Published in final edited form as:

Int Conf Digit Signal Process Proc. 2011 July 6; 2011: . doi:10.1109/ICDSP.2011.6004976.

ESTIMATION OF DIRECTIONAL BRAIN ANISOTROPY FROM EEG SIGNALS USING THE MELLIN TRANSFORM AND IMPLICATIONS FOR SOURCE LOCALIZATION

Catherine Stamoulis* [Member IEEE] and

Harvard Medical School, Children's Hospital Boston, Departments of Neurology and Radiology, Clinical Research Program, 300 Longwood Ave., Boston MA 02115, USA

Bernard S. Chang

Harvard Medical School, Beth Israel Deaconess Medical Center, Department of Neurology, 330 Brookline Ave., Boston MA 02215, USA

Abstract

This paper presents a novel approach for the estimation of frequency-specific EEG scale modulations by the directional anisotropy of the brain, using the Mellin transform [1, 2, 3]. In the case of epileptic sources, the activity recorded by routine scalp EEG includes contributions not only from a seizure's primary propagation path but also from secondary paths and unrelated to the seizure activity. In addition, the anisotropy of the brain directionally modulates the seizure-related signal component. We estimated patient-specific direction-specific, frequency-locked scale shifts. During the ictal interval, these shifts occurred at frequencies ~ 50 Hz. We further estimated the effect of scale modulations on time-delay estimation. Larger time-delays were estimated from EEGs that had been corrected by a scale factor prior to this estimation. Thus, corrections for non-linear scaling of EEGs may ultimately improve time-delay estimation for source localization, particularly in cases of seizures rapidly propagating to large areas of the brain.

Keywords

Mellin transform; Source localization; Epileptic EEG

1. INTRODUCTION

Localization of epileptic sources from non-invasive (scalp) electroencephalograms (EEG) has been an area of active research for decades [4]. However, a number of technical and conceptual problems have limited accurate localization of the epileptic focus from scalp EEGs. First, routine systems, such as the international 10-20 EEG system, sample the surface of the brain only sparsely, limiting the localization to the spatial resolution of the electrode grid. In addition, scalp EEGs measure neural activity from relatively large areas of the brain. Thus, not only seizure-related activity from primary and possibly secondary propagation paths, but also contributions from unrelated brain activity may be superimposed in EEG signals. Finally, a large number of source localization studies that involve solutions of the inverse problem, are based on the assumption that the epileptic source is adequately represented by a dipole model. However, the epileptic source current density distribution is complex and dipole models may be too simplistic to adequately represent the source [5]. Consequently, the assumption of models for this ill-posed inversion, that are not

*caterina@mit.edu, caterina.stamoulis@childrens.harvard.edu..

directly derived from the data, may lead to non-unique and inaccurate localizations. Finally, not only directional (non-dipolar) seizure propagation but also the structural/electrical anisotropy of the brain modulate the EEG signal and affect source localization. Epileptic seizures are often caused by structural brain abnormalities, including brain malformations and tumors, but it is unclear how these modulate EEGs measuring seizure activity. A few studies have investigated the effects of tissue anisotropy on source reconstruction from EEGs using simulations. It has been shown that anisotropic white and gray matter, the skull anisotropy and source depth-dependent anisotropic tissue all affect the generated electric field and consequently EEG signals [6], particularly their amplitude [7]. Direct estimation of the effects of structural/electrical anisotropy from measured EEGs has been limited, since it is difficult to decouple signal components primarily modulated by directional anisotropy. Thus, potential non-linear scale changes in EEGs due to the directional anisotropy induced by structural abnormalities have not been studied extensively.

In this study, we investigated the directional modulation of ictal scalp EEGs using the Mellin transform, which has the desirable property of scale invariance [1, 2, 8, 3]. Specifically, we hypothesized that specific EEG components, identified via signal decomposition, are scaled versions of the source signal. Thus, the Mellin transform may be used to estimate the scaling factor associated with the directional, anisotropy-related modulation of this signal. For this purpose, we analyzed EEGs from 5 patients with diagnosed focal epilepsy and multiple seizures. We first decomposed filtered, artifact-free signals into principal components in order to decouple contributions from primary and secondary propagation paths, and to isolate the primary contribution of the source to the EEG signal. Then, the Mellin transform was applied to each EEG spectrum. In essence this is a problem of estimating signal features associated with a single (or dominant) source from multiple EEGs, and corresponding scale shifts associated with signal modulation by the directional anisotropy of the brain, along each source-electrode direction. Finally, we assessed the effect of incorporating a scale correction in the estimation of time delays between EEGs, which has important implications for source localization. Although the Mellin transform has been applied to a wide range of problems in image and signal processing, to the best of our knowledge it has not been previously applied for estimating anisotropy-induced EEG signal modulations, particularly in the presence of structural abnormalities.

2. METHODS

2.1. Data

Ictal scalp EEGs from 5 patients with diagnosed focal epilepsy, abnormal structural imaging (MRI), and at least 2 seizures (a total of 24 seizures) were analyzed. All EEG data were recorded at Beth Israel Deaconess Medical Center, Boston MA, in the Clinical Neurophysiology Laboratory of the Comprehensive Epilepsy Center. All data were part of inpatient clinical electrophysiology studies for patient evaluation and monitoring. Data were recorded with a standard international 10-20 EEG system, a referential montage and 500 Hz sampling rate. Power-line noise was attenuated with a stopband filterbank, centered at the 60 Hz harmonics of the noise, in the range 60-250 Hz, with a 1 Hz bandwidth for center frequencies < 150 and a 2 Hz bandwidth for center frequencies > 150 Hz. Second order elliptical filters (10 dB attenuation in the stopband, 0.5 dB ripple in the passband) were used. Signals were filtered in both forward and reverse directions to eliminate potential phase distortions due to the non-linear phase of the filter. Artifacts associated with eye blinking and muscle movement also contaminate EEGs. Studies have used a wide range of methods to eliminate them, including a number of decomposition and source separation methods [9, 10]. In this study, these artifacts were eliminated using a stopband-type matched-filter. The respective signatures of the artifacts were estimated from their multiple occurrences in the

data, during quiescent (baseline) periods, and were used as templates for pattern matching. Matched-filtering then identified time intervals of higher correlation between the templates and the data, during pre-ictal and ictal intervals, and the data signal-to-noise ratio (SNR) was subsequently reduced within the respective template bandwidths, to suppress the artifacts [11].

Based on our previous studies [12], analyzed seizures appeared to modulate baseline neural activity at least several minutes prior to clinical onset (visually identified by an epileptologist). In addition, approximately 30-60 s following ictal onset seizure activity also appeared to have spread to large areas of the brain (depending on the patient), resulting in non-measurable time delays between electrodes after that time. Thus, in this study we restricted the analysis of EEG signals to 60 s prior to 60 s following clinical seizure onset.

2.2. Principal component analysis

EEGs were decomposed into principal components, in order to isolate the signal contribution of the primary seizure propagation path, from which both scaling factors and time-delays were estimated. Principal component analysis (PCA) is a widely used signal decomposition method, which has been applied for extraction of source-related contributions to the EEG signal [10, 13], as well as artifact-related components [9], although in this study it was not applied for this latter purpose. It can, therefore, in principle be used for decoupling components associated with primary and secondary seizure propagation paths. Although there may not be a one-to-one mapping between the number of propagation paths that contribute to the EEGs and the number of principal components, we assumed that the first principal component corresponds to the contribution of the primary propagation path. PCA rather than independent component analysis (ICA) is preferred since the independence of components is not relevant in multi-path propagation. Paths are typically coupled giving rise to complex seizure propagation patterns.

Principal components were estimated from multi-channel EEG matrices. During focal seizure evolution, and particularly shortly before clinical onset, the primary signal contributions are expected to be seizure-induced rather than associated with other unrelated (possibly spontaneous) potentials in the brain. An example of the first three principal components of a (pre-ictal/ictal) EEG signal matrix is shown in Figure 1. Note that the second component includes both pre-ictal spiking activity and potential artifacts. Also, the EEG amplitude distribution is different in the first and third components, possibly corresponding to primary and secondary propagation, with distinct spreading patterns, possibly evolving prior to clinical onset. All further analysis was focused on the first principal component of the EEG.

2.3. The Mellin Transform

The Mellin transform has been used extensively in pattern recognition as well as in signal and image processing, e.g., [1, 2, 8, 14, 15], among many others. The Mellin transform $\mathcal{M}(\cdot)$ of real-valued function $g(t)$ is defined by:

$$\mathcal{M}\{g(t)\}(s) = \int_0^{\infty} t^{s-1} g(t) dt = \phi(s) \quad (1)$$

There is a direct relationship between the Mellin $\mathcal{M}(\cdot)$, Laplace $\mathcal{L}(\cdot)$ and Fourier $\mathcal{F}(\cdot)$ transforms. The Fourier transform of $g(t)$ is given by:

$$\mathcal{F}\{g(t)\}(s) = \int_{-\infty}^{\infty} g(t) e^{2\pi i s t} dt \quad (2)$$

If we define $\tilde{g}(t) = g(e^t)$, then Equation 2 becomes

$$\begin{aligned}
\mathcal{F}\{\tilde{g}(t)\}(s) &= \int_{-\infty}^{\infty} \tilde{g}(t) e^{2\pi i s t} dt = \\
&= \int_{-\infty}^{\infty} g(e^t) e^{2\pi i s t} dt = \\
&= \int_{-\infty}^{\infty} g(e^t) (e^t)^{2\pi i s} \frac{d(e^t)}{e^t} = \\
&= \int_0^{\infty} g(\tau) (\tau)^{\zeta-1} d\tau = \mathcal{M}\{g(\tau)\}(\zeta)
\end{aligned} \tag{3}$$

where $\zeta = 2\pi i s$ and $\tau = e^t$. The last expression in Equation 3 corresponds to the Mellin transform. The Fourier transform is invariant with respect to translation. The effect of time delay (translation) on the Fourier transform is a corresponding phase change in the transform, while its magnitude remains translation-invariant. Similarly, the Mellin transform is scale invariant, i.e., $\forall k$, where $k \in \mathbb{R}$, is any scale factor:

$$\mathcal{M}\{kg(t)\}(s) = k^{\Delta} \mathcal{M}\{g(t)\} \tag{4}$$

with Δ a scaling exponent. Thus, a scale change results in a phase change in the transform, while its magnitude remains scale-invariant.

Therefore, we assumed that observed EEGs $y_i(t)$ may be described as functions of the source signal $s(t)$ are follows:

$$y_i(t) = h_i s(t - k_i) + \varepsilon_i(t) \tag{5}$$

where h_i is the modulation of the source signal $s(t)$ in the direction of electrode i , k is the corresponding time delay and $\varepsilon_i = \mathcal{N}(0, \sigma_i^2)$ a noise term. Evidently, this is a simplified model of the EEG signal. Using a 5 s processing window, corresponding to a quasi-optimal window length for ictal EEGs estimated in a previous study [12], we calculated the Short-Time Fourier transforms (STFT) of the first principal component of the EEG matrices. We then applied the Mellin transform to the magnitude of the STFT, which is time-invariant.

3. RESULTS

We arbitrarily chose channel T3 as the reference in the estimation of relative scale shifts. Figure 2 shows an example of the T3 signal, the magnitude of its STFT and corresponding magnitude of the Mellin transform. As previously mentioned, these are respectively time- and scale-invariant.

For each patient and seizure, we examined the relative phase $\Delta\phi = \phi_{ref} - \phi_i$ of the Mellin transform of channel i , as scaled versions of the same signal are shifted by $e^{i\Delta\phi}$. Figure 3 shows an example of relative (with respect to T3) phase shifts between transforms across channels, at 4 time frames, 0-10 s (where $t=0$ marks the beginning of the selected EEG segment) to 60-70 s ($t=60$ s is the time of clinical onset), for one patient with bilateral temporal lobe seizures. Frequency-specific shifts, at various times within the selected intervals were estimated across patients. About 60 s prior to seizure onset scale shifts primarily at low frequencies <15 Hz were observed. About 30s prior to clinical onset, scale shifts at frequencies <40 Hz were estimated. The spatial variation of these shifts was approximately linear (as indicated from the peak-to-peak lines, in Figures 3 and 4). Note that these are scaling differences, not time-delays, and thus amplitude and polarity differences indicate directional modulation, in addition to small frequency shifts. Furthermore, after clinical onset differential shifts were estimated in different frequency ranges, including higher frequencies (60-90 Hz). This is expected, significantly higher frequency activity is measured by EEGs during the ictal interval (see the spectrum in Figure 2) that may be modulated by directional anisotropy.

Figure 4 shows a second example of relative phase shifts for a patient with seizures originating in the frontal lobes. Note that in addition to higher amplitude Mellin transform phase shifts, e.g., such as those at 10, 20, 25, and 60 Hz in the bottom right panel of this figure, there are several other frequency-locked shifts, with smaller amplitudes and variable polarity. Note that the polarity of the scale shift, is an indicator of relative expansion or compression of a signal in a particular direction. For example, in the preictal interval 20-30 s, and frequencies ~8 Hz, negligible scale shifts between T3 and FP2, P4, T1, T5, T4, O1, O2, but larger shifts between T3 and Fp1, F7, F3, P3, Pz were observed. At other frequencies, different spatial shift distributions were estimated. For example, at ~35 Hz, shifts of the same polarity were observed across all channels with slightly higher magnitude in channels of the left hemisphere. During ictal intervals, distinct spatial shift distributions, locked at different frequencies, were observed.

We examined the directional distribution of frequency-dependent scaling factors for all patients and seizures. Figure 5 shows an example of the radial distribution of estimated scale shifts for 4 seizures, at 3 frequencies, 10, 40 and 70 Hz, in the interval 60-70 s, i.e., in the first 10 seconds of the ictal interval. At 70 Hz, negative shifts (indicating signal compression) are observed bilaterally, below the T3-T4 axis. At 40 Hz higher shifts are observed in the right hemisphere, with significantly lower shifts at 10 Hz. Evidently these are based on rough estimates of the direction between electrodes, without using precise measurements of the electrode position and head dimensions. Scale shift distributions were consistent across seizures for individual patients. However, given the heterogeneous etiologies and associated potential structural brain abnormalities, estimated shifts were locked at different frequencies for different patients, within the same time interval.

In addition to providing information about the differential anisotropy of the brain, which may cause compression/expansion of the non-stationary EEG signal, estimated scale changes may also be useful in time-delay estimation, a problem of significant interest for source localization. Despite significant research, estimation of measurable time delays prior to or at ictal onset from non-invasive EEGs is a challenging problem. It is often thought that given the limitations of the scalp EEG, such estimation may not be possible. We have estimated measurable time delays at ictal onset from scalp EEGs and multiple seizures, using a signal decomposition method and a square law detector for relevant event detection [16]. Here, we compared these estimates to corresponding delays obtained following scale corrections, based on the the phase of the Mellin transform. Figure 6 shows an example of 4 seizures from two different patients. Time delays with (dashed line) and without (solid line) scale correction are superimposed.

In general, correction for scale modulation increased amplitude of estimated time delays, at least in some channels, but typically preserved the spatial distribution of these delays.

4. DISCUSSION

We have presented a novel approach based on the Mellin transform, for estimating scale-related modulations of EEG signals, potentially associated with the directional electrical/structural anisotropy of the brain. We have estimated measurable, frequency-locked phase differences in the Mellin space, corresponding to scale shifts in specific directions. Based on their polarity, we have identified directions of signal compression and expansion, both in preictal and ictal intervals (typically in the first ~10-20 s following clinical seizure onset). These shifts appeared to be patient-specific but not seizure-specific. This may be expected given that patients had distinct structural abnormalities, potentially resulting in similar *intra-patient* but distinct *inter-patient* anisotropy-induced scale modulations of EEGs. Furthermore, at a preliminary level, we examined the effects of scale correction on time-

delay estimation, for source localization purposes. It is often difficult to estimate measurable delays between scalp EEG signals, particularly for rapidly propagating seizures. Although, as desired, the spatial distribution of estimated time delays did not change significantly after scale corrections, their magnitude increased, possibly due to improved estimation from higher amplitude scale-corrected signals. Thus, correction for scale-related modulations may be a necessary pre-processing step for improved time delay estimation from scalp EEGs. Evidently this is a preliminary study, based on 5 patients and multiple seizures. A more detailed estimation of EEG scale shifts, based on specific measurements of electrode position and the geometry of the head is necessary. Finally, to establish a correlation between estimated scale factors and structural abnormalities, these results need to be further superimposed to structural imaging data (MRI).

Acknowledgments

This work was funded by a Harvard Catalyst pilot grant, and the Harvard Clinical and Translational Science Center (NIH grant 1UL1RR025758-2).

5. REFERENCES

- [1]. Baraniuk RG, Jones DL. Unitary equivalence: a new twist on signal processing. *IEEE Trans. Sig. Proc.* 1995; 43(10):2269–2282.
- [2]. Altes R. The Fourier-Mellin transform and mammalian hearing. *J. Acoust. Soc. Am.* 1978; 63(1): 174–183. [PubMed: 632408]
- [3]. Davies, G. *Integral transforms and their applications*. Springer-Verlag; New York, NY: 1984.
- [4]. Plummer C, Harvery AS, Cook M. EEG source localization in focal epilepsy: where are we now? *Epilepsia*. 2008; (2):201–208. [PubMed: 17941844]
- [5]. Ebersole JS, Hawes-Ebersole S. Clinical application of dipole models in the localization of epileptiform activity. *J. Clin. Neurophysiol.* 2007; 24:120–129. [PubMed: 17414967]
- [6]. Wolters CH, Anwander A, Tricoche X, Weinstein D, Kocj MA, MacLeod RS. Influence of tissue conductivity anisotropy on EEG/MEG field and return current computation in a realistic head model: a simulation and visualization study using high resolution finite element modeling. *Neuroimage*. 2006; 15(30):813–26. [PubMed: 16364662]
- [7]. Haueisen J, Tuch DS, Ramon C, Schimpf PH, Wedeen VJ, George JS, Belliveau JW. The influence of brain tissue anisotropy on human EEG and MEG. *Neuroimage*. 2002; 15:159–166. [PubMed: 11771984]
- [8]. Bertrand J, Bertrand P, Ovarlez JP. Discrete Mellin transform for signal analysis. *ICASSP-90*. 1990; 3:1603–1606.
- [9]. Shoker L, Sanei S, Chambers J. Artifact removal from electroencephalograms using a hybrid BSS-SVM algorithm. *IEEE Signal. Proc. Lett.* 2005; 12(10):721–724.
- [10]. Corsini J, Shoker L, Sanei S, Alarcon G. Epileptic seizure predictability from scalp EEG incorporating constrained blind source separation. *IEEE Trans. Biomed. Eng.* 2006; 53(5):790–799. [PubMed: 16686401]
- [11]. Stamoulis, C.; Chang, BS. Application of matched-filtering to extract EEG features and decouple signal contributions from multiple seizure foci in brain malformations; *IEEE Proc. 4th Int. Conf. Neural Eng.*; 2009; p. 514–517.
- [12]. Stamoulis C, Gruber LJ, Chang BS. Differentially modulated pre-ictal brain network dynamics at frequencies > 100 hz encoded in information theoretic measures. *Epilepsia*. submitted to. pending.
- [13]. Hess CW, James CJ. On semi-blind source separation using spatial constraints with application in EEG analysis. *IEEE Trans. Biomed. Eng.* 2006; 53(12 Pt1):2525–2534. [PubMed: 17153210]
- [14]. Zwicke PE, Kiss I. A new implementation of the Mellin transform and its application to radar classification of ships. *IEEE Trans. Pattern Anal. Machine Intell.* 1983; 5(2):191–199.

- [15]. Chen QS, Defrise M, Deconinck F. Symmetric phase-only matched filtering of Fourier-Mellin transforms, for image registration and recognition. *IEEE Trans. Pattern Anal. Machine Intell.* Dec; 1994 16(12):1156–1167.
- [16]. Stamoulis C, Schachter SS, Chang BS. Time delay estimation in scalp and intracranial EEGs during propagating seizures. pending.

Fig. 1. Signal decomposition by PCA. The vertical line indicates the clinical seizure onset.

Fig. 2. Time series (top plot), STFT (middle plot) and Mellin transform (bottom plot) for channel T3.

Fig. 3. Patient with temporal lobe seizures. Relative phase of Mellin transform across channels, using the transform phase of T3 as the reference. Each panel represents a different 10 s window from the start of the segment, at 10, 20, 40, and 60 s. Relative phase is shown for every channel as a function of frequency in the range 1-100 Hz.

Fig. 4.
Relative phase of Mellin transform across channels, for one patient with frontal lobe seizures.

Fig. 5. Radial distribution of scale shifts estimated from the Mellin transform, for 4 seizures. Radial distance corresponds to the amplitude of the shift and angle corresponds to the direction between T3 and all other electrodes.

Fig. 6. Estimated time delays prior to (solid line) and following (dashed line) scale correction, for 4 seizures from two different patients.