Estimation of the neuronal current via Electro-Magneto-Encephalography using real data

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Abstract. The medical significance of Electroencephalography (EEG), and Magnetoencephalography is well established, see for examples [1–3]. EEG and MEG are considered two of the most important imaging techniques for real time brain imaging. In order to generate images of the brain activation using either EEG or MEG, it is necessary to analyse certain mathematical inverse problems. The definitive answer to the inverse source problem for the case of EEG and MEG was finally obtained by [4]. Here, we present reconstructions of the current using real data via the formulation proposed in [4]. The data was provided by the Medical Research Council (MRC) Cambridge, UK and involves a visual stimuli. We show comparisons of the reconstructed irrotational component of the neuronal current using EEG measurements and the radial component of the neuronal current using MEG measurements. Based on the results, we argue that EEG imaging technology has the potential to become the dominant real time, low cost brain imaging tool.

1 Introduction

The inverse problems of the important imaging techniques of Electroencephalography (EEG), and Magnetoencephalography (MEG) involve the estimation of the neuronal current from the knowledge of the electric potential on the scalp, and of the magnetic field outside the head, respectively. However, as it was known to Helmholtz since 1853, both these techniques suffer from non-uniqueness. Following the efforts of several researchers, see for example [5–20], the complete answer to the non-uniqueness question for both EEG and MEG for arbitrary geometry was presented in [4]. Furthermore, effective formulas for the “visible” component of the current associated with EEG and MEG were also presented in [4]. Let the bounded domain \( \Omega_c \) represent the cerebrum, which has conductivity \( \sigma_c \). A shell \( \Omega_f \) with conductivity \( \sigma_f \) representing the cerebrospinal fluid (CSF) which surrounds the domain \( \Omega_c \). The CSF is surrounded by the skull characterized by the domain \( \Omega_b \) with conductivity \( \sigma_b \). Finally, the skull is surrounded by the scalp which is modelled as a shell \( \Omega_s \) with conductivity \( \sigma_s \). The domain exterior to the head is denoted by \( \Omega_e \) which is not conductive. The permeability of all domains is equal to the permeability \( \mu_0 \) of empty space. Let
J^p(\tau), \tau \in \Omega_c, denote the current which is supported within the cerebrum \Omega_c. An arbitrary vectorial function with support in \Omega_c, can be expressed as

\[ J^p(\tau) = \nabla\Psi(\tau) + \nabla \times A(\tau), \quad \tau \in \Omega_c, \]  

(1.1)

where A(\tau) satisfies the constraint \nabla \cdot A(\tau) = 0. This constraint implies that J^p(\tau) involves three arbitrary scalar functions, namely the scalar function \Psi(\tau) and the two independent scalar functions characterizing A(\tau).

It is shown in [4] that the electric potential on the scalp is given by the expression

\[ u_s(r) = \frac{1}{4\pi} \int_{\Omega_s} \left( \nabla^2 \Psi(\tau) \right) v_s(r, \tau) dV(\tau), \quad r \in \partial \Omega_s, \]  

(1.2)

where the function v_s(r, \tau) is independent of the current and is governed by a well defined boundary value problem [4]. Moreover, the measured magnetic field in \Omega_e satisfies the expression

\[ \frac{4\pi}{\mu} B(r) \cdot r = -\int_{\Omega_e} \nabla^2 (\tau \cdot A(\tau)) \frac{dV(\tau)}{|r - \tau|} \]

\[ + \frac{1}{4\pi} \int_{\Omega_c} \left( \nabla^2 \Psi(\tau) \right) r \cdot H(r, \tau) dV(\tau), \quad r \in \Omega_e, \]  

(1.3)

where H(r, \tau) is a vectorial function which can be computed in terms of v_s(r, \tau).

A numerical implementation of the above formulas for the physiologically unrealistic case of spherical geometry was presented in [21]. Numerical examples were presented using the exact sensor (gradiometers) positions of the Elekta Neuromag system; the orientation of the gradiometers was assumed to be in the direction of the unit normal vector. For the case of the spherical geometry, v_s(r, \tau) can be computed explicitly and \nabla \cdot H(r, \tau) = 0. In a recent important development, it was shown in [22] that v_s(r, \tau) can be expressed in terms of an auxiliary functions w_s(r, \tau), and furthermore a numerical code was presented for computing w_s(r, \tau) in arbitrary geometry. In what follows we present the reconstructions of the current employing this code, to corresponding real EEG and MEG data associated with a particular visual experiment.

2 Visual Experiment

Two visual checker board patterns, one on the left and one on the right of a central fixation cross, were simultaneously flashed for 32 ms approximately every 2.5 s. These stimuli were intermixed with other trials that contained checker boards only on the left or right, tone beeps on the left or right, as well as tone beeps in both ears. Participants had to indicate by button press whether a stimulus appeared on the left, right or on both sides. EEG and MEG data were recorded using an Elekta Neuromag Vectorview system. The set-up consisted of 306 MEG sensors and 70 EEG electrodes. Data were off-line low-pass-filtered at 40 Hz, and subjected to averaging. The analysis was applied to data from one participant.
3 Results and Discussions

The numerical solution of the boundary value problem associated with the function \(v_s(r, \tau)\) and the numerical construction of the vectorial function \(H(r, \tau)\) is discussed in [22].

(a) A sagittal view of the MEG reconstructions for the visual left-right stimulus.

(b) A sagittal view of the EEG reconstructions for the visual left-right stimulus.

Fig. 1. A sagittal view reconstruction for visual left-right stimulus.
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References


