Canonical Decomposition of scalp EEG as preprocessing for source localisation.

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ABSTRACT

Long-term electroencephalographic (EEG) recordings are important in the presurgical evaluation of refractory partial epilepsy for the delineation of the ictal onset zones. In this paper we introduce a new concept for an automatic, fast and objective localisation of the ictal onset zone in ictal EEG recordings and show with a simulation study that canonical decomposition of ictal scalp EEG allows a robust and accurate localisation of the ictal onset zone.

I. INTRODUCTION

Epilepsy is one of the most common, severe neurological diseases. People suffering from epilepsy, who are not helped by medication, can potentially benefit from epilepsy surgery [6]. In order to remove the epileptogenic region, a precise localisation of the epileptic focus is mandatory. One of the diagnostic tools to localize this region of seizure onset zone is recording of ictal scalp electroencephalogram (EEG) [13]. The EEG measures electric potential distributions at discrete recording sites on the scalp. EEG recordings have an excellent temporal resolution, but a rather poor spatial accuracy due to the limited number of recording sites and the shielding effect of the skull. A challenging problem in neuroscience is to estimate in a more objective and precise way the regions of the brain that are active, given only the measured potential distributions.

Estimating the electrical source in the brain from the scalp EEG is a difficult problem since an infinite number of internal electrical currents can generate the same potential distribution on the scalp. Several different approaches to solve this source localisation or inverse problem exist based on different assumptions [1], [11]. Dipole modeling is a well-established technique for localising interictal spikes, see e.g. [10], [9] and references herein. Ictal EEG recordings have been subjected to dipole modeling much less often than interictal spikes. The seizure discharge is a very complex pattern. Mainly artifacts render modeling difficult [7]. Moreover, the low signal to noise ratio of the seizure signal can render the correct localisation very diffuse. However, when source localisation of seizure onset would be possible, it can reduce the need for invasive intracranial EEG recordings. So far, the results of ictal EEG source localisation have been discouraging, except in [5]. The localising value of dipole modeling of ictal EEG can be improved by first removing artifacts and afterwards estimating the sources [8]. Another possibility is to decompose the measured EEG in a sum of individual contributions of distinct brain sources and localising the epilepsy-related source(s) in order to estimate the epileptic focus. Recently, we have shown that a space-time-frequency decomposition of a three way array containing wavelet transformed EEG by the Canonical Decomposition (Candecomp), also known as Parallel Factor Analysis (Parafac), reliably separated a seizure atom from the noise and background activity with a sensitivity of more than 90 % [4]. This research was inspired by [12]. After the decomposition, the potential distribution over the electrodes of the epilectical activity was obtained, and displayed as a 2D image. Electrodes with large potential amplitudes could be considered as close to the focus. The aim of the present study was twofold. First, we wanted to investigate whether it was possible to localise the ictal onset zone in the head by applying dipole source localisation after Canonical Decomposition

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of ictal EEG recordings. Second, we wanted to investigate the accuracy of this localising method with realistic simulations under different conditions. We were especially interested (i) in the influence of the signal-to-noise ratio (SNR) of the seizure activity on the localisation, (ii) how the dipole localisation would be influenced by changes in frequency, and (iii) if multiple dipolar sources could be reliably estimated. A larger simulation study is described in [3].

We start by revising the Canonical Decomposition of a higher-order array (§II-A). We then define how we constructed realistically simulated EEG (§II-B), assessed the accuracy of our method (§III) and finally discuss our results (§IV).

II. Materials and Methods

A. Method

In our application, a three-way data array $\mathcal{X}$ with dimensions (space, scale, time) is obtained by wavelet-transforming every channel of the original EEG matrix. The continuous wavelet transform C at scale a and time t of a signal $x(t)$ is defined as

$$C(a, time) = \int_{-\infty}^{\infty} x(t)\phi^{\star}(a, time, t)dt \quad (1)$$

with $\phi^{\star}$ the chosen wavelet. Different real wavelets can be used. In this study, we used a biorthogonal wavelet with decomposition order 3. From the scale a of the wavelet, the frequency f of the signal can be estimated as:

$$f = f_c/(a\Delta t) \quad (2)$$

with f_c the center frequency of the wavelet and Δt the sampling period.

Candecomp [14] is a generalisation of the singular value decomposition (SVD). For a three-way array $\mathcal{X}(I \times J \times K)$ it is defined as:

$$x_{ijk} = \sum_{r=1}^{R} a_{ir}b_{jr}c_{kr} + e_{ijk} \quad (3)$$

where R is the number of components used in the Candecomp model and e_{ijk} are the residuals containing the unexplained variation. A pictorial representation of the Candecomp model is given in Figure 1. The Candecomp model is a trilinear model: fixing the parameters in two modes, x_{ijk} is expressed as a linear function of the remaining parameters.

The Canonical Decomposition is usually computed by means of an Alternating Least Squares (ALS) algorithm [14].

When Candecomp is used for seizure localisation, 2 seconds of EEG at the seizure onset is wavelet transformed. The obtained three-way array is decomposed with Candecomp with R atoms. Several techniques exist to determine the optimal number of atoms [14]. We used Corcondia. After decomposition, the 'epileptic' atom is well localised in space and frequency. The spatial component of the epileptic atom provided localising information of the focus [4].

Dipole estimation then determines the (*one*) dipole's coordinates and orientation that best generates the given potential distribution in a least squares sense. For computational simplicity, we used a spherical head model in this study.

B. Simulation

Consider a matrix $\mathbf{X}$ of dimension 500-by-21 representing a 21 channel EEG section of 2.0 s long. Each vector $\mathbf{x_s}, s = 1, \ldots, 21$ of $\mathbf{X}$ contains the time course of an EEG channel:

$$\mathbf{X} = [\mathbf{x_1}, \mathbf{x_2}, \ldots, \mathbf{x_{21}}]^T. \quad (4)$$

In this simulation study $\mathbf{X}$ includes both seizure activity, and superimposed noise. Both signals are described below.

1) Synthetic seizure activity: The EEG of the ictal activity was generated using a fixed dipole in a three-shell spherical head model (conductivities 0.33/0.2/0.33 S/m). The different time courses generated by the dipole are described below. The amplification factors at each electrode were computed by solving the forward problem for a dipole in a three-shell spherical head model consisting of a brain, a skull and a scalp compartment. 21 electrodes were placed according to the 10-20 system for electrode placement and additional sphenoidal electrodes.

Unless otherwise stated, dipole coordinates x (left ear to right ear), y (posterior to anterior) and z (up, through the Cz electrode) were [-0.5 0 0.1] and the dipole orientations $d_x,\ d_y$ and d_z were [1 0 0].

Following seizure characteristics were simulated:

- Seizure activity in patients with epilepsy is typically expressed by a sinusoidal waveform. In a first simulation we estimated the dipole localisation error when seizure activity was represented by a 4 Hz sinusoid at different noise levels (figure 2).
- Epileptic seizure activity can rapidly change in frequency. Ictal EEG activity is often characterized by low voltage fast activity in the beta range which gradually slows down to

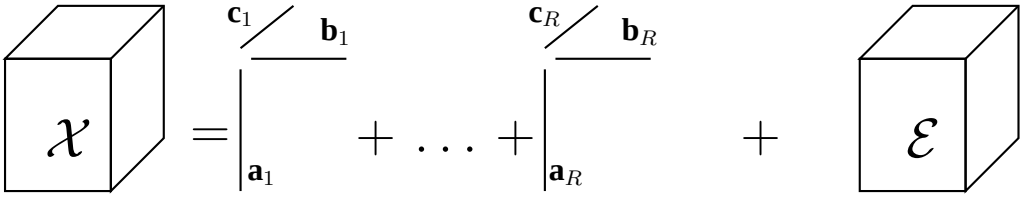

Fig. 1. The Candecomp model with R components.

alpha or theta frequencies with increasing amplitude. The Canonical decomposition exploits frequency information during the decomposition. In a second simulation, we wanted to estimate the dipole localisation error when the frequency changed during the 2 seconds under investigation. We simulated a chirp that linearly changed in frequency from 16 Hz at the start to 8 Hz at the end of the considered 2 seconds. The signal also doubled in amplitude.

- In our previous study [4], two atoms were obtained after the decomposition of *in vivo* seizures and a distinction could be made between a seizure and a non-seizure atom. An interesting question is whether seizure activity will always be represented by one atom and thus 1 dipole. In a third simulation, we considered also two rhythmical sources firing at the same frequency separated from each other by about 1 cm: the second dipole had coordinates [-0.4 0 0.1]. These dipoles generated similar potential distributions at the scalp.

2) Noise: A 500-by-21 noise matrix $\mathbf{B}$ contained 2 seconds of awake background EEG activity, recorded with the same electrode configuration as in (A), from a normal subject. On this matrix $\mathbf{B}$, muscle artifacts were superimposed. These muscle artifacts were separated from contaminated background activity using BSS-CCA [2].

3) The simulated signal: In the simulation study the noise matrix $\mathbf{B}$ is superimposed on the signal matrix $\mathbf{A}$ containing the epileptical activity:

$$\mathbf{X}(\lambda) = \mathbf{A} + \lambda \cdot \mathbf{B}, \tag{5}$$

with $\lambda \in \mathbb{R}$. The Root Mean Squared (RMS) value of the signal is then equal to

$$RMS(\mathbf{A}) = \sqrt{\frac{1}{S \cdot N} \sum_{s=1}^{S} \sum_{n=0}^{N-1} (\mathbf{A}(n,s))^2}, \tag{6}$$

with N the number of time samples. The signal-to-noise ratio (SNR) is then defined as follows,

$$SNR = \frac{RMS(\mathbf{A})}{RMS(\lambda \cdot \mathbf{B})}. \tag{7}$$

Changing the parameter λ alters the noise level of our simulated signal.

III. Results

Figure 2 shows the dipole localisation error for different SNRs when Candecomp and SVD are used prior to dipole modeling. It can be seen that Candecomp gives a much more reliable dipole estimate. At a SNR of 0.7, the error between the simulated and the fitted dipole was only 5 mm.

Figure 3 shows the dipole localisation error in function of the SNR when the simulated epileptic signal changed in frequency and amplitude during the considered 2 seconds. This figure is very similar to figure 2. This means that, although the seizure signal is not well localised in frequency, the decomposition still reliably detects the correct location. When we looked at the frequency component of the epileptic atom, this component seemed to be localised around 12 Hz, i.e. the average of the start (16 Hz) and end frequency (8 Hz), while the frequency component in simulations with fixed frequency peaked at the simulated frequency. Also the Candecomp fit percentage in this simulation was lower than the fit percentage in simulations with fixed frequency, because the signal was not perfectly trilinear anymore.

Figure 4 shows in (a) the simulated localisation of two close dipoles and in (b) the estimated localisation with the proposed method when 3 atoms were estimated. SNR was 0.7. The localisation error was for both sources about 5mm, which indicates that a reliable separation and localisation of multiple dipoles was obtained.

IV. Discussion

We present here the framework for seizure onset localisation with Candecomp as preprocessing step for EEG source localisation. We have shown that in a spherical head model with realistically simulated EEG, our algorithm correctly localised the seizure-related atom with an accuracy smaller than 1 cm, even at SNR ratios that are lower than one encounters during real ictal recordings. The second simulation investigated a more challenging, but maybe more realistic situation in which the frequency of the seizure changed during the considered time interval. The resulting atom could not fully capture the exact frequency-varying signal, as indicated by

a lower fit-percentage of Candecomp. However, the best tri-linear approximation still reliably localised the signal. We investigated also the situation in which two dipolar sources generating the same signal were placed near each other. The Corcondia [14] indicated that three atoms was the correct number of atoms for this simulated EEG. Two of them corresponded to the 2 dipolar sources. When matrix decomposition techniques like SVD or Independent Component Analysis (ICA) are used, only 1 rhythmical source would be extracted as the 2 simulated sources are not independent or uncorrelated. This example illustrates the interesting uniqueness property of Candecomp [14] for EEG source localisation. Tensor decomposition techniques clearly outperform matrix decomposition techniques as preprocessing technique for EEG source localisation.

In the future, we plan to validate our method on *in vivo* seizures with a gold standard.

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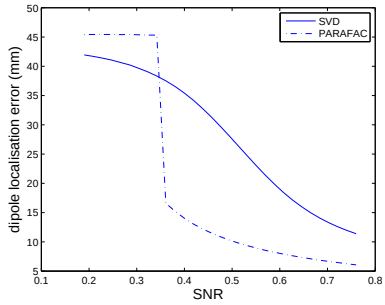


Fig. 2. The dipole localisation error as a function of the seizure frequency

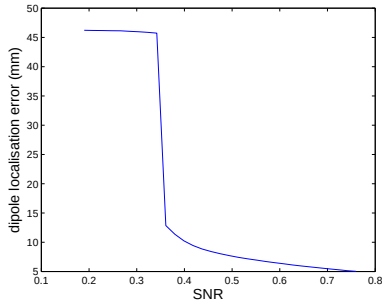
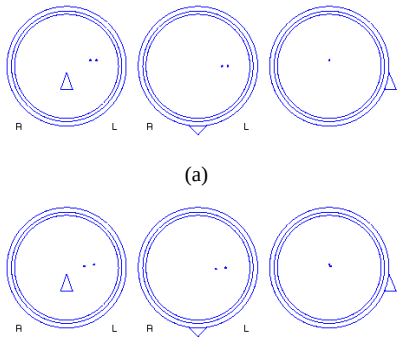


Fig. 3. The dipole localisation error as a function of the noise level, when the seizure activity is changed in frequency during the time interval under investigation



(a)

(b)

Fig. 4. (a) The original dipole localisation of two simulated dipoles. (b) The dipole localisation when three atoms were estimated with Candecomp