Improved Detection of Deep EEG Sources: Optimization of Weighted Multielectrode Leads

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Abstract. Weighted multielectrode EEG leads can be applied to improve the signal-to-noise ratio (SNR) of EEG generated deep in the brain. In an optimal multielectrode lead the sensitivity distribution is uniform within the brain volume. When the head is modelled with a three-layer spherical model and the electrodes are uniformly distributed over the scalp surface, the weights of individual unipolar leads can be calculated based on the electrode coordinates. When the head is modelled with a realistically shaped volume conductor model, an optimization method is needed to define the weights. In the present study the purpose was to study such an optimization method in a three-layer spherical head model. As a measurement data we had brainstem auditory evoked potentials (BAEPs) measured with a 124-channel EEG cap. We calculated the SNRs obtained with different multielectrode EEG leads. The results show, that the SNR can be slightly improved, when the weights are calculated with the optimization method. It can be concluded that the optimization method can be applied in defining the weights of multielectrode EEG leads.

Keywords: multielectrode EEG leads, lead field theory, sensitivity distribution, deep source, signal-to-noise ratio

1. Introduction

The purpose of the present paper is to study the performance of multielectrode lead method, which has been developed to improve the detection of deep EEG sources [Väisänen and Malmivuo, 2007]. In a multielectrode lead, several electrodes are applied to form a new single measurement lead. We have evaluated this method with preliminary experimental measurements in [Väisänen and Malmivuo, 2007]. The basis of the method lies in optimization of the sensitivity distribution of the multielectrode lead. Two factors improve the signal-to-noise ratio (SNR) in a multielectrode lead: 1. the lead field of a multielectrode lead is more specific in measuring signals generated deep in the brain, and 2. spatial averaging of noise occurs when several electrodes are applied in the synthesis of a multielectrode lead.

The improved detection of deep EEG sources is important especially in evoked potential (EP) studies. Such EPs include brainstem auditory evoked potentials (BAEPs), which originate in the brainstem and midbrain. The traditional bipolar measurement set-up is relatively insensitive to such sources and thousands of epochs are thus traditionally averaged to obtain a reasonable SNR. Great interest therefore attaches to the development of new methods, which would improve the SNR of signals generated deep in the brain.

The main interest here is to evaluate the performance of an optimization method, which is applied to define the weights for multielectrode EEG lead. When the head is modeled with a three-layer spherical head model, and the electrodes are uniformly distributed over the scalp surface, the optimal weights can be calculated based on the electrode coordinates. If the head is modeled with a realistically shape head model, new methods are needed to define the optimal weights. In the present study the head is modeled with a three-layer spherical model and 124 electrodes are located on the scalp surface based on realistic electrode locations.

In the present study, we evaluated with experimental measurements the performance of the multielectrode lead technique, when the weights are defined with the optimization method. As an experiment we choose to measure BAEPs with 124 electrodes. The performance of the multielectrode lead was evaluated by comparing its SNR to that of traditional bipolar BAEP leads.

2. Material and Methods

2.1 Lead field and reciprocity theorem

The lead field theory can be applied to calculate the sensitivity distributions of different EEG leads. The lead field in the volume conductor can be calculated by feeding a unit reciprocal current I_r to the lead. The reciprocal current gives rise to a current density field $\overline{J}_L$ in the volume conductor. This current density field is the lead field, which is the sensitivity distribution of the lead.

Equation (1) describes the relationship between the measured signal V_L in the lead, the current sources $\overline{J}^i$ and the lead field $\overline{J}_L$ in the volume conductor, when a reciprocal unit current I_r is applied to the lead [Malmivuo and Plonsey, 1995].

$$V_L = \frac{1}{I_r}\int \frac{1}{\sigma}\overline{J}_L \bullet \overline{J}^i dv \tag{1}$$

In Eq. (1) σ is the conductivity of the volume conductor.

An ideal sensitivity distribution is such that its magnitude is concentrated at the source location and its orientation is parallel to the source dipole moment. The sensitivity distribution of bipolar EEG lead is not optimal for measuring signals generated deep in the brain, because the sensitivity is concentrated on the cortex [Malmivuo and Suihko, 2004]. When the measurement electrodes are located on the surface of the scalp, it is impossible to obtain a sensitivity distribution where the magnitude is concentrated at the center of the brain. We have assumed that the best sensitivity distribution realized for deep sources is the uniform sensitivity distribution. By uniform sensitivity distribution, we mean that its magnitude and orientation are uniform throughout the source volume.

We have previously shown how the uniform sensitivity distribution can be obtained in a spherical three-layer head model [Väisänen and Malmivuo, 2007]. To obtain a uniform lead field in the direction of the positive z-axis, the reciprocal current applied to each electrode needs to be incrementally proportional to:

$$I \propto \cos\theta \cos\phi \tag{2}$$

where θ is the angular displacement of the electrode location from the negative z-axis and ϕ is the angular displacement from the xz-plane. The more point electrodes are placed at the surface of the volume conductor, the more uniform is the lead field.

One significant benefit of a dense EEG array is that since hundreds of electrodes may be applied to the scalp of the patient, the multielectrode lead can be so oriented that the sensitivity distribution is parallel to the source orientation. According to Eq. (1) the parallel orientation of the lead field and the source maximizes the signal amplitude.

2.2 Optimization of weights

The weights for multielectrode EEG lead can be calculated with Eq. (2), if the head is modeled with a concentric spherical model and the electrodes are uniformly distributed over the scalp surface. When considering the deep EEG sources, the properties of the spherical volume conductor model differ substantially from the actual anatomy. Thus it would be beneficial to apply a realistically shaped head model. If the head is modeled with a realistically shaped head model, an optimization method is needed to define the weights of a multielectrode lead to synthesize a uniform sensitivity distribution.

In the optimization method the lead fields of all individual unipolar leads are first calculated. From these lead fields $3m$xn matrix A is constructed. m is the number of the locations, where the orthogonal components of the lead fields are calculated. n is the number of measurement electrodes. w is defined as a nx1 matrix, which includes the weights of n unipolar leads. If A is multiplied with w, we obtain a $3m$x1 matrix c, which includes the orthogonal components of the lead field of a multielectrode lead:

$$Aw = c \tag{3}$$

In the optimization of the weights, the purpose is to optimize the sensitivity distribution of a multielectrode lead so that it is uniform within the brain volume. For this purpose, the matrix c is defined in a way, that it includes the components of a uniform lead field. Based on the calculated A and c, an estimate for the weights w can be obtained by solving the inverse of Eq. (3). In the present study, the inverse was calculated by applying singular value decomposition. The lead fields of unipolar leads were calculated within the volume conductor model at the total of 46247 locations.

2.3 BAEP measurements

BAEPs were obtained from four volunteers (3 male, 1 female, mean age 28) with normal BAEP. As a stimulus we used a rarefaction click (0.1 ms, 80 dB nHL) with a presentation frequency of 9.9 Hz. The stimuli were delivered to the ear through a tubal phone, while a mask noise was presented to the contralateral ear. EEG was measured with a 124-channel EEG cap (SynAmps, NeuroScan, Compumedics). The high-pass filter was set at 0.05 Hz and the low-pass at 2000 Hz. The electrode impedances were kept below 10 kΩ. From each testee 4000 epochs were colleted in two separate sets per ear. Subsequently and prior to the offline analysis the data were filtered with a 100 Hz high-pass filter.

For the SNR analysis we selected a time interval including the BAEP peaks IV-V from the epochs. It has been shown in a dipole localization study, that the sources generating peaks IV-V are directed almost towards the vertex [Schreg and von Cramon, 1985].

2.4 Multielectrode lead analysis

The EEG can be measured with a dense EEG electrode array and the recorded signal of the multielectrode lead can be calculated either online, during the measurement, or offline thereafter, by adjusting the weight of each unipolar lead. The signal x of the multielectrode lead is calculated with Eq. (4), where x_i is the signal measured with unipolar lead i and w_i is the weight assigned for unipolar lead i. n is the number of unipolar leads in original measurement.

$$x = \sum_{i=1}^{n} w_i x_i \qquad (4)$$

We calculated the weighted signal x offline after the recording. All the calculations were done in Matlab. EEGLAB was applied to read the Neuroscan files into the Matlab [Delorme and Makeig, 2004].

The BAEPs were measured with 124 electrodes referred to an electrode located on the forehead. The digitized electrode coordinates were fitted on the surface of a three-layer spherical head model. The positive x-axis, the positive y-axis and the positive z-axis go through right mastoid, nasion and vertex, respectively. If the impedance of some electrode was notably high or the signal looked extremely noisy, that electrode was excluded from the multielectrode lead. Also artifact contaminated epochs were excluded from the analysis. We calculated the amplitude SNRs with equations from [Raz *et al.*, 1988], which give an SNR estimate for one epoch.

In [Väisänen and Malmivuo, 2007] we calculated the weights with Eq. (2) and optimized the orientations of the multielectrode lead fields by changing the lead field orientation until the optimal orientation giving highest SNR was found. In the present study we assumed, that the lead field orientation at the center of the volume conductor model calculated in [Väisänen and Malmivuo, 2007] is optimal, and thus it was defined as the lead field orientation in matrix c in Eq. (3).

2.5 Head model

As a volume conductor model we applied the three-layer spherical head model with radii of 92 mm, 85 mm and 80 mm, respectively, as in the Rush and Driscoll head model. As in [Oostendorp *et al.*, 2000] the resistivity ratio between the tissues was 1:15:1. We applied the lead field and reciprocity theorems to calculate the sensitivity distributions within the volume conductor. In the calculations we used the analytical equations derived in [Rush and Driscoll, 1969]. The realistic electrode coordinates of a 124-channel EEG cap were fitted on the surface of the model. After the bad electrodes were removed, the total number of electrodes in multielectrode leads varied between 110 and 118, average being 114.

3. Results

Examples of optimized sensitivity distributions in multielectrode leads are illustrated in Fig. 1 and Fig. 2. for testee MP, when the stimulus is delivered to the right ear. In Fig. 1 the weights are calculated with Eq. (2) [Väisänen and Malmivuo, 2007] and in Fig. 2 the weights are calculated with Eq. (3).
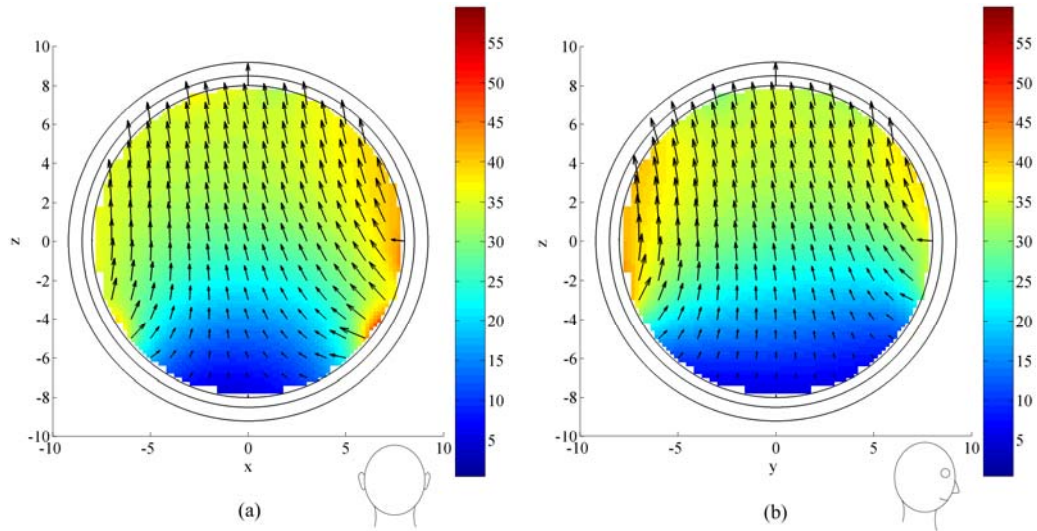
Figure 1. Lead field of the multielectrode lead illustrated within the brain area in (a) xz-plane and (b) yz-plane. Colorscale gives the magnitude of sensitivity distribution $\overline{J}_L$ (A/m²). The head is modelled with a three-layer spherical model. 115 electrodes applied in the measurements are fitted on the surface of the model. The weights are calculated with Eq. (2).
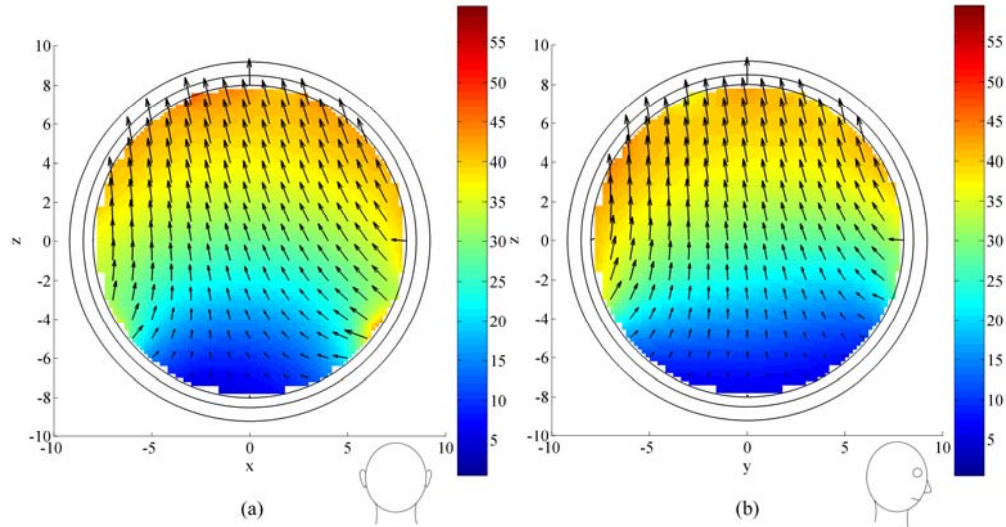

Figure 2. Lead field of the multielectrode lead illustrated within the brain area in (a) xz-plane and (b) yz-plane. Colorscale gives the magnitude of sensitivity distribution $\overline{J}_L$ (A/m²). The head is modelled with a three-layer spherical model. 115 electrodes applied in the measurements are fitted on the surface of the model. The weights are calculated with Eq. (3).

Traditional BAEP electrode montage consists of electrodes placed at ipsilateral mastoid (Ai), contralateral mastoid (Ac) and vertex (Cz). The calculated signal-to-noise ratios of traditional BAEP leads and of multielectrode leads are listed in Table 1. The ML_COS refers to a multielectrode lead, where the weights are calculated with Eq. (2) and ML_OPT refers to a multielectrode lead, where the weights are optimized with Eq. (3). We compared the SNR obtained with multielectrode lead to the SNR obtained with the best traditional BAEP lead. The values marked with (*) in Table 1 were excluded from the averages. In the case the weights were calculated with Eq. (2), the average improvement was 1.59 and in the case the weights were calculated with Eq. (3), the average improvement was 1.65.

Table 1. Calculated SNRs and average improvements obtained with multielectrode EEG leads. The subscript L stands for the left ear stimulus and subscript R for the right ear stimulus.
(*) The contact impedance in electrode Ai or Ac is not optimal.

Testee	Electrodes in ML	SNR Cz - Ai	SNR Cz - Ac	SNR ML_COS	SNR ML_OPT
EH_R	110	0.088	0.13	0.182	0.186
EH_L	111	0.11	0.025[*]	0.162	0.169
MP_R	115	0.067[*]	0.070[*]	0.143	0.144
MP_L	116	0.044[*]	0.039[*]	0.134	0.140
MR_R	116	0.11	0.11	0.190	0.199
MR_L	113	0.090	0.084	0.151	0.158
TV_R	118	0.10	0.11	0.194	0.202
TV_L	116	0.16	0.13	0.238	0.246

Average SNR improvement → **1.59** → **1.65**

4. Discussion

This paper shows that the detection of the signals generated deep in the brain can be improved with the multielectrode lead technique. The results show that the amplitude SNR of BAEPs obtained with multielectrode lead consisting of 110-118 electrodes is on average 1.65 times that of traditional BAEP leads. The improvement in SNR obtained with multielectrode EEG lead can be utilized by reducing the number of epochs included in the average. When the amplitude SNR is improved by a factor of 1.65, instead of averaging 2000 epochs with traditional BAEP leads, it is sufficient to average 730 epochs with multielectrode lead to obtain similar SNR.

In a theoretical case, when the head is modeled with a layered concentric spherical model and the electrodes uniformly cover the scalp surface, the weights for optimal multielectrode lead can be calculated with Eq. (2). However, the need for realistically shaped head model is clear, because of the inaccuracy of the spherical model at deep structures of the head. Thus realistically shaped head models including e.g. the ventricles should be used when defining the optimal weights for multielectrode leads. In the present study we evaluated a numerical optimization method, which can be used in the future studies to define the weights for multielectrode leads, when the head is modeled with a realistically shaped model.

In the present study the head was modeled with a three-layer spherical model. In the experimental measurements the SNR of multielectrode leads was slightly improved, when the weights were defined with an optimization method instead with a method based on the cosines of electrode angles. The main reason for the improvement was the application of realistical electrode system. The realistical electrode system covers only slightly more than half of the head surface, and thus the calculation of weights discretely based on the electrode coordinates with Eq. (2) does not result in optimal lead field. In the future studies in addition for applying a realistically shaped head model, new electrode montages should be studied, which would cover also the lower part of the head more comprehensively than traditional EEG montages. Based on the results the optimization method can be applied in the future studies to define the weights for multielectrode EEG leads.

The new multielectrode leads will bring many benefits to EEG studies. The first applications include evoked potential studies. If the same number of epochs is recorder with multielectrode lead than with traditional leads, the signal quality will be substantially improved. On the other hand, fewer epochs are sufficient for an average without reducing the signal quality. Thus for example the effects of changes in response signal will become smaller. New multielectrode leads more sensitive to measure deep EEG sources might also reveal new features of brain activity in spontaneous recordings.

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