Biomedical Signal Processing and Control 8 (2013) 830–837

Contents lists available at ScienceDirect

Biomedical Signal Processing and Control

journal homepage: www.elsevier.com/locate/bspc

Analysis of parametric estimation of head tissue conductivities using Electrical Impedance Tomography

Mariano Fernández-Corazza[a,b,*], Leandro Beltrachini[a,b], Nicolás von Ellenrieder[a,b], Carlos H. Muravchik[a,c]

[a] Laboratorio de Electrónica Industrial, Control e Instrumentación (LEICI), Departamento de Electrotecnia, Facultad de Ingeniería, Universidad Nacional de La Plata, Calle 48 y 116, CC91 (1900), La Plata, Buenos Aires, Argentina
[b] Consejo Nacional de Investigaciones Científicas y Técnicas (CONICET), Argentina
[c] Comisión de Investigaciones Científicas de la Provincia de Buenos Aires (CICpBA), Argentina

ARTICLE INFO

Article history:
Received 6 February 2013
Received in revised form 24 June 2013
Accepted 5 August 2013
Available online 3 September 2013

Keywords:
Electrical Impedance Tomography
Cramér-Rao Bound
Maximum Likelihood Estimator
Electrical conductivity estimation
Skull anisotropy
Finite Element Method

ABSTRACT

We study the theoretical performance of using Electrical Impedance Tomography (EIT) to measure the conductivity of the main tissues of the head. The governing equations are solved using the Finite Element Method for realistically shaped head models with isotropic and anisotropic electrical conductivities. We focus on the Electroencephalography (EEG) signal frequency range since EEG source localization is the assumed application. We obtain the Cramér-Rao Lower Bound (CRLB) to find the minimum conductivity estimation error expected with EIT measurements. The more convenient electrode pairs selected for current injection from a typical EEG array are determined from the CRLB. Moreover, using simulated data, the Maximum Likelihood Estimator of the conductivity parameters is shown to be close to the CRLB for a relatively low number of measurements. The results support the idea of using EIT as a low-cost and practical tool for individually measure the conductivity of the head tissues, and to use them when solving the EEG source localization. Even when the conductivity of the soft tissues of the head is available from Diffusion Tensor Imaging, EIT can complement the electrical model with the estimation of the skull and scalp conductivities.

1. Introduction

Electric models of the human head, as used in Electroencephalography (EEG) source localization, require an electrical conductivity map of the head. Usually, this map is built as a layered model, with each layer representing a different tissue. The shape of the layers is either assumed spherical, ellipsoidal, or obtained by segmenting Magnetic Resonance (MR) images. Then, the electrical conductivity values for each tissue are usually selected from existing studies [1–5]. However, there is a large variation in the values found in the literature, depending on the measurement method, and probably due to inter-subject variability [6]. The use of incorrect values in the head model could lead to erroneous solutions of the neural source estimation [7–9]. A possible solution to this problem is the use of Electrical Impedance Tomography (EIT) measurements from the subject under study to estimate the conductivity of the main tissues of the head. As EIT is portable and relatively cheap, this is an advantage over other methods such as MR (particularly, Diffusion Tensor Imaging (DTI)) or Computed Tomography (CT) that could be used with the same purpose. Also, EIT can estimate the skull conductivity, which cannot be obtained from DTI. Most of the skull conductivity values found in the above mentioned literature are based on *in vitro* or *in vivo* measurements with a four electrode system. In EIT, the same electrodes and acquisition system of EEG could be used in combination with an electric current source to estimate *in vivo* the tissue conductivities.

Electrical Impedance Tomography (EIT) on the head consists of injecting electrical currents on known points on the scalp and measuring the resulting electric potential distribution on the scalp to infer the electrical impedance map of the whole head volume. An inconvenience with EIT impedance mapping is that even when using an array with a large number of electrodes, the spatial resolution of the resulting map is quite low [10]. To overcome this, it is possible to combine the high spatial resolution of the MR segmentation in tissues, with the EIT technique to estimate the electrical

* Corresponding author at: Laboratorio de Electrónica Industrial, Control e Instrumentación (LEICI), Departamento de Electrotecnia, Facultad de Ingeniería, Universidad Nacional de La Plata, Calle 48 y 116, CC91 (1900), La Plata, Buenos Aires, Argentina. Tel.: +54 2214259306; fax: +54 2214259306.
E-mail addresses: marianof.corazza@ing.unlp.edu.ar, marianofco@gmail.com (M. Fernández-Corazza), lbeltra@ing.unlp.edu.ar (L. Beltrachini), nellen@ieee.org (N. von Ellenrieder), carlosm@ing.unlp.edu.ar (C.H. Muravchik).

1746-8094/$ – see front matter © 2013 Elsevier Ltd. All rights reserved.
http://dx.doi.org/10.1016/j.bspc.2013.08.003

conductivity of each layer [11,12]. In this way, parametric estimation tools are used to solve the EIT inverse problem (EIT IP), i.e. the construction of the conductivity map.

The skull is highly resistive compared to the surrounding tissues, acting as an electrical shield between the scalp and the brain. It is mostly composed by a sandwich of two layers of compact bone with a layer of spongy bone in the middle [4]. Several models have been proposed for the skull adopting isotropic conductivities to each type of bone [13], a single homogeneous isotropic conductivity, localized but isotropic conductivities [14], or homogeneous but anisotropic models [9,13], where the tangential conductivity is higher than the radial (or transversal) one. The relevance of considering the skull anisotropy has also been analyzed [9]. Several studies measured the skull conductivity and its anisotropy ratio [1,2,15,16]. Existing reports show skull conductivity values ranging from 0.04 S/m to 0.004 S/m [1,2,4,16] (and even some values outside this range have been discussed). The radial:tangential anisotropy ratio was initially supposed to be 1:10 [1], but recent studies suggest that it is lower, ranging from 1:1.6 to 1:6 [3,9] considering the 1:10 ratio as an upper limit. Our hypothesis is that the ratio is still uncertain and a wide range of anisotropic conductivity ratios (from 1:1 to 1:10) is evaluated.

The scalp is the first compartment that the current passes through, playing a major role in EIT [4]. It can also be modeled as inhomogeneous because it has zones with muscles, fat and different skin thickness. Homogeneous but anisotropic models with transversal to tangential ratio of 1:1.5 have been also used [8]. As the conductivity of this tissue is higher than the skull's, its relative inhomogeneities are usually neglected and a single isotropic conductivity is assigned to this layer. This value is also uncertain since only few studies have been performed to measure it [4]. So, EIT can also be applied to obtain a scalp conductivity estimate.

The innermost cavity is the most complex as it includes cerebrospinal fluid (CSF), gray matter, and white matter. Assigning to it a single conductivity value is nowadays an oversimplification of the problem and recent studies also highlight the relevance of considering anisotropy within the white matter [8,9,17]. The diffusion of water over the tissues (MR-DTI) can be used to build an anisotropic conductivity map of the CSF, white and gray matter, by means of a linear transform of each tensor eigenvalue [18]. Averaged conductivity distributions can be used if the patient specific resonance is not available [19].

We assess the performance of using EIT to estimate the conductivity values of the scalp, skull, and brain. For the first two tissues we analyze homogeneous isotropic and anisotropic conductivities, whereas for the brain, we assign a unique isotopic value or a realistic inhomogeneous and anisotropic map based on a DTI atlas. We use this to analyze the influence of the *a priori* knowledge of the tensorial map of the brain conductivity (that could be obtained from DTI) in the EIT parametric estimation. The Cramér-Rao Lower Bound (CRLB) is computed for the unknown parameters, allowing us to quantify the performance of the method independently of the specific algorithm used to estimate the conductivity values, and to detect convenient electrode pairs for the current injection. This builds upon our preliminary study [20], where we analyzed convenient pairs using a simple spherical head model with a particular estimation algorithm. In contrast to other studies [11,12], in this work we use detailed realistic head models including the estimation of the anisotropic components of the tissue conductivities. To the best of our knowledge, our parametric EIT estimation analysis introduces accounting for anisotropic components of the scalp and skull conductivity tensors. Finally, we compute the Maximum Likelihood Estimator (MLE) to solve the EIT IP using simulated signals with additive noise and we show that its performance is close to the bound even for a limited number of measurements.

2. Methods

2.1. Forward problem solution

The EIT forward problem (FP) consists in calculating the electric potential distribution on the scalp as a result of current injection, assuming that the electrical conductivities are known. In the EEG frequency range, this is a quasistatic problem with Neumann boundary conditions [21]. The governing equations are

$$\begin{cases} \vec{\nabla}\cdot\left(\boldsymbol{\sigma}\left(\vec{x}\right)\vec{\nabla}\boldsymbol{\Phi}\left(\vec{x}\right)\right)=0, & \text{in}\,\Omega \\ \boldsymbol{\sigma}\left(\vec{x}\right)\left(\vec{\nabla}\boldsymbol{\Phi}\left(\vec{x}\right)\right)\cdot\hat{n}=j(\vec{x}), & \text{in}\,\delta\Omega \end{cases} \tag{1}$$

where $\vec{x}=(x,y,z)$ represents a generic point in the head, $\boldsymbol{\Phi}$ is the electric potential, $\boldsymbol{\sigma}$ represents the conductivity tensor, Ω is the head volume and $\delta\Omega$ is its outer surface, $\hat{n}$ is the normal unit vector, and j is the normal component of the current density on the boundary.

The FP can be solved analytically for spherical geometries and isotropic conductivities [20,22]. When using arbitrary shapes, numerical methods such as the Boundary Element Method (BEM) or the Finite Element Method (FEM) are required to solve the problem. We use the FEM because it admits the anisotropy of the tissues. Existing studies have validated the use of FEM for EIT purposes [23].

Using linear basis functions in the FEM with tetrahedral elements (and n nodes), the problem is converted into a linear system of equations with the form $KU=F$ (see Appendix A) where K is the $n\times n$ *stiffness matrix* that includes information of the head geometry and conductivity, U is the unknown $n\times 1$ vector of the potential at the n nodes of the volume tessellation, and F is the $n\times 1$ independent vector that includes the information of the electric current.

For simplicity, the electrodes are assumed to be a point with no surface as the electrode area is much smaller than the total area of the external head surface [24]. Complete electrode models could also be considered as in [8].

2.2. Signal model

Based on the FP we may write a sample Y of the signal at the electrodes as

$$\mathbf{Y}=MU+W=MK^{-1}F+W, \tag{2}$$

where M is an $m\times n$ sparse selection matrix that selects the elements of U corresponding to the nodes located at the position of the m measurement electrodes, and W is a noise term. The noise in the measurements should contemplate two sources; the noise of the amplifiers and the skin-electrode contact impedance, and the electrical activity of the brain, which is an undesired noise term in this EIT application. The latter may have larger amplitude and could lead to a large variance of the estimated parameters [20]. However, as it is spatially and temporally correlated, an appropriate selection of the current waveform and data processing mitigates its effect if a sufficient number of temporal samples are available [29].

2.3. Head models

The head models adopted for this study have a realistic shape and are composed by three layers representing brain, skull, and scalp. In order to obtain general results, average models are chosen for the head shape and the surfaces that delimit the layers. They are obtained from the ICBM152 atlas [27], which is an average of 152 healthy subjects. We analyze three models:

Isotropic: The three layers are considered homogeneous and isotropic with their three parameters of interest; the electrical conductivities of each layer (σ_{sc}, σ_{sk}, σ_{br}).