Proceedings of the 29th Annual International
Conference of the IEEE EMBS
Cité Internationale, Lyon, France
August 23-26, 2007.

Improved Detection of Deep Sources with Weighted Multielectrode EEG Leads

Outi Väisänen, and Jaakko Malmivuo, *Fellow, IEEE*

Abstract—**The purpose of the present study is to conduct preliminary experimental measurements to validate the improvement in the detection of deep EEG sources achieved with new multielectrode EEG leads. As a measurement we had brainstem auditory evoked potentials (BAEPs), which include deep generators in the brainstem and midbrain. The BAEPs were measured with a 124-channel EEG cap. We have previously developed a multielectrode lead technique, which has its basis in optimization of the sensitivity distribution of a multielectrode lead for detecting signals generated by deep sources. The purpose of the present study is to validate with experimental measurements the results previously obtained with theoretical approach and simulations. The results show that the amplitude SNR of BAEPs obtained with multielectrode lead is on average 1.6 times that of traditional bipolar BAEP lead. Though improvement obtained in experimental measurements is smaller than was theoretically approximated it encourages for further development of the multielectrode leads.**

I. INTRODUCTION

THE purpose of the present paper is to evaluate with experimental measurements the performance of new multielectrode lead method, which has been developed to improve the detection of deep EEG sources [1]. Modern EEG equipment, comprising up to 256 channels, enable the synthesis of new kind of measurement leads. In a multielectrode lead several electrodes are applied to form a new single measurement lead. We have introduced the method in [1], where we also evaluated it's performance both with theoretical calculations and with simulations. The basis of the method lies in optimization of the sensitivity distribution of the multielectrode lead. We have shown in [1] that 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.

Most of the methods previously applied to improve the SNR of EPs compared to traditional averaging comprise different signal-processing techniques that are applied to signals measured with traditional EEG leads. Such approaches include e.g. median and weighted averaging and trimmed estimators [2-4]. Also wavelet based denoising methods have been applied [5].

In the present study we evaluated with experimental measurements the performance of the multielectrode lead technique. 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.

II. METHODS

A. 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. The 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 [6].

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

In (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 [7]. 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

This work has been supported by the Finnish Cultural Foundation, and the Ragnar Granit Foundation.

O. Väisänen and J. Malmivuo are with Ragnar Granit Institute, Tampere University of Technology, FIN-33101 Tampere, Finland. (e-mail: outi.vaisanen@tut.fi).

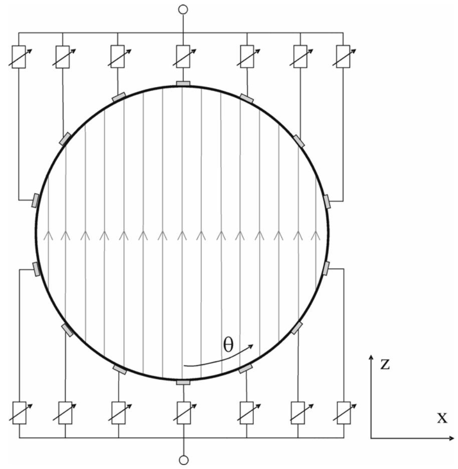


Fig. 1. Application of the reciprocal current to obtain a homogeneous lead field in a spherical volume conductor. The origin of the coordinate system is at the center of the model. The lead field illustrated with arrows is oriented in the positive z-direction. Variable resistors reflect that the current applied to different leads is adjusted based on the angle θ.

realized for deep sources is the homogeneous sensitivity distribution [1].

We have previously showed how the homogenous sensitivity distribution can be obtained in a spherical three-layer head model [1]. The spherical model can be thought to consist of thin cylindrical disks with circular cross-section. Within each of these disks the injected reciprocal current at every electrode is incrementally proportional to $\cos\theta$, where θ is the angular displacement from the negative z-axis as illustrated in Fig.1. The lead field is oriented along the positive z-axis. The shape of the volume conductor along the y-axis is considered by multiplying the $\cos\theta$ with $\cos\phi$, where ϕ is the angular displacement from the xz-plane. Thus, to obtain a homogeneous lead field throughout the volume conductor, the reciprocal current injected at each electrode needs to be incrementally proportional to:

$$I \propto \cos\theta \cos\phi \qquad (2)$$

The more point electrodes are placed at the surface of the volume conductor, the more homogeneous is the lead field. After the currents/weights are calculated with (2) the sum of positive and negative weights is normalized to correspond to a unit reciprocal current.

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 (1) the parallel orientation of the lead field and the source maximizes the signal amplitude.

B. BAEP measurements

BAEPs are generated in the auditory pathway passing through the cochlear nerve, pons and midbrain [8]. BAEP peaks are traditionally named with roman numbers from I to VII. 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 two different time intervals from the epochs. The first interval included the peaks I-V and the second interval peaks IV-V. In a multielectrode lead the purpose is to orientate the lead field in the direction of the dipolar source. In a dipole localization study the dipole sources generating the peaks I-V were localized. Judging by the results obtained with only 10 measurement electrodes, the activity generating peaks I-III proceeds more horizontally and the sources generating peaks IV-V are directed almost towards the vertex [9].

C. Multielectrode lead analysis

With modern digital signal processing there is no need to physically construct a resistor network as illustrated in Fig. 1. 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 bipolar lead. The signal x of the multielectrode lead is calculated with (3), where x_i is the signal measured with bipolar lead i and w_i is the weight assigned for bipolar lead i. L is the number of leads in original measurement.

$$x = \sum_{i=1}^{L} w_i x_i \qquad (3)$$

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 [10].

The BAEPs were measured with 124 electrodes referred to an electrode located on the forehead. The electrode coordinates for each testee were digitized with Polhemus FASTRAK. The positive x-axis goes through right mastoid, positive y-axis through nasion and positive z-axis through vertex. Before the calculation of the weights w_i, the electrode positions were first recentered, so that the locations best fit a spherical surface [10]. The weights were calculated with (2) from the recentered electrode coordinates. The orientation of the lead field was studied by rotating the electrode coordinatates before calculating the weights. 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.* [11], which give an SNR estimate for one epoch.

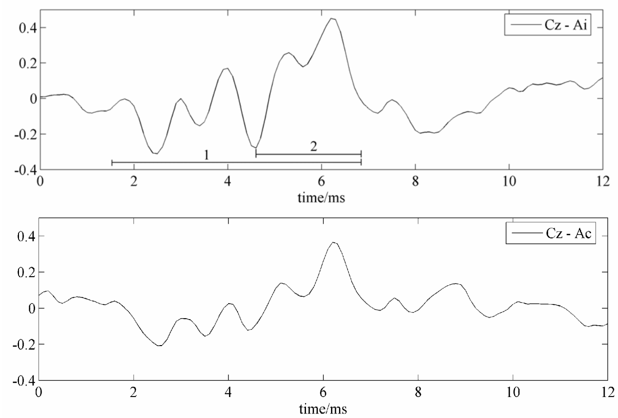


Fig. 2. Original BAEP waveforms from MP, when the stimulus was delivered to right ear. 3970 epochs are averaged. The time interval 1 includes peaks IV-V and time interval 2 peaks I-V.

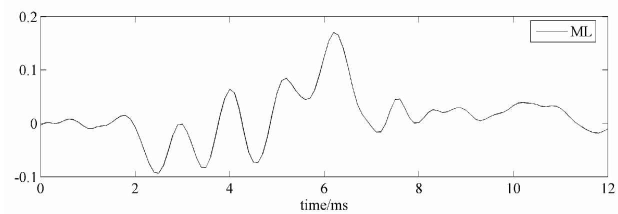


Fig. 3. BAEP waveform of multielectrode lead (ML) from MP, when the stimulus was delivered to right ear. 3970 epochs are averaged.

III. Results

Traditional BAEP electrode montage consists of electrodes placed at ipsilateral mastoid (Ai), contralateral mastoid (Ac) and vertex (Cz). An example of BAEP measured with traditional leads is presented in Fig. 2, where the two time intervals studied are also marked. The same signal measured with an optimal multielectrode lead is illustrated in Fig. 3.

During time interval 1, all of the peaks I-V are best seen in channel Cz-Ai. In Table I the calculated SNRs for time interval 1 are given for each testee. The SNRs are calculated in the lead Cz-Ai and in the multielectrode lead (ML).

In Table II the calculated SNRs for time interval 2 are given for each testee. The SNRs are calculated in traditional BAEP leads Cz-Ai and Cz-Ac. Sometimes the peaks III- IV are best defined if the Cz is referred to neck or occipital reference [12]. Thus if within the 124 measurement electrodes there existed another reference location to Cz, which gave higher SNR than mastoid references, it is also given in Table II. For both time intervals 1 and 2, the orientations of the multielectrode lead fields were optimized by changing the lead field orientation until the optimal orientation giving highest SNR was found.

The average improvement in SNR obtained with multielectrode leads was calculated in three different cases. The values marked with (*) in Tables I and II were excluded from the averages. In the first case the SNR improvement in time interval 1 was studied. In the second and third case, the SNR improvement in time interval 2 was studied:

In the first case the SNR obtained with ML was compared to SNR obtained with traditional lead Cz-Ai. On average the SNR of ML was 1.6 times that of lead Cz-Ai.

In the second case the SNR of ML was compared to the SNR of best possible traditional BAEP lead (Cz-Ai or Cz-Ac). On average the SNR of ML was 1.6 times that of best traditional BAEP lead.

In the third case the SNR of ML was compared to best possible of bipolar lead (i.e. Cz referred either to Ac, Ai or neck reference). In this case on average the SNR of ML was 1.5 times that of best bipolar lead.

An example of the sensitivity distribution of a multielectrode lead within the brain area is given in Fig. 4. The multielectrode lead is the same than in Fig. 3.

Table I. SNRs in time interval 1. 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 ML
EH_R	110	0.070	0.13
EH_L	111	0.081	0.12
MP_R	115	0.047(*)	0.12
MP_L	116	0.044(*)	0.11
MR_R	116	0.082	0.13
MR_L	113	0.065	0.10
TV_R	118	0.097	0.17
TV_L	116	0.12	0.17

Table II. SNRs in time interval 2. 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.

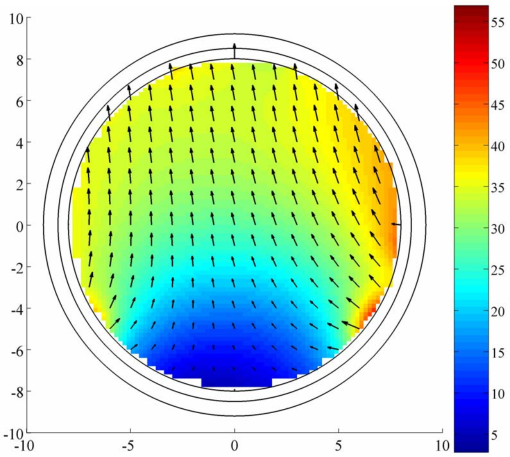
Testee	Electrodes in ML	SNR Cz - Ai	SNR Cz - Ac	SNR Cz-neck	SNR ML
EH_R	110	0.088	0.13	-	0.18
EH_L	111	0.11	0.025(*)	-	0.16
MP_R	115	0.067(*)	0.070(*)	0.088	0.14
MP_L	116	0.044(*)	0.039(*)	0.082	0.13
MR_R	116	0.11	0.11	0.14	0.19
MR_L	113	0.090	0.084	0.12	0.15
TV_R	118	0.10	0.11	0.14	0.19
TV_L	116	0.16	0.13	-	0.24

IV. Discussion

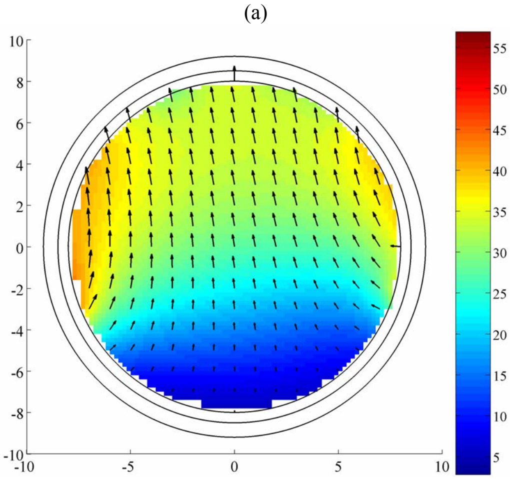
This paper shows that the detection of the signals generated deep in the brain can be improved with the new multielectrode lead technique. The results show that the amplitude SNR of BAEPs obtained with multielectrode lead consisting of 110-118 electrodes is on average 1.6 times that of traditional BAEP leads.

If on the other hand the multielectrode lead is compared with the best possible bipolar lead measuring activity generated deep in the brain, the SNR of multielectrode lead is on average 1.5 times that of the bipolar lead.

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 1.6, instead of averaging 2000 epochs with traditional BAEP leads, it is sufficient to average 780 epochs with multielectrode lead to obtain similar SNR.



(a)



(b)

Fig. 4. 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/m^2). The multielectrode lead is the same as in Fig. 3.

During the time interval 1 the orientation of the multielectrode lead was not optimal, because the source orientation changes as the signal proceeds from the cochlea through the pons to the midbrain. Anyhow the improvement in SNR obtained with multielectrode lead was notable. The plausible reason for this is the relatively large amplitude of peaks IV-V and the orientation of the multielectrode lead field, which is optimal for these sources.

In [1] we studied with simulations the improvement obtained with a multielectrode lead including 102 electrodes uniformly distributed over the surface of a three-layer spherical head model. In the simulations the noise was modeled as Gaussian white noise, which was uncorrelated between different bipolar leads. The SNR of that multielectrode lead was 3.0 times that of optimal bipolar lead. The electrode angle in optimal bipolar lead is 180°. Thus the improvement obtained here with the experimental measurements is approximately half of that obtained with simulations.

In the present paper the weights for the multielectrode lead were calculated based on the assumption that the head was modelled with a three-layer spherical head model and the electrodes were evenly distributed over the whole scalp surface. The need for realistically shaped head model is apparent. Also the electrodes covered only slightly more than half of the head surface. In the future studies new electrode montages should be studied, which would cover also the lower part of the head more comprehensively than traditional EEG montages. Also an optimization method needs to be applied to define the weights for the different bipolar leads which form the multielectrode lead. The improvements obtained in the present study encourage for these future developments of the method.

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.

REFERENCES

[1] O. Väisänen and J. Malmivuo, "Improved Detection of Deep EEG Sources with Multielectrode EEG Leads," submitted.

[2] O. Ozdamar and T. Kalayci, "Median averaging of auditory brain stem responses," *Ear Hear*, vol. 20, pp. 253-64, 1999.

[3] C. E. Davila and M. S. Mobin, "Weighted averaging of evoked potentials," *IEEE Trans Biomed Eng*, vol. 39, pp. 338-45, 1992.

[4] Z. Leonowicz, J. Karvanen, and S. L. Shishkin, "Trimmed estimators for robust averaging of event-related potentials," *J Neurosci Methods*, vol. 142, pp. 17-26, 2005.

[5] E. Causevic, R. E. Morley, M. V. Wickerhauser, and A. E. Jacquin, "Fast wavelet estimation of weak biosignals," *IEEE Trans Biomed Eng*, vol. 52, pp. 1021-32, 2005.

[6] J. Malmivuo and R. Plonsey, *Bioelectromagnetism: principles and applications of bioelectric and biomagnetic fields*. New York: Oxford University Press, 1995.

[7] J. A. Malmivuo and V. E. Suihko, "Effect of skull resistivity on the spatial resolutions of EEG and MEG," *IEEE Trans Biomed Eng*, vol. 51, pp. 1276-80, 2004.

[8] K. Chiappa and R. Hill, "Brain Stem Auditory Evoked Potentials: Interpretation," in *Evoked Potentials in Clinical Medicine*, K. Chiappa, Ed., 3rd ed. Philadelphia: Lippincott-Raven, 1997, pp. 709.

[9] M. Scherg and D. von Cramon, "A new interpretation of the generators of BAEP waves I-V: results of a spatio-temporal dipole model," *Electroencephalogr Clin Neurophysiol*, vol. 62, pp. 290-9, 1985.

[10] A. Delorme and S. Makeig, "EEGLAB: an open source toolbox for analysis of single-trial EEG dynamics including independent component analysis," *J Neurosci Methods*, vol. 134, pp. 9-21, 2004.

[11] J. Raz, B. Turetsky, and G. Fein, "Confidence intervals for the signal-to-noise ratio when a signal embedded in noise is observed over repeated trials," *IEEE Trans Biomed Eng*, vol. 35, pp. 646-9, 1988.

[12] M. R. Nuwer, M. Aminoff, D. Goodin, S. Matsuoka, F. Mauguiere, A. Starr, and J. F. Vibert, "IFCN recommended standards for brain-stem auditory evoked potentials. Report of an IFCN committee. International Federation of Clinical Neurophysiology," *Electroencephalogr Clin Neurophysiol*, vol. 91, pp. 12-7, 1994.