicular to the cortex and oriented along the normal vectors $\mathbf{n}_i = \mathbf{n}(\mathbf{r}_i)$, where $\mathbf{n}(\mathbf{r})$ denotes the normal vector at point $\mathbf{r}$. The linearity of Poisson's equation implies the existence of a linear transformation between the current intensity of the sources, $\mathbf{I} = (||\mathbf{j}(\mathbf{r}_1)||, \ldots, ||\mathbf{j}(\mathbf{r}_S)||)^{\mathrm{T}}$ and the vector of measured potentials, $\mathbf{V} = (v(\mathbf{r}'_1), \ldots, v(\mathbf{r}_N))^{\mathrm{T}}$, which can be expressed as

$$\mathbf{V} = \mathbf{G}\mathbf{I} + \mathbf{E} \tag{2}$$

where $\mathbf{G}$ is the $N \times S$ gain matrix, and $\mathbf{E}$ is the measurement noise. The forward EEG problem consists of finding $\mathbf{V}$ given $\mathbf{I}$ and can be essentially captured as calculating the gain matrix $\mathbf{G}$. The inverse EEG problem aims at finding $\mathbf{I}$ given $\mathbf{V}$ [2]. The rank deficiency of $\mathbf{G}$ makes the problem hugely under-determined, a fact that is frequently compensated for by imposing regularization on $\mathbf{I}$. As an alternative, here we propose to enrich $\mathbf{G}$ with new independent measurements by modulating the EEG signal with CC vibrations.

2.2. Vibrations of the cerebral cortex

The interest in vibrational analysis of the human head is motivated mostly by brain injury prediction in an accident [6]. The CC consists of tissue having finite elasticity and its surface contains multiple convex and concave parts (called gyri and sulci). This forms a spring-like structure with multiple resonant frequencies. If harmonic force of a certain resonant frequency is applied to the CC, the later responds with mechanical oscillations of a specific to this frequency shape (i.e., distribution of amplitudes, phases and directions of the oscillation on the surface) specific to that frequency. The spatial components of such resonant frequency oscillations are called the vibration or resonant modes of the CC. The lowest resonant frequencies of the CC are of the order of hundreds of Hz and corresponding to spatially unlocalized modes. Higher resonant frequencies are characterized by more localized modes roughly corresponding to the resonant modes of the individual sulci and gyri. The vibrational modes in the CC can be induced by attaching a single transducer to the head or by using a phased array of transducers to more effectively generate localized vibrations [8]. Similar to vibroacoustgraphy, the imaging modality determining the tissues mechanical properties from their vibrations, high frequency ultrasound (typically 2MHz) can be modulated by low frequency (typically 1kHz) to induce low frequency tissue vibrations in the focal point of the phased array [7]. These vibrations can be monitored using MRI similar to the MR-elastography technique [9].

2.3. VibroEEG

Vibrations of the electrically active CC will periodically displace the current sources residing on it modulating the EEG signal. The vibroEEG signal is defined as the modulation of the regular EEG signal due to the vibration of the electrically active cortex and, at least conceptually, resembles regular amplitude modulation (AM). Let us denote by $\mathbf{G}(\mathbf{r}, \mathbf{r}')$ the fundamental solution of the EEG forward problem at point $\mathbf{r}'$ due to a unit current dipole placed normal to the cortex at point $\mathbf{r}$. Perturbation analysis of Poisson's equation shows that the change of the EEG signal due to a small displacement of the source would be given by the scalar product of the gradient of $\mathbf{G}(\mathbf{r}, \mathbf{r}')$ with respect to the second argument with the displacement vector. The overall vibroEEG signal resulting from cortical activity $\mathbf{I}(\mathbf{r})$ undergoing displacement $\mathbf{d}(\mathbf{r})$ is given by the integral

$$v(\mathbf{r}) \quad = \quad \int \nabla_{\mathbf{r}'} \mathbf{G}(\mathbf{r}, \mathbf{r}') \cdot \mathbf{d}(\mathbf{r}')\, \mathbf{j}(\mathbf{r}') da(\mathbf{r}'), \tag{3}$$

where the integration is performed over the entire cortex surface and $da(\mathbf{r})$ denotes the area element on the cortex at point $\mathbf{r}$.

Demodulating the acquired vibroEEG signal into its spectral components produces independent measurements that effectively increase the rank of the gain matrix $\mathbf{G}$. Moreover, for resonant frequencies of different CC folds the activations can be essentially probed directly, leading to a set of approximately independent inverse problems for each fold.

3. NUMERICAL STUDY

In what follows, we present a numerical simulation study assessing the advantages of the proposed modality. In order to obtain the vibroEEG gain matrix it is sufficient to calculate the vibrational modes of the CC and the gradient of the EEG forward problem. The forward EEG problem was calculated by boundary element method (BEM). The head model we used was an averaged MNI model provided by Brainstorm toolbox [10]. To calculate the vibrational modes of the CC the mesh of 557478 faces and 278743 vertices of the MNI model provided by Freesurfer toolbox was used. As a simplification we decoupled the CC from its surrounding. The mechanical parameters we used were based on [6]: Youngs modulus 0.497 MPa, Poissons ratio 0.48 and density 1.14 g/cm^3. In order to determine the modal responses, finite element (FE) analysis was performed using Dassault Abaqus v6.1.

In this experiment, we evaluated the separation of spatially close sources residing on different gyri using vibroEEG. Close deep sources are indistinguishable in regular EEG. However, the small Euclidean distance between the sources does not imply small distance on the cortex surface, where the geodesic distance provides a more meaningful measure of proximity. This distinction can lead to very different functionalities of the sources and can be very important in brain functional imaging.

Figure 1 (first row) visualizes vibration modes 327 and 368 corresponding to resonant frequencies of 3089Hz and 3275Hz. These modes concentrate vibration energy in two adjacent folds of the cortex. For the sources situated on these folds and being separated by a relatively large geodesic distance the Euclidian distance is very small (about 2mm) making them indistinguishable on EEG. However, as can be seen in Figure 1 (bottom rows), the vibroEEG signal for these sources varies substantially for the appropriate frequencies. In this way the internal geometry of the cortex manifests itself for the close sources separation using vibroEEG technique.

In previous example we set the maximal vibration amplitude to 1mm which can be harmful especially in continuous regime (though worthwhile for cases such as planning surgery for epilepsy). However, the amplitude can be reduced substantially while maintaining reasonable SNR if repetitive recordings are considered (for regular EEG averaging of hundreds of triggered by external stimulus repetitions is routinely used). For $3kHz$ the EEG spectral density is negligible and the main source of noise is the amplifier with achievable spectral noise density of $5nV/\sqrt{Hz}$. Next we assume that after the concatenation of multiple triggered epochs the effective length of the concatenated vibroEEG signal sums to $1s$. Accordingly the effective bandwidth of such a signal is about $1Hz$. Hence the instrumental noise RMS amplitude can be reduced to $5nV$. If instead of $1mm$ amplitude the $10\mu m$ amplitude is used the amplitude of the vibroEEG signal would be $30nV$ (instead of $3\mu V$) which is still 6 times above the noise level. Increasing effective signal temporal length permits to further reduce the vibration amplitude while maintaining detectable signal level.

In the following experiment, we explore the impact of vibroEEG modulation on the rank of the gain matrix. First the high resolution cortex mesh was resampled to 12000 vertices. The electrical potential was measured at 65 vertices of the head mesh corresponding to the standard $10-20$ EEG electrodes placement [10]. Interpolating the vibration modes to the rough cortex mesh the vibroEEG gain matrices for first 400 modes were calculated and concatenated. The singular values of the obtained matrix are depicted in Figure 2 in comparison to the singular values of the EEG matrix. The superiority of vibroEEG is clearly visible: 65 EEG singular values span the range of $0.02--1$ as opposed to 320 singular values in the case of vibroEEG. In other words, with some abuse of notation, one can say that the vibroEEG gain matrix is about five times better conditioned.

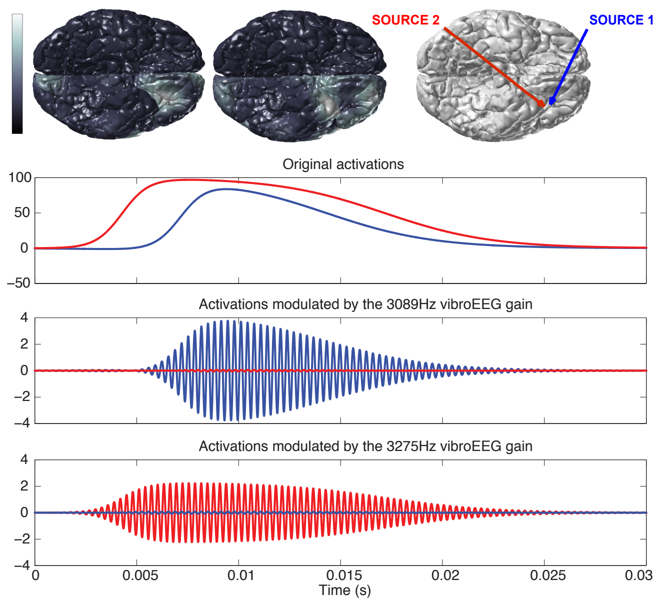

Fig. 1. First row: vibrational mode of the CC for frequencies 3089Hz (left) and 3275Hz (middle), and the locations of the simulated sources on the CC (right). Second row: simulated activations of the sources (in nAm). Third and fourth rows: vibroEEG-moduldated source voltages (in microvolts) recorded at the 'Fp1' electrode for both frequencies.

4. DISCUSSION AND CONCLUSION

We have presented vibroEEG , a new combined acoustic-electric modality for brain imaging. The method in general enriches the regular EEG measurements with new information based on the change of the electrophysiological signal as the result of the externally induced cortex vibrations. The

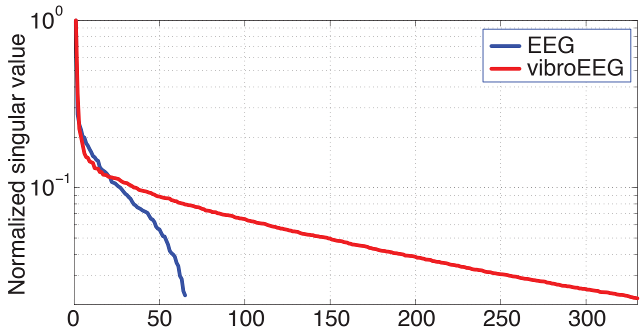

Fig. 2. Singular values corresponding to the vibroEEG gain matrix (red) based on 400 modes and sampled at 65 points of the $10-20$ system vs the singular values of the corresponding EEG gain matrix (blue).

pattern of the induced vibration can be monitored via MR imaging as described in [9]. Not only does this feedback allows to induce the vibrations controllably, it compensates for the imprecise forward model: if a spatially localized source is active on a certain vibrating fold vibroEEG signal should be much stronger compared to the case when the said fold is inactive.

An important issue requiring further study is the impact of the vibrations on the brain electrical activity itself, i.e., acoustic neuromodulation. Such an impact has been previously reported in the literature [11]. Obviously, not every vibration influences the neural activity; as a negative example one can argue that mental activity hardly changes while riding a train. The detailed mapping of the acoustical intensity threshold for appreciable neuromodulation effects in different brain regions over large frequency ranges is presently an active research area [4].

Instead of inducing the vibration externally, it also seems possible to employ the mechanical oscillations of the brain tissue due to blood/CSF pulsation related to the cardiac cycle. While the main power of blood pulsation is concentrated at a relatively low frequency range [12], it is possible to average EEG signal with ECG gating. Information about brain tissue displacement can be obtained from cardiac-gated MRI of the brain. Such "cardio-vibroEEG" approach can possibly be relatively easy to implement as it requires no hardware modifications in the traditional EEG setup. Yet, as the internal pulsation spectrum overlaps with the EEG spectrum, the more involved signal processing will be required. This research direction is left for future studies.

5. ACKNOWLEDGEMENTS

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