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Sensitivity of rheoencephalographic measurements to spatial brain electrical conductivity

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Abstract— **Rheoencephalography (REG) is impedance plethysmography applied to the head, and provides an indirect measurement of the pulsatility of the cerebral blood volume. To extend REG as a clinical and research tool, it is necessary to evaluate the sensitivity of REG measurement to local brain conductivity changes. By means of the analytical solution of a four-sphere geometrical model of the head, maps of impedance sensitivity were assessed for different electrode arrangements. Results showed a selective distribution of sensitivities, with a preference for cortical areas under electrodes. This suggests a potential for application of REG to regional evaluation of cortical cerebral perfusion.**

I. Introduction

Rheoencephalography (REG) is a technique that arose as an extrapolation of impedance plethysmography applied to the head; this technique was first reported by Polzer and Schuhfried in 1950 [1]. It provides an indirect measurement of the pulsatility of the cerebral blood volume (CBV) associated with the cardiac cycle, which is supposed to be related to the cerebral blood flow (CBF). The principles of the technique are based on the fact that the electrical conductivities of blood and brain tissue are different, so that the effective impedance of the mixture depends on the mixture ratio of both elements [2]. This technique has a number of features that make it of great interest in clinical practice: it is non-invasive and painless, available at bedside, provides continuous measurement and is cost-effective.

Nevertheless, despite the large number of previous reports [3], there are no studies on the cerebral regions reflected in REG recordings. Furthermore, REG capability to reflect local, regional or global CBF is still unknown. As a previous step to adopting REG as a clinical and research tool, it will be necessary to evaluate the REG sensitivity to local brain conductivity changes, as has previously been done in the case of EEG and MEG [4].

The main goal of this work is to obtain the sensitivity map of the tetrapolar REG, referred to as REG II, to cerebral electrical conductivity changes for the set of electrode arrangements used in previous works [5], [6]. For this purpose, a geometrical four-shell model of the head is used to derive the analytical solution of the sensitivity, which could be further used to inspect the influence of geometrical parameters on the solution.

II. Methods

A. Model

Head was modeled using the conventional four–shell spherical model, which represents from inner to outer: brain, cerebrospinal fluid (CSF), skull and scalp [6], [7] (Fig. 1). Henceforth, these layers will be labeled consecutively from 1 to 4. The outer radii and the mean isotropic conductivity of the layers were r_1=7.8 cm and σ_1=0.33 S/m; r_2=8.2 cm and σ_2=1.79 S/m; r_3=8.7 cm and σ_3=0.0042 S/m; and r_4=9.2 cm and σ_4=0.33 S/m.

To avoid the use of mixed boundary conditions on the outer surface of the model, current injection electrodes were modeled as two points on the scalp surface. The infinite set of tetrapolar electrode arrangements was limited by fixing the current injection points at opposite sites of the model and forcing symmetry of the arrangement to the central plane

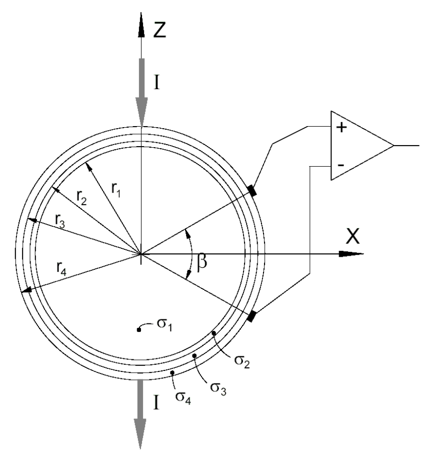

Fig. 1. Schematic representation of the four-shell head model used in this study. From the inner to the outermost, the spheres represent the brain, CSF, skull and scalp. Current injection was considered to be at different points on the scalp surface and diametrically opposed, whereas the pickup electrodes were placed keeping their symmetry with the *XY* plane and separated by some angle *β*.

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orthogonal to the current injection axis (Z). The angle between radial vectors of pickup electrodes will hereafter be referred to as β.

The REG II impedance signal was assessed as the voltage output of the differential amplifier when a unitary current was injected, as shown in Fig. 1.

B. Sensitivity of impedance to brain tissue conductivity

To assess the impedance change measured using an arbitrary REG II electrode arrangement when the electrical conductivity of an elemental volume changes differentially, we use the equation proposed by Geselowitz [8], which relates the measured impedance changes ΔZ with the conductivity variations of a particular region of a volume conductor being,

$$\Delta Z = -\Delta\sigma \int_v \frac{\nabla(\phi + \Delta\phi)}{I_\phi} \frac{\nabla(\theta)}{I_\theta} dv \tag{1}$$

where $\Delta\sigma$ is the increment of the volume conductivity; I_ϕ is the current injected through a pair of electrodes, and ϕ is the potential field generated by I_ϕ; I_θ and θ are the equivalent ones, but using a second pair of electrodes. The integral in (1) extends throughout the volume in which the conductivity change occurs.

For differential conductivity changes, (1) is expressed as

$$dZ = -d\sigma \int_v \frac{\nabla(\phi + d\phi)}{I_\phi} \frac{\nabla(\theta)}{I_\theta} dv \approx -d\sigma \int_v \frac{\nabla(\phi)}{I_\phi} \frac{\nabla(\theta)}{I_\theta} dv\,. \tag{2}$$

From (2), let us define a function of point S whose value is the impedance sensitivity to the electrical conductivity of an infinitesimal volume surrounding the point P:

$$S(P) = \frac{d^2 Z}{d\sigma\, dv} \approx -\frac{\nabla(\phi)|_P}{I_\phi} \frac{\nabla(\theta)|_P}{I_\theta}\,. \tag{3}$$

Therefore, function S is proportional to the dot product of the potential gradients at point P generated by the injection of a unit current through both pairs of electrodes, and reflects the contribution of the local conductivity changes at point P to the REG II signal.

C. Analytical solution for the model

Electric potential at a point in the innermost sphere created by the current injection can be expressed in terms of Legendre polynomials [9] as

$$U_1 = \sum_{n=1}^{\infty}\sum_{m=1}^{n} A_{mn}\, r^n \cos(m\phi)\, P_n^m(\cos\theta) \tag{4}$$

where U_1 is the potential at a point of coordinates (r, ϕ, θ) inside the sphere; and A_{mn} is the serial coefficient which depends on the geometrical and electrical parameters of the model (see Appendix).

Therefore, according to (3), the function S at a point P of the inner volume is

$$S_1 = -\nabla \sum_{n=1}^{\infty}\sum_{m=1}^{n} A_{mn}^{INJ}\, r^n \cos(m\phi)\, P_n^m(\cos\theta)\Big|_P \bullet \bullet \nabla \sum_{n=1}^{\infty}\sum_{m=1}^{n} A_{mn}^{REC}\, r^n \cos(m\phi)\, P_n^m(\cos\theta)\Big|_P \tag{5}$$

where the superscripts *REC* and *INJ* on the coefficient A_{mn} represent those obtained when current I is injected through pickup electrodes, or through the injection electrodes, respectively.

Sensitivity S_1 of REG II was calculated for those points P belonging to the plane that contains the electrodes, with a spatial resolution of 0.5 mm, and for β angles of 30º, 60º, 90º, 120º and 150º. Additionally, S_1 was also calculated for the plane orthogonal to the aforementioned one, but containing both the origin and the zones of maximum sensitivity.

D. Convergence

Sensitivity S1 is expressed in (5) as an infinite sum of terms. Therefore, an estimation of S1 is obtained by truncating the sum when the addition of the last 16 terms accounts for less than 1%, or up to a maximum of 200 terms.

III. RESULTS

Maps of the sensitivity of impedance to the regional brain conductivity are shown in Fig. 2 to Fig. 6 for β values ranging from 30º to 150º. Sensitivity values are negative for all the points and electrode positions under study.

The results show that highest sensitivity is located in the vicinity of the recording points, for any β angle. The influence of the current injection points increases for higher β angle values.

The maximum sensitivity area is located near the surface of the sphere, increasing its depth for closer recording electrodes, whereas the minimum sensitivity area is located, for any β angle, on the sphere surface, at the point opposite to the midpoint between pickup electrodes.

The ratio between the maximum and minimum value of sensitivity S_1, identified as the heterogeneity index (H_1), is shown in Table 1.

TABLE 1
HETEROGENEITY INDEX OF SENSITIVITY

Recording angle β (º)	$S_{1\,MAX}$ (Ω^2/m^2)	$S_{1\,MIN}$ (Ω^2/m^2)	H_1
30	11911	406	29.4
60	12433	1463	8.5
90	14533	2228	6.5
120	19994	3119	6.4
150	33994	4162	8.2

Index H_1 keeps a quite homogeneous range, varying from 6.4 to 8.5, with the exception of β=30º, for which it rises to 29.4.

IV. DISCUSSION

Sensitivity maps show that, even for a simple four shell

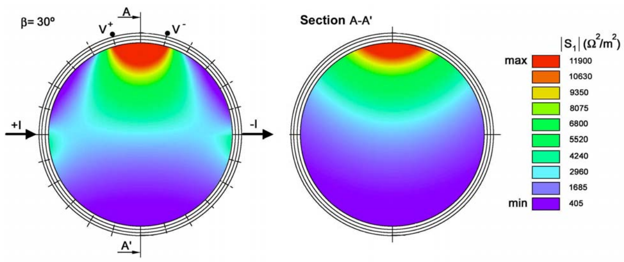

Figure 2. Sensitivity to the impedance registered by a REG II for a recording electrode angle of β=30º.

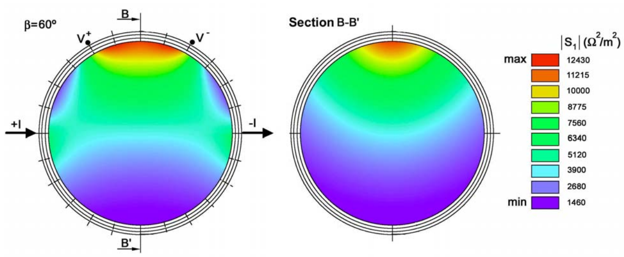

Figure 3. Sensitivity to the impedance registered by a REG II for a recording electrode angle of β=60º.

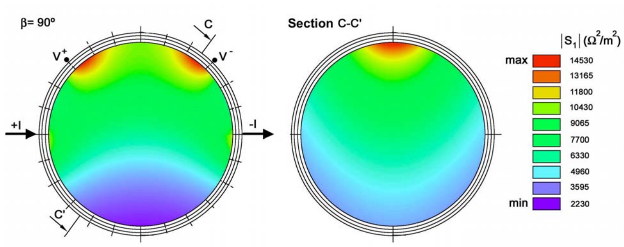

Figure 4. Sensitivity to the impedance registered by a REG II for a recording electrode angle of β=90º.

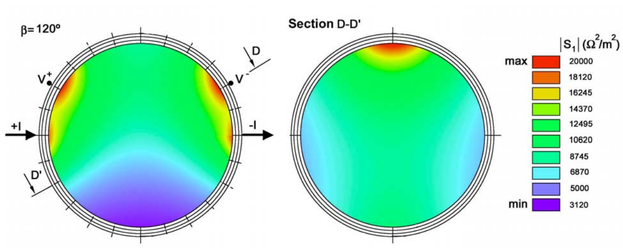

Figure 5. Sensitivity to the impedance registered by a REG II for a recording electrode angle of β=120º.

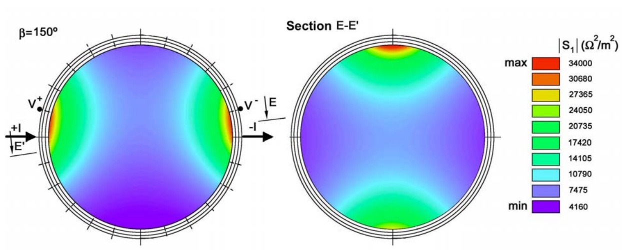

Figure 6. Sensitivity to the impedance registered by a REG II for a recording electrode angle of β=150º.

spherical model, local changes in electrical conductivity induce different impedance variations depending on the point and on the electrode arrangements. This fact could be exploited to focus the REG measurements on a specific brain region, mainly in cortical areas. According to the H_1 index, focal measurements are best achieved with the narrowest β angles.

This suggests a potential application of REG to local evaluation of cortical cerebral perfusion, similar to optical methods such as NIRS.

This study could also help to explain the paradoxical behavior of REG throughout the cardiac cycle. Traditionally, the conductivity values used in head models are of 0.33 S/m for the brain, 0.66 S/m for blood, and 1.79 S/m for CSF. Intracranial space is occupied by these components and, according to the Monroe-Kelly theory, an increasing volume of any element displaces an equivalent volume of the other components. Under normal conditions, entering blood must evacuate CSF from the intracranial space. Therefore, head impedance should increase, instead of attenuating, as was observed in practice.

This behavior of the impedance registered by an REG could be due, according to Lifshitz [2], to a redistribution of the spaces occupied by blood and CSF in the cranial cavity. Blood and CSF displacements have been studied by means of movement-sensitive magnetic resonance imaging. These studies [10] show that the brain shifts sharply in the systole, and returns progressively during diastole to its initial state. Basically, there is no movement of the brain above the *corpus callosum* in head-closed subjects. In addition, important displacements of blood and CSF occur during the cardiac cycle, when the arterial systolic blood supply is accompanied by venous outflow and CSF migration to the spinal space [11].

The results obtained in this study show a higher sensitivity of the measurement for the points located on the cortex, near the pickup electrodes. On the other hand, the brainstem, where main CSF displacement is produced, is located in the lowest sensitivity areas. These facts suggest that the impedance reduction of the REG signal could be due to the blood filling high sensitivity areas while CSF is is reduced in those of low sensitivity. Moreover, the uniform distribution of blood in the brain tissue, instead of the localized volumes for CSF, could reinforce this different behavior [2].

Finally, the conclusions derived from these results, outside the established framework, may be imprecise. Although a more realistic model would give results nearer to experimental data, geometrical models provide analytical and parametric solutions that are non-dependent on the anatomical peculiarities of the subject. The results presented in this work therefore provide an outline of the spatial distribution of REG sensitivity.

APPENDIX

In order to derive the necessary equations to solve the model, it must be kept in mind that the potential field created by the current injection has to satisfy Laplace's equation at any inner point of the four spheres [9].

Due to the electrode location with respect to the axis, the solution is an even function of ϕ, so that the solution can not contain terms in $sin(m\phi)$.

The innermost sphere can not contain $1/r$ terms, because this region includes the origin, so potential for this region is in the form

$$U_1 = \sum_{n=1}^{\infty}\sum_{m=1}^{n} A_{mn}\ r^n\ \cos(m\phi)\, P_n^m(\cos\theta)\cdot$$

For the other shells (j=2, 3, 4), potential is in the form

$$U_j = \sum_{n=0}^{\infty}\sum_{m=0}^{n}\left(B_{mn,j}\ r^n + C_{mn,j}\ \frac{1}{r^{n+1}}\right)\cos(m\phi)\, P_n^m(\cos\theta)\cdot$$

To assess these coefficients, it is sufficient to impose Dirichlet and Neumann boundary conditions on the appropriate surface of spheres. The former requires that the potential at every point of the outer surface of domain *i* must be identical to the potential at the same point of the inner surface of domain *i+1*. On the other hand, the Neumann boundary condition establishes that the current density that emerges normal to every differential of the outer surface of domain *i* must be identical, but opposite to that of the inner surface of domain *i+1*. The seventh and last equation can be obtained by applying the Neumann boundary condition to the outermost surface of the model. On this surface, the radial current density should be nil at every point, with the exception of the current injection points, where it is infinite.

This equation system was solved by using the software Mathematica (Wolfram Research, Inc. Illinois, USA) and is easily reproducible.

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