Localized Cortical Dipole Imaging using a Small Number of Electrodes Based on Independent Component Analysis

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Abstract—**The spatial resolution of scalp potential mapping is limited because of low conductivity of a skull. Cortical dipole layer imaging has been proposed as a method to visualize brain electrical activity with high spatial resolution. According to this method, about 100 electrodes were required to measure whole brain electrical activity. In the present study, we investigated simplified cortical dipole imaging with a small number of electrodes. The density of electrodes and the spatial resolution are in a trade-off relation. Thus, the number of electrodes was reduced by limiting the visualization region of interest, without lowering the density of electrodes. Moreover, independent component analysis was applied to the multiple signal sources to extract an attention signal from the other signals and noise. In simulation, even if the number of electrodes was reduced to 25, the obtained results were almost equivalent to the case with whole brain electrodes. The proposed method was applied to human experimental data of movement-related potential. We confirmed that the proposed method provided high resolution cortical dipole imaging with localized distribution.**

I. Introduction

Noninvasive electroencephalogram (EEG) recordings with low restriction on the measurement environment are effective to analyze brain function in daily life. However, the spatial resolution of the EEG data is limited because of low conductivity of a skull. Cortical dipole imaging that estimates the equivalent dipole source distribution on a virtual layer within a brain has been proposed as a method to visualize brain electrical activity with high spatial resolution [1], [2]. According to this method, brain electrical activity is represented without being restricted in the number and the direction of the dipole sources. We have investigated several spatial inverse filters to solve the EEG inverse problem [3]-[6]. The parametric projection filter that incorporated statistical noise information was effective for high quality imaging [3], [4]. In cortical dipole imaging, about 100 electrodes were required to visualize whole brain electrical activity in high precision. Such EEG measurements took a lot of time for preparation and setting and it was a burden on the subject and the experimenters. Moreover, the cost of multi-channel amplifier is too high to use in daily life.

In the present study, we investigated simplified cortical dipole imaging with small number of electrodes. The density of electrodes and the spatial resolution of the images are in a trade-off relation [7]. The scalp potential map should be sampled with sufficiently short distance between electrodes. Thus, the number of electrodes was reduced by limiting the visualization region of interest, without lowering the density of electrodes. Moreover, independent component analysis (ICA) [8] was applied to the EEG signals to extract an attention signal from the other signals and noise. In simulation, the region of interest was limited at the motor areas of the brain considering the application of brain-computer interface (BCI). In human experiments, the proposed method was applied to movement-related potential (MRP) of fast repetitive finger movement protocols which is preliminary study for motor imaginary BCI [9]-[12]. We confirmed that the proposed method provided high resolution cortical dipole imaging with localized distribution in simulation and experimental studies.

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II. Method

A. Cortical Dipole Imaging

In the present study, the cortical dipole imaging was used to accomplish high resolution brain imaging. In this method, the head was modeled by a volume conductor approximated by the inhomogeneous three-concentric spheres [1]. This head model consists of the scalp, the skull, and the brain with different conductivities. An equivalent dipole layer consists of radial dipoles is established to represent the brain electrical activity inside of the brain. The detail of the cortical dipole imaging technique is described in [2].

The observation process of brain electrical activity on the scalp surface shall be defined by the following equation:

$$\mathbf{g} = \mathbf{A}\,\mathbf{f} + \mathbf{n} \tag{1}$$

where $\mathbf{f}$ is the vector of the equivalent source distribution on a dipole layer, $\mathbf{n}$ is the vector of the additive noise and $\mathbf{g}$ is the vector of scalp-recorded potentials. $\mathbf{A}$ is the transfer matrix from the equivalent cortical dipole sources to the scalp potentials measured with surface electrodes. The estimated source distribution of the dipole layer $\tilde{\mathbf{f}}$ is represented by

$$\tilde{\mathbf{f}} = \mathbf{B}\,\mathbf{g} \tag{2}$$

where $\mathbf{B}$ is the spatial filter that demonstrates the inverse function of $\mathbf{A}$. Several inverse techniques have been proposed to solve Eq. (2). In the present study, we paid attention to the parametric projection filter (PPF) described in II.B.

B. Spatial Inverse Filter

The PPF, which allows estimating solutions in presence of information on noise covariance structure, has been introduced to solve the inverse problem [3], [4]. The PPF is given by

$$\mathbf{B} = \mathbf{A}^{T}\,(\mathbf{A}\,\mathbf{A}^{T} + \gamma\,\mathbf{Q})^{-1}. \tag{3}$$

where $\mathbf{A}^{T}$ the transpose matrix of $\mathbf{A}$, $\mathbf{Q}$ the noise covariance

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matrix derived from the expectation over the noise ensemble E[$\mathbf{n}$ $\mathbf{n}^T$], and γ a small positive number known as the regularization parameter. The determination of the value of parameter γ is left to the subjective judgment of the user. We have developed the parameter estimation method by minimizing the approximation of the square error between actual and estimated dipole distributions [3]. In a clinical and experimental setting, the noise covariance was estimated from data that is known to be source free, such as pre-stimulus data in evoked potentials [3]. Alternatively, the noise covariance matrix was estimated by applying ICA to the scalp potentials [5], [6]. That is, the scalp potential was separated into the signal and the noise components according to a priori physiological information. To extract the attention signal, the other signal components and noise components were assumed as noise. Detail is described in II.D.

C. Localized Cortical Dipole Imaging

To reduce the number of electrodes, the imaging area was restricted at the limited area. In this case, the density of the electrodes was as same as that of the original electrode arrangement. We assumed that a priori information on the broad region of interest was already known from the localization of brain function in clinical study. The electrodes were located around the region of interest. For example, if we aim to discriminate right-hand movement, the attention area is set to motor cortex in the left hemisphere. The optimum number of electrodes may be estimated based on Akaike's information criterion in the preliminary simulation study that considering both the error and simplification of the model.

D. Independent Component Analysis

In the case of single dipole source, the localized cortical dipole imaging may perform effectively. However, in the case of multiple dipole sources, the signal from the other dipole sources located outside of the limited area affects to the attention signal. In such cases, each component was separated using ICA. ICA extracts independent sources from the observed signal based on statistical independence of the original signal. Equation (3) requires adequate noise covariance to improve the restorative ability. By calculating the noise covariance adequately using ICA, signal source would be extracted from noisy observed signals. In the present study, FastICA algorithm proposed by Hyvarinen was used for estimating the ICA [8]. This algorithm is based on a fixed-point iteration scheme maximizing non-Gaussianity as a measure of statistical independence.

The noise covariance for the spatial inverse filter was estimated by the following method (Fig.1). The number of independent components was empirically estimated from a priori physiological information. Moreover, separated independent components were divided into the signal and the noise components according to the anatomical knowledge of the spatiotemporal distributions. By applying a mixing matrix, we obtained the separated signal and noise components:

$$\mathbf{s}_1, \mathbf{s}_2, \ldots, \mathbf{s}_N, \mathbf{n}_1, \mathbf{n}_2, \ldots, \mathbf{n}_M \tag{4}$$

where $\mathbf{s}_i$ (i = 1, 2, ..., N) and $\mathbf{n}_i$ (i = 1, 2, ..., M) are the signal and noise components, respectively and N and M are the number of signal and noise components, respectively.

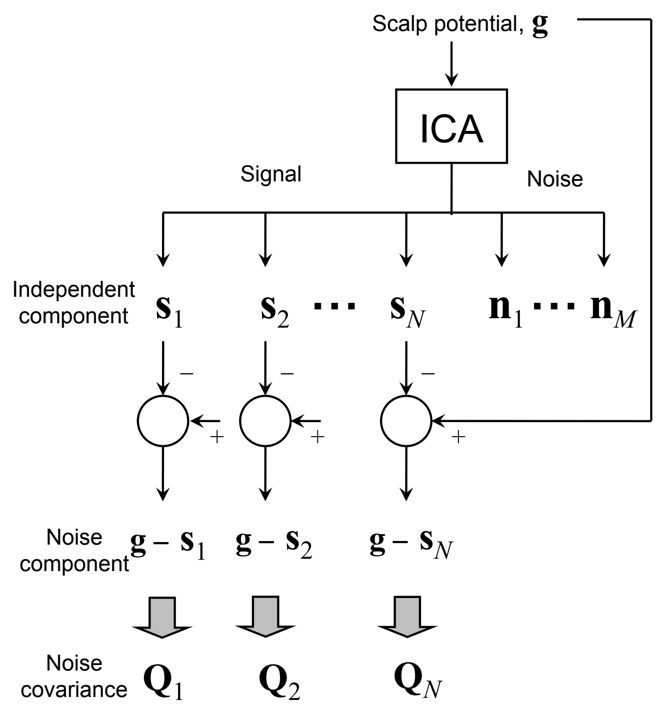

Figure 1. Estimation of noise covariance matrices incorporated with the inverse filters. Separated signal and noise components with ICA were used for localized cortical dipole imaging.

From the results of the previous study [5], [6], the differential noise between the original EEG signal and the separated signal using ICA was suitable for the calculation of the noise covariance matrix to improve the performance.

The differential noises $\mathbf{n'}_i$ (i = 1, 2, ..., N) between the scalp potential and separated signal source were calculated for each dipole sources.

$$\mathbf{n'}_i = \mathbf{g} - \mathbf{s}_i \tag{5}$$

Thus, the noise covariance matrix was calculated by

$$\mathbf{Q}_i = \mathrm{E}[\, \mathbf{n'}_i \, \mathbf{n'}_i^{\mathrm{T}}] \tag{6}$$

The spatial inverse filter for each signal source was constructed using this noise covariance matrix derived from Eq. (6). The estimated dipole distribution for single source $\mathbf{s}_i$ is derived by

$$\tilde{\mathbf{f}}_i = \mathbf{B}(\mathbf{Q}_i)\, \mathbf{g} \tag{7}$$

In this method, the components without the attention signal were assumed as noise. If several dipole sources exist simultaneously, individual signal component may be separately extracted from the mixed EEG signals using this noise covariance-based inverse filter. That is, one dipole source would be extracted by each estimated dipole distribution while the noise and other signal sources were reduced. In other words, the optimized inverse filters can be designed for each signal source component.

III. Experiments

A. Simulations

The present simulation studies were conducted using the inhomogeneous spherical volume conductor head model. 128 electrodes uniformly distributed over the upper hemisphere were used in the simulation. A dipole layer with 1280 radial dipoles at a radius of 0.7 was used to represent the cortical electrical activity.

Figure 2 (a) shows actual dipole distribution for two radial dipole sources with the eccentricity of 0.6. This simulation represents multiple localized brain electrical sources. We paid attention to the left dipole source among these signal sources. Fig.2 (b) shows 128-channel simulated EEG signals observed on the scalp surface. The scalp potential was contaminated with Gaussian white noise with the noise level of 0.2. Electrode arrangements for whole brain (128 channels) and limited area (25 channels) were shown in Figs.2 (c) and (d), respectively. If 128 electrodes were used for whole brain, two signals can be clearly observed in the estimated cortical dipole distribution (Fig.2(e)). However, if the electrodes are reduced to limited area, the estimated dipole distribution was distorted under the influence of the inattention signals and noise (Fig.2 (f)). As a result of signal extraction by applying the ICA, the dipole distribution was improved even if the number of the electrodes was reduced to 25 channels (Fig.2 (h)).

B. Human Experiments

A right-handed female normal subject with age of 24 years took part in the experiments. Experiments were performed after informed consent was obtained according to the Institutional Review Board of University of Illinois at Chicago. The subject performed fast repetitive finger movements which were cued by visual stimuli on the display. 30 second blocks of 2 Hz thumb oppositions for right or left hand were recorded, During movement, the subject was instructed to avoid eye blinks, swallowing, or any movement other than the required finger movements. 94-channel scalp potentials were recorded using multichannel EEG system (NeuroScan Lab, TX). A/D sampling rate was 250Hz. All data were visually inspected, and trials containing artifacts were rejected. After the EEG recording, the electrode positions were digitized using a 3D localization device with respect to the anatomic landmarks of the head. About 450 single epochs were averaged using EMG-locked trigger.

Cortical dipole imaging for the MRPs was analyzed during the period of the motor field. That is, the time point with the highest activity of the scalp potential waveform was determined to be around 50ms after the peak of EMG. Figure 3 displays the estimated results of the cortical dipole imaging for right hand movement. Note that the dipole layer distributions estimated by means of our methods are well-localized as compared with blurred scalp potential maps. Especially, more localized distribution was obtained by limiting the measurement electrodes to 25 channels (Fig.3 (e)). The localized areas for the motor field were located in the premotor cortex, which is consistent with the hand motor representation. Most activities of the source in right-hand movement in the period of the motor field covered the precentral sulcus.

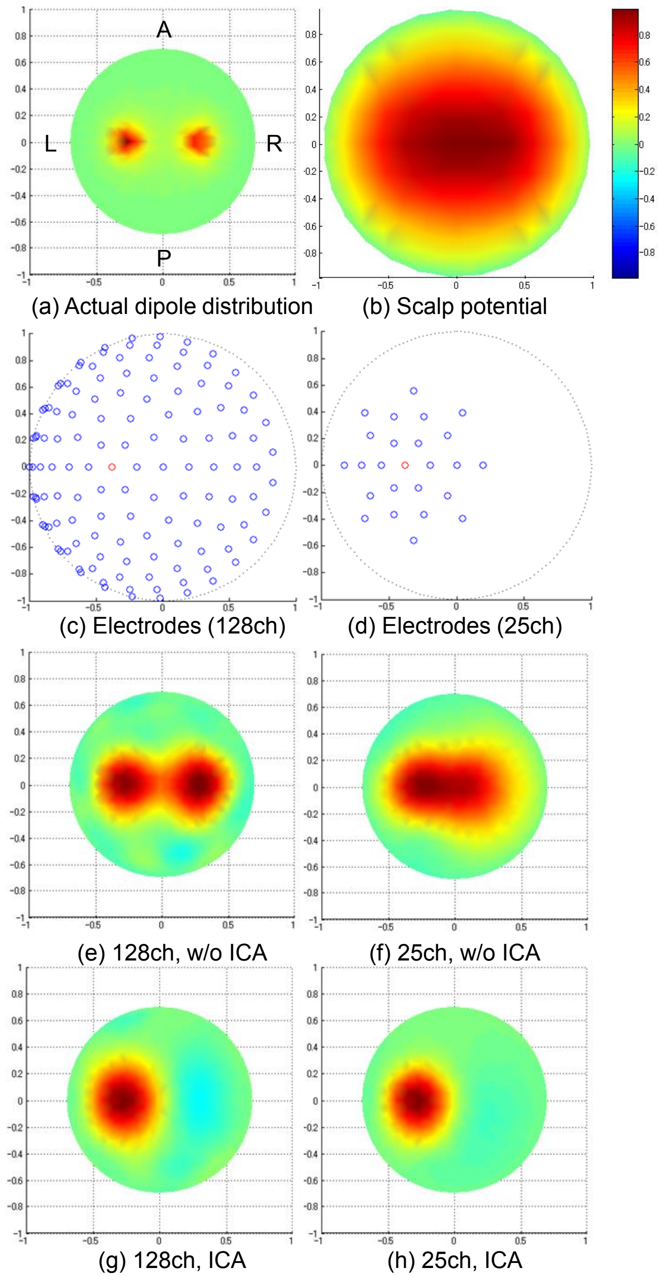

Figure 2. Estimated results of coritical dipole distributions for two dipole sources (e)-(h). In this case, we paid attention to the dipole source in left hemisphere using limited electrodes and ICA.

IV. Discussion

Noninvasive cortical dipole imaging is very effective method to visualize brain electrical activity with high spatial resolution. However, it has disadvantage that many electrodes must be attached to the subject's head. The set up for measurement should be simplified in order to apply the cortical dipole imaging to daily use such as BCI. We attempted to reduce the electrodes effectively. From the previous study, we confirmed that the imaging quality was

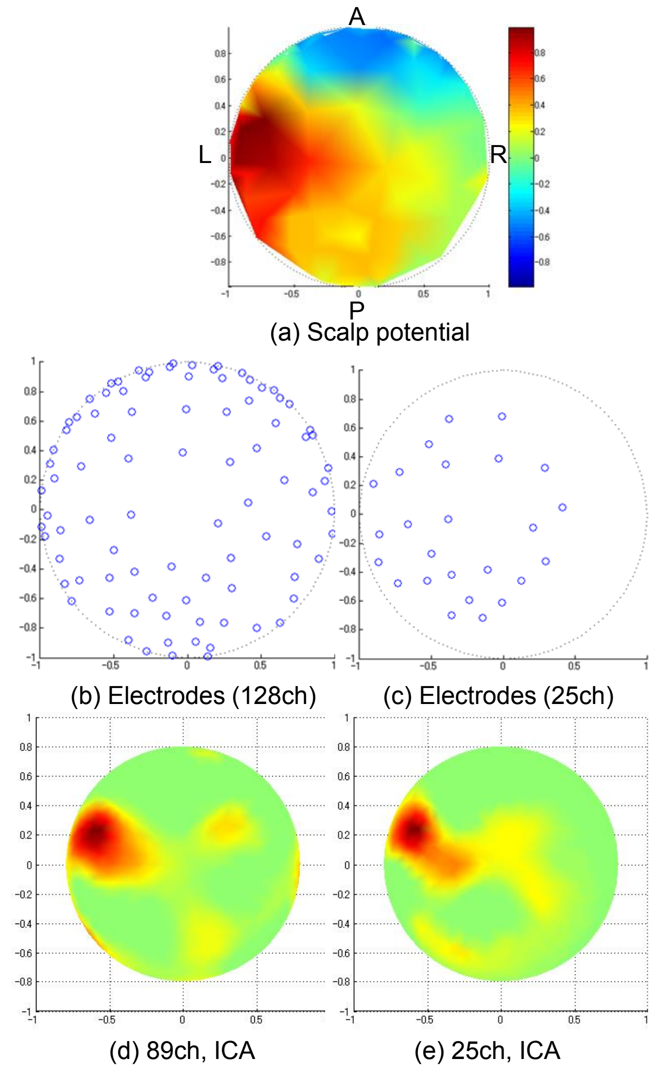

Figure 3. Estimated results of cortical dipole distributions for the MRP (d), (e). Estimated dipole distribution using 25 electrodes was more localized at premotor cortex than that of 89 electrodes.

degraded by spatially thinning out the electrodes. Thus, the number of measurement electrodes was reduced by limiting the observation region to the notice area.

As shown in Fig.2 (f), if several dipole sources exist simultaneously, the cortical dipole distribution was distorted by limiting the electrode arrangement. This distortion caused by the influence from other signals outside of the observation region. In such cases, ICA was applied to extract the attention signal source from other inattention signals and noise. As a result, we could obtain localized dipole distribution as shown in Fig.2 (h) even if the number of electrodes was reduce to 25. It was considered that 25 electrodes is limited number for localized imaging of brain electrical activity. The number of electrodes should be reduced without imaging by considering the application in daily life.

The present experimental study demonstrates that if the subject moves her right-hand, the motor cortex of left hemisphere was activated at about 50ms after the EMG peak (Fig.3(e)). In contraction, we confirmed that left-hand movement derived activation in right hemisphere. These results indicate that contralateral predominant activity of the motor field would occur after the hand movements, which extends previous evidence supporting a hemispheric functional asymmetry of motor control demonstrated by functional magnetic resonance [13] and positron emission tomography studies [14]. In order to indicate the activated location precisely, we are planning to expand our method to realistic-shaped head model in near future. Moreover, we are going to add more number of subjects to validate the proposed method.

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