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Extraction of the Intracranial Component from the Rheoencephalographic Signal: A New Approach

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***Abstract*—The well-known inherent artifact on the rheoencephalogram (REG) caused by the pulsatility of the scalp blood flow left the REG out of the clinical practice. In fact, depending on the selected electrode arrangement, the measurement of the brain impedance changes time-locked with the heartbeat can be completely buried on that of the scalp. In this work, a novel mathematical method based on the physiological differences between the brain and scalp perfusions is proposed to extract the intracranial information from REG. This method is experimentally applied to REG signals recorded at five electrode positions and results are compared with those derived from our previous theoretical works. Intracranial components extracted from the REG signals are consistent with the stated hypothesis and reproduce the unexpected results obtained with our theoretical models. Although further studies would be needed, the evidences found in this work suggest that the method proposed in this work extracts the intracranial information from the REG signal.**

I. Introduction

A close surveillance of cerebrovascular parameters in neurological pathologies is essential to prevent secondary brain insults. A persistent increase of intracranial pressure caused by oedema, tumours or haematomas may decrease the cerebral blood flow (CBF) to values above the minimum required for neuronal survival [1].

As a method to evaluate CBF, Polzer and Schuhfried proposed in 1950 to apply the well-known plethysmographical techniques to the head, which was specifically referred to as Rheoencephalography (REG) [2].

For this purpose, bipolar and tetrapolar electrode arrangements (named REG I and REG II respectively) were used to measure from the scalp surface the electrical impedance changes of the head time-locked with the heartbeat. The REG signal was assumed to be induced by the time-varying nature of the mixture ratio of two biological conductors with different electrical conductivity; i.e. blood and brain. However, some experimental [3], [4] and theoretical [5] works suggested that most of the REG signal was actually caused by the pulsatile blood flow of the extracranial tissues.

In this sense, our previous theoretical results obtained from head models suggest that (i) most of the REG I signal is caused by the pulsatility of the extracranial blood flow [6]; (ii) in some subjects, a set of tetrapolar electrode arrangements may exist to record a REG II free of extracranial information; and (iii) such electrode positions and even their existence are strongly dependent on the subject's physical constitution, particularly on their scalp thickness. In summary, our previous theoretical results suggest that there is no universal electrode arrangement suitable for all individuals to register a REG II free of extracranial contamination.

Therefore, a REG II recorded from an arbitrary tetrapolar electrode arrangement contains information from both intra and extracranial blood flow, mixed in unknown proportions which, furthermore, depend on the used electrode arrangement. One unexplored way of extracting intracranial information from a REG signal could be to approach the problem under the Blind Source Separation (BSS) perspective [7]. In this manner, observations would comprise two or more REG signals recorded simultaneously from different electrode arrangements, and the intra and extracranial components would be the sources to be separated. However, no statistical properties of the extra and intracranial components are known to be exploited to separate one source from the other.

Unlike conventional BSS approaches which are based on the statistical differences of the sources, in this work we propose to use the physiological differences between both components as an approach to solve the mixture problem.

II. Methodology

A. Data model and separation method

Let $R_1[i]$ and $R_2[i]$ be a pair of REG signals recorded simultaneously from a given subject at different electrode arrangements. Assuming the morphological invariability and negligible phase shift of both REG components [8], we can write

$$R_1[i] = a_{11}C_{En}[i] + a_{12}C_{In}[i] \quad (1)$$

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$$R_2[i] = a_{21} C_{En}[i] + a_{22} C_{In}[i], \quad (2)$$

where $C_{En}[i]$ and $C_{In}[i]$ are, respectively, the extra and intracranial components of the recorded REG signals normalized to unit variance (sources), and a_{ij} are the mixture coefficients that depend on the electrode arrangement. It should be noted that both components and coefficients of the mixture matrix are unknown. Now, subtracting an arbitrary ratio K of (1) from (2), we can define a function $F_K[i]$ as

$$\begin{aligned} F_K[i] &= R_2[i] - K\, R_1[i] = \\ &= (a_{21} - K\, a_{11}) C_{En}[i] + (a_{22} - K\, a_{12}) C_{In}[i]. \end{aligned} \quad (3)$$

Then, the goal is to find a K^* value that cancels out the extracranial component term in (3). For this purpose, we exploit the following difference between the intra and extracranial blood perfusion. On the one hand, the quick rise in arterial blood pressure that follows a heartbeat causes an abrupt increment of the extracranial blood volume, which is responsible for the sharp drop of the extracranial component [8]. In fluid dynamics terms, the extracranial blood flow will show a markedly pulsatile waveform. On the other hand, fast increases in cerebral blood volume should not be allowable within the rigid cranial structure and, therefore, CBF should show, comparatively, a smoother waveform. According to this, since K modulates the amount of extracranial blood volume information contained in $F_K[i]$, the searched K^* value will give a minimum value of the function cost

$$J(K) = E\left\{\left(R_2[i] - R_2[i-1] - K(R_1[i] - R_1[i-1])\right)^2\right\}. \quad (4)$$

Then, the K^* value that minimizes (4) will transform (3) into

$$F_{K^*}[i] = R_2[i] - K^*\, R_1[i] = (a_{22} - K^*\, a_{12}) C_{In}[i] \quad (5)$$

Although this procedure could be used to extract the morphology of the intracranial component from two REG II signals, a more interesting result can be obtained by using a REG I signal as $R_1[i]$. In this case, and assuming that the REG I signal is caused exclusively by the extracranial blood flow, the a_{12} term in (1) can be considered negligible and $F_{K^*}[i]$ becomes the intracranial component of $R_2[i]$. In this work, we applied this method on a set of five pairs of REG I and REG II signals.

B. Experiment description

In this section, a preliminary evaluation of the proposed method was performed experimentally on one subject. Twelve electrode sites were properly marked and cleaned on the subject's scalp surface: C5; C6; five equidistant electrode sites between C5 and Cz (labelled as Ca, Cb, Cc, Cd and Ce); and other five between C6 and Cz (labelled as Ca', Cb', Cc', Cd' and Ce') (Fig. 1). Current injection electrodes were permanently fixed at C5 and C6, whereas the pair of REG II pickup electrodes was moved between recordings. An additional pair of electrodes to record ECG (Lead I) was used for synchronization purposes.

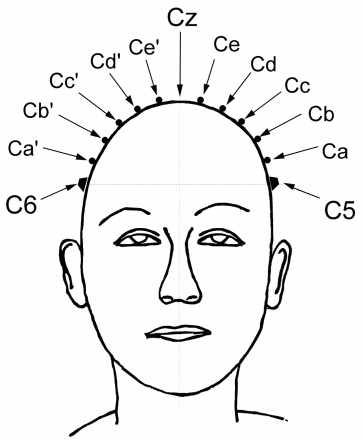

Fig. 1. Schematic representation of the electrode positions used in the experimental section. Tetrapolar impedances were measured by injecting current through C5 and C6 electrodes, whereas electrode pairs symmetrical to the coronal plane were used for voltage measurement.

Five records of REG I, REG II and ECG channels were acquired during three minutes. Whereas the current injection electrodes were kept at C5 and C6 during all the experiment, the pairs of pickup electrodes used in each recording stage were placed at Ca-Ca'; Cb-Cb'; Cc-Cc'; Cd-Cd'; Ce-Ce'. This set of REG recordings matches well with the studied in our previous theoretical works [6], [9].

To improve signal-to-noise ratio, REG sweeps time-locked with the R wave of the ECG were averaged using a time-window extended from 100 ms prior to the R wave to 700 ms after it. The aforementioned method was used to extract the intracranial component of the REG II signal of each pair of the five averaged data sets.

III. Results

Averaged REG I waveforms are shown overlapped in Fig. 1. For clarity, the traces have been normalized to unit variance, so they can be read as $C_{En}[i]$ components if the premise of the REG I pure extracranial origin is assumed. The traces show a morphology very close to that previously identified as REG extracranial component [8]. The time interval in which the impedance falling edge occurs is marked on Fig. 1 and transferred to the next figures for comparison. As it could be expected, the morphology of the REG I traces is practically constant along the recordings since injection electrodes keep invariable throughout the experiment.

On the contrary, a high variability of the $R_2[i]$ traces can be observed in Fig. 2. Furthermore, the mean and dispersion (α=0.05) of the correlation coefficient between each pair of $R_2[i]$ traces are 0.6949 (min= - 0.6069; max= 0.9843). Within the time interval of the $R_1[i]$ falling edge previously

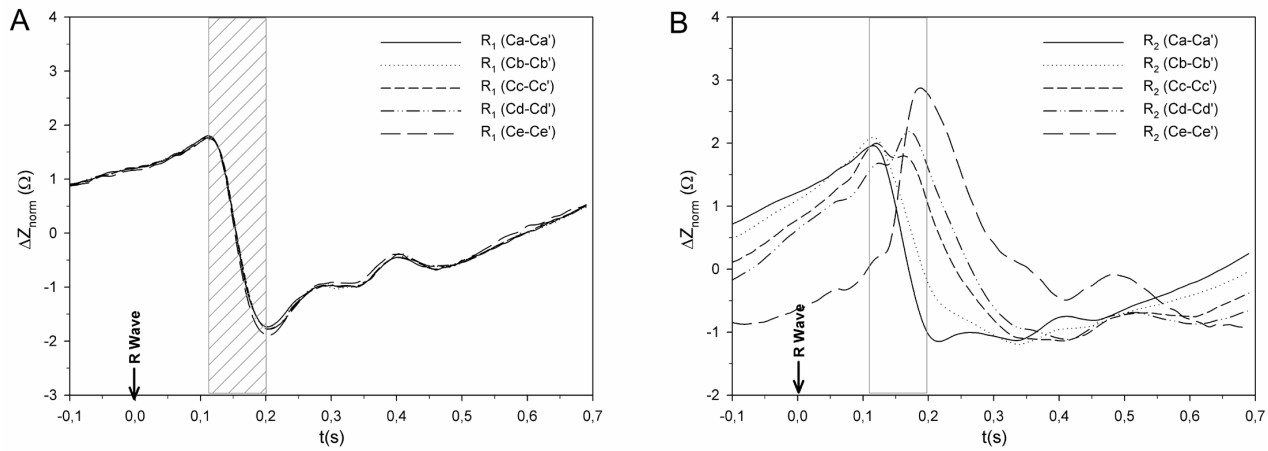

Fig. 2. Set of five averaged REG waveforms recorded from the subject and normalized to unit variance. (A) REG I traces and time interval during which REG I impedance falls. (B) Overlapped REG II traces and previous time interval transferred from A.

described in Fig. 1, replicas of that edge can be observed in the two electrode pairs closer to the injection one (Ca-Ca' and Cb-Cb'). Moreover, a falling edge of the impedance trace can also be observed, but inverted, in the impedance measurement recorded at the farthest electrode pair from the injection one (Ce-Ce').

Finally, the intracranial $C_{In}[i]$ components extracted from $R_2[i]$ traces show similar morphologies (Fig. 3). In these traces, the mean and dispersion (α=0.05) of the correlation coefficient between each pair of $C_{In}[i]$ traces are 0.9813 (min= 0.8763; max= 0.9973). Within the $R_1[i]$ time interval referred in Fig. 1, no sign of the falling edge is observed, the waveforms being similar to the $R_2[i]$ traces recorded at intermediate electrode positions (Cc-Cc' and Cd-Cd').

IV. Discussion

Despite the developments on tomographical techniques, the idea of a non-invasive, portable and cost-effective device that evaluates continuously the CBF at bedside still seems attractive. Research on REG could meet these requirements, but was limited because of the inherent extracranial contamination of the signal.

The experiment described in this work has been designed to validate our previous theoretical results on REG using a spherical model of the head, by injecting an electric current at opposite sides of the head and moving the pickup electrodes symmetrically to the coronal plane [6][8]. Both works predict that, for some subjects, the extracranial component of a REG II could be nil for some pickup electrode positions and, furthermore, may be inverted when both electrodes are brought together. This result, that can be hardly explained theoretically, is also found experimentally in the REG II recorded at Ce-Ce'. Therefore, assuming that the extracranial component of a REG II signal predominates at Ca-Ca', we should be able to find a pure non-extracranial REG signal at intermediate electrode positions.

On the other hand, although the conventional BSS methods use statistical properties to separate the sources of the observed signals, we propose to exploit the physiological differences between the intra and extracranial component to separate one from the other. In this sense, our approach is based on two premises: (i) most of the REG I signal reflects the extracranial perfusion, and (ii) the changes of the cerebral blood volume are markedly smoother than those of the extracranial tissues.

In this sense, the assumption that most of the REG I is caused by the scalp blood flow agrees well with early reports (e.g. [3]) and, even if some intracranial signal were buried in the REG I, it would only affect the amplitude of the result but would not modify the morphology of the intracranial component. With respect to the second premise, although we have no knowledge of comparative studies between intra and extracranial perfusion processes, it seems reasonable to think that expansion of the cerebral arteries is constricted by the cranial structure, as it can be inferred from radiological studies about brain and cerebrospinal fluid dynamics [10], [11].

To the best of our knowledge, this is the first time that a

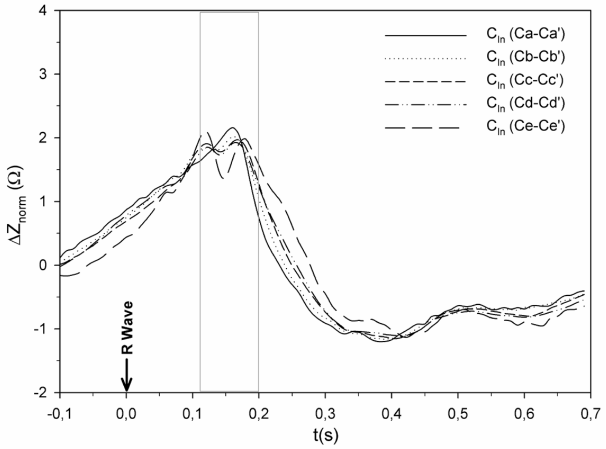

Fig. 3. Intracranial components derived from each pair of REG signals by the proposed separation method. Boxed time interval has been transferred from that shown in Fig. 2A.

mathematical model has been proposed to extract the intracranial information from a REG signal. In our opinion, several facts suggest an intracranial origin of the extracted component: in particular, the inversion of the extracranial component of the REG II at Ce-Ce' suggests the existence of an electrode arrangement, probably in the neighbouring of Cc-Cc' or Cd-Cd', from which a pure intracranial component could be directly recorded; the intracranial components generated by the model are morphologically similar among them and to those recorded directly at Cc-Cc' and Cd-Cd'; their morphologies coincide with that reported in previous works in which the cerebral impedance changes were measured using intracranial electrodes [12]; and, finally, the temporal derivative of the extracted component is similar to a typical transcranial Doppler trace.

However, further studies on the signal extracted with this novel method are needed to show that, as we suggest, this signal could reflect exclusively the intracranial perfusion. Moreover, since the measure variable is the intracranial impedance changes associated with the cardiac cycle, additional studies should be required to find the appropriate signal parameters that could be useful in clinical practice.

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