Sindh Univ. Res. Jour. (Sci. Ser.) Vol.44 (2) 215-220 (2012)

SINDH UNIVERSITY RESEARCH JOURNAL (SCIENCE SERIES)

Volume conduction effects on the coherence analysis of alcoholic's EEG

I.A. MEMON, M. S. KALHORO

Institute of Physics, University of Sindh, Jamshoro

*Corresponding Author: I.A. Memon, Email: memon904@btinternet.com Cell. No 03368510917

Received 2nd February 2012 and Revised 30th April 2012)

Abstract: An electroencephalography (EEG) coherence function is one of important signal processing too which is used to measure the linear correlation between different regions of the brain. Most of EEG coherence methods applied to examine alcoholic effects on the brain are based on difference of coherence function obtained by comparing the EEG activities of alcoholic and non-alcoholic subjects. However, volume conduction effects of various uncorrelated sources (VCUS) present in the brain introduce the biased estimates into its true value. This study has assessed the importance of minimizing VCUS effects on coherence analysis of alcoholic's EEG by comparing the coherence methods based on both scalp Laplacian (SL) and EEG. It is important, because most of the studies have interpreted the results of coherence analysis of alcoholic's EEG without keeping in account the VCUS effects. Results in this study have shown significant effects of VCUS on the statistics of EEG coherence analysis based on the coherence difference of alcoholic and non-alcoholic subjects.

Keywords: EEG, volume conduction effects, alcoholism, coherence function.

1. INTRODUCTION

Electroencephalography is the study of electric potentials from the brain and it is abbreviated as EEG. It is recorded by affixing electrodes on the human head. These signals provide important information for studying underlying brain processes. Various signal processing techniques are used for studying these signals for different purposes. Among these techniques, a coherence function is useful technique which is used to measure a linear correlation between signals. It is frequently used in EEG analysis to reveal functional relation between different brain areas. The coherence function between signals x (t) and y (t) is defined by the following relation (Nunez *et al.*, 1997):

$$C_{xy}(\omega) = \frac{P_{xy}(\omega)}{\sqrt{P_{xx}(\omega)P_{yy}(\omega)}} \quad (1)$$

where $P_{xy}(\omega)$ is the cross-spectrum normalized by the auto-spectra $P_{xx}(\omega)$ and $P_{yy}(\omega)$. The cross-spectrum of two time series $x(t)$ and $y(t)$ whose Fourier transforms are $X(t)$ and $Y(t)$ respectively is represented by the following relation:

$$P_{xy}(\omega) = X(\omega)Y^*(\omega) \quad (2)$$

where asterisk symbol shows the conjugate operation. When $x(t)$ is equal to $y(t)$ for all values of t, Equation (2) is called the auto-spectrum of signal $x(t)$ or $y(t)$. Auto and cross spectra are usually estimated on the basis of averages drawn from individual spectra of segments of corresponding signals. The square of absolute value of Eq. (1) is called the magnitude squared coherence (MSC). Coherence analysis of EEG signals has brought reliable results for a wide range of neurocognitive and clinical studies that corroborate the validity of this method. For example, it has been used to locate an epileptogenic focus from epilepsy patients (Mormann *et al.*, 2000), to identify neuroanatomic pathways for a seizure propagation (Sherman *et al.*, 1997), and to detect a seizure activity (Brazier, 1972; Gotman, 1983; Bahcivan *et al.*, (2001); Xiaoli Li *et al.*, (2006)).

Apart from the various statistical and recording problems, volume conduction from various uncorrelated sources (VCUS) present in the brain is the major problem for the estimation of robust coherence value. VCUS effects alter coherence estimates by introducing artificial coherence between a corresponding pair of electrodes (Srinivasan *et al.*, 1998; Nunez and Srinivasan, 2005).

Few signal processing techniques have been developed in order to address the issue of VCUS effects on EEG coherence analysis. These techniques use scalp Laplacian (SL) to estimate a coherence function rather than EEG signals. The SL, which is the useful approximation of scalp current density (SCD) is less affected by VCUS effects. The SCD is generated in the brain and reaches to the scalp by flowing through cerebral spinal fluid (CSF) and skull such that most of the current in the skull is radial which spreads tangentially when enters into the scalp (Nunez and Srinivasan, 2005). Perrin *et al.*, (1987) reported that potentials from deeper sources in the brain at a depth of half the radius of the head are not reflected in the SCD. The SCD is sensitive to superficial sources, with sensitivity falling off at approximately r^4, where r is the distance from the current source to the scalp surface (Oostendorp and Van Oosterom, 1996; Pernier *et al.*, 1988). The implication is that superficial cortical sources will have greater impact on the SCD than deeper sources. Due to this important characteristic, the SCD has an ability to eliminate much of the VCUS effects from distant sources, and electrical activity generated by reference electrodes (Hjorth, 1975).

Based on this important characteristic of the SL, some important results regarding VCUS effects on Fourier-based coherence of EEG signals were deduced by Nunez and Srinivasan (2005). According to these results, conventional EEG-based coherence due to VCUS effects does not depend on frequency and decreases as interelectrode distance decreases. The SL-based coherence function showed absence of VCUS effects as most of its part was independent of frequency. Various other studies have reported the ability of SL-based coherence in order to detect EEG correlation with minimum VCUS effects (Nunez and Pilgreen, 1991; Nelson and Nunez, 1993; Nunez and Westdorp, 1994; Srinivasan *et al.*, 1998; Srinivasan, 1999).

Coherence analysis for different neurological disorders including disorders caused due to alcoholism has avoided the issue of VCUS effects. This is due to the assumption that VCUS effects are additive and therefore can be ignored while comparing two different EEG activities for two or more groups of subjects (Sherman et al., 1997; Andrew et al., 1998; Winterer *et al.*, 2003; Nunez and Srinivasan, 2005; Marsdlek *et al.*, 2006; Rangaswamy and Porjesz, 2008). The major aim of this study is to examine the consequences of avoiding the issue of VCUS effects on the study based on comparing the coherence functions of alcoholic and non-alcoholic's EEG.

2. MATERIAL AND METHODS

EEG recording

Fifteen (15) male alcoholic subjects were recruited as paid volunteers from the program for alcoholism in the King's College London's School of Medicine. The alcoholic subjects were diagnosed according to diagnostic and statics manual of mental disorders (DSM-lV). The 15 male non-alcoholics subjects were recruited as paid volunteers through advertisements. The period of alcoholism was at least 12 years and drinking rate was 45 units per week. One unit of alcohol was equal to 10ml. The subjects in each group had no history of neurological disease and had normal hearing and normal sight. The subjects were in the age range of between 16 and 25 years. The subjects were seated on comfortable seat one meter away from the computer screen. EEG data in order to produce event related potentials (ERPs) was recorded from 64 electrodes based on the extended 10-20 EEG recording system of electrodes (Oostenveld and Praamstra, 2001). **(Table 1)** shows the position of the electrodes used in this study.

Table 1: The positions of the electrodes according to 64 electrode system based on the extended 10-20 system.

Alphabetical representation of electrodes	Numerical representation of electrodes	Location
F1, F2, F3, F4	7, 9, 8, 5, 6	Frontal Lobe
CPZ, PZ, P1, P2, P3, P4, CP3, CP4	61, 25, 60, 59, 23, 24, 48, 49	Parietal lobe
C5, FC6, FC5, CP5, CP6, P5, P6	42,10,11,19,20,50,51	Temporal lobe
POZ	57	Occipital Lobe

An analog to digital converter was used to change the EEG signals from a continuous analog wave form to a digital signal at the sampling rate of 256 Hz. Artifacts caused by the subjects were carefully minimized during the recording of EEG data. In order to minimize artefacts caused by the various body parts of the body, subjects were asked to remain in relax position and avoid unnecessary movement of their body and eyes. In addition to this, the bandpass filter between 0.02 and 50 Hz was also used for filtering out the artefacts caused by subjects such as electrogalvanic signals and movement artefacts as well as high-frequency artefacts such as electromyographic signals. Artefacts caused due to external sources such as the electrode-pop artifact, electro smog, etc were avoided by taking important measures during the recording. This included recording ERPs in a sound attenuated radio frequency shielded room blocking electromagnetic radiation from outside, the use of proper electrode/electrolyte combinations, and proper attachment of the leads.

ERPs were elicited by modified delayed matching-to-sample paradigm following the method used in the study of Zhang et al., (1995). Each subject was shown 18 pictures of objects as the stimuli chosen from the Snodgrass and Vanderwart (1980) picture set. Subjects were exposed to either a single stimulus (P1) or to two stimuli (P1 and P2) in succession. The pictures were shown to the subjects using three ways. The first condition was called single picture trial in which only single picture was shown to the subject. In second condition, which was called no-similarity trial, two picture stimuli, shown to the subjects, were completely different in terms of semantic category. The third condition was called similarity trial in which same picture stimulus P1 was repeated two times. The task of the subjects was to differentiate similarity and no-similarity trial by pressing the mouse button. The trial was called change trial if subject was successful in detecting no-similarity trial. The trial was called no-change trial if response of the subject was successful in detecting similarity trial. The trial was called failed trial if subject was not successful in detecting either no-similarity trial or similarity trial. The time duration between each trial was fixed to 3.2s and the interstimulus interval was 1.6s. This experiment yielded an ERP waveform consisting of three components which were most clearly discernible at the more posterior electrodes: component I (c110) ranging between 100 and 125 ms, component 2 (c175) ranging between 160 and 190 ms, and component 3 (c247) ranging between 220 and 260 ms.

Coherence analysis

First any two trials were randomly selected and transformed into the SL using the nearest neighborhood method as following (Hjorth, 1975):

$$l_x(n) = x_0(n) - \frac{x_1(n) + x_2(n) + x_3(n) + x_4(n)}{4} \qquad (3)$$

where $x_1(n), x_2(n),\ x_3(n)\ and\ x_4(n)$ are recorded time series of ERP signals at electrode's position 1, 2, 3, and 4 respectively and $x_0(n)$ is recorded time series of ERP signal at position 0 nearest and at equal distance to these electrodes. The transformed SL was then used for estimating the cross-spectrum and cross spectra using the equation (2). In order to avoid spectral leakage effects, the cross and auto-spectra were estimated after applying the Hanning window to each transformed SL.

This procedure was repeated for each pair of repeated trials. In next step, averages of all estimated cross and auto-spectra for each transform were estimated. Finally these averaged cross and auto-spectra were used in the square of the absolute value of Eq (1) in order to estimate magnitude squared coherence (MSC). The conventional EEG-based coherence function was estimated using the same procedure as described for SL-based coherence function.

The average EEG-based MSCs for 15 non-alcoholic subjects was compared to average EEG-based MSCs for 15 alcoholic subjects. The difference in statistical level of significance was examined using both the coherence methods.

Statistical Analysis

The group difference

The difference in SL-based and EEG-based MSCs between alcoholic and non-alcoholic subjects was assessed using the t-test. In order to access this difference using t-test, averages of both SL-based and EEG-based MSC values across the alpha and beta frequency bands were estimated and then these average MSCs were normalized using the Fisher's Z transformation.

Confidential interval of MSC

An important method to control statistical errors in coherence estimate is based on the confidential interval of a coherence function. It is the probability that the true coherence at a certain time lies within a certain interval about the estimated coherence. The approximate relation for the 95 % confidence interval in terms of standard error e for Fourier coherence was estimated using the following relation (Bendat and Piersol, 2001):

$$\frac{C_{xy}^2(\omega)}{1+2e} \le coh^2 \le \frac{C_{xy}^2(\omega)}{1-2e} \qquad (4)$$

where

$$e = \sqrt{\frac{2}{N_c}}\,\frac{1 - C_{xy}^2(\omega)}{C_{xy}(\omega)}$$

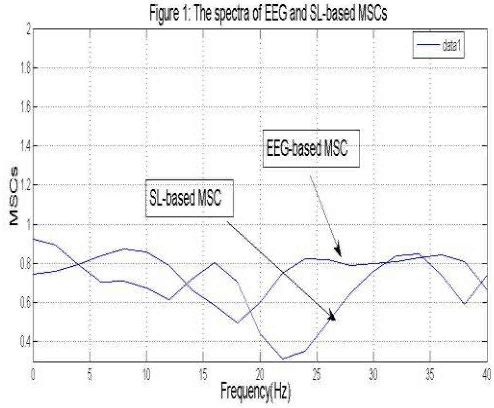
N_c is the number of samples in the interval over which the coherence function is estimated and coh^2 is true coherence. Any MSC value outside this interval was rejected. In order to apply student's t-test, the normal distribution of MSC values were obtained using the Fisher's z-transform.

3. RESULTS AND DISCUSSION

Fig. 1 clearly indicates that EEG-based MSC is independent of frequency around 25Hz to 37Hz as compared to the SL-based MSC.

It was found that EEG-based MSC due to VCUS effects does not always decrease as interelectrode distance increases. However decrease in coherence was always observed when interelectrode distance was along some fixed direction. It is important to mention here that existing studies on EEG coherence analysis also predict decrease in coherence due to VCUS as interelectrode distance increases (Nunez, 1981; Srinivasan *et al.,* 1996; Nunez *et al.,* 1997; Srinivasan et al., 1998; Nunez and Srinivasan, 2005). However, these studies do not mention effects of the direction between corresponding electrodes. Most of results of these studies have been derived using EEG signals of electrodes attached to the scalp of a subject along some fixed direction.

Figure 1: The spectra of EEG and SL-based MSCs



Coherence methods based on both SL and EEG were used in the study of examining MSC difference between alcoholic and non-alcoholic subjects. The average of MSC across the alpha and beta frequency band (henceforth called the alpha and beta MSCs) was used in order to assess the difference in MSCs between alcoholic and non-alcoholic subjects using the t-test.

The results for EEG-based MSC difference between alcoholic and non-alcoholic subjects are shown in Table 2. As shown in this table that the alpha and beta MSCs of EEG signals are larger in value for alcoholic subjects as compared to those estimated for non-alcoholic subjects. These larger MSCs were mostly observed for parietal and frontal, regions of the brain. Results were also consistent with studies on EEG-based coherence analysis for alcoholics (Moselhy *et al.,* 2001; Winterer *et al.,* 2003; Marsdlek *et al.,* 2006). **(Table 2)** shows that t-test reveals significant difference in alpha and beta MSCs of EEG between alcoholic and non-alcoholic subjects.

The results for SL-based MSC difference between alcoholic and non-alcoholic subjects are shown in Table 3. Evan though SL-based alpha and beta MSCs for alcoholic subjects appear to be larger in value than those estimated for non-alcoholic subjects but t-test reveals larger values of P as compared to those based on EEF-based MSC method. Because of these larger values of P, the difference in SL-based MSCs between alcoholic and non-alcoholic subjects is not statistically significant as it has been examined for EEG-based MSC difference. These results show a considerable loss in significant level at various positions of electrodes.

The larger values of EEG-based MSCs do not necessarily include true coherence. This study while analyzing EEG-based MSCs have examined various MSCs affected by VCUS effects. Contrary to this, SL-based MSCs were not found affected by the VCUS effects which is because of the characteristics of SL that filter out VCUS effects. Therefore difference in statistical significances between EEG-based and SL-based methods of MSC provides an evidence that VCUS effects can biase the results of statistical analysis based on examining MSC coherence difference between two different EEG activities.

Table 2: The EEG-based alpha and beta MSCs (means+standard deviations) for alcoholics and non- alcoholics subjects and scores of t-test for various positions of the electrodes

Frequency Band	EEG-based MSC(Mean+SD) Non-Alcoholics	EEG-based MSC(Mean + SD) Alcoholics	Electrode Positions	T-test for EEG P	T
Alpha Beta	1.20±0.11 1.02±0.09	1.26±0.11 1.08±0.07	F1-F3	0.025 0.002	2.281 3.113
Alpha Beta	1.02±0.06 1.15±0.10	1.05±0.08 1.22±0.10	F2-F4	0.080 0.004	1.774 2.928
Alpha Beta	1.06±0.10	1.10±0.07	F3-F5	0.056 0.031	1.938 2.198

	1.16± 0.09	1.21±0.10		
Alpha Beta	1.2±10.12 1.2±50.12	1.27±0.14 1.31±0.13	F4-F6	0,058 1.925 0,048 2.006
Alpha Beta	1.19±0.11 1.04±0.09	1.25±0.11 1.130±.14	PZ-P1	0.025 2.281 0.002 3.199
Alpha Beta	1.20±0.11 1.04±0.06	1.27±0.10 1.06±0.08	PZ-P2	0.006 2.785 0.240 1.183

Table 3: The SL-based alpha and beta MSCs (means+standard deviations) for alcoholics and non- alcoholics subjects and scores of t-test for various positions of the electrodes

Frequency Band	SL-based MSC(Mean+SD) Non-Alcoholics	SL-based MSC(Mean+SD) Alcoholics	Electrode Positions	T-test for EEG	
				P	T
Alpha Beta	1.17±0.11 1.33±0.15	1.21±0.10 1.36± 0.09	F1-F3	0.116 0.313	1.591 1.014
Alpha Beta	1.06±0.07 1.16±0.11	1.08±0.08 1.12±0.10	F2-F4	0.269 0.116	1.113 1.598
Alpha Beta	1.19±0.10 1.20±0.11	1.23 ±0.09 1.27±0.10	F3-F5	0.083 0.006	1.759 2.785
Alpha Beta	1.37±0.13 1.18±0.11	1.43± 0.11 1.2±50.11	F4-F6	0.040 0.009	1.737 2.662
Alpha Beta	1.70± 0.11 1.13±0.14	1.23±0.07 1.18±0.14	PZ-P1	0.177 0.139	1.361 1,494
Alpha Beta	1.26±0.11 1.17±0.14	1.29± 0.08 1.23±0.14	PZ-P2	0.196 0.077	1.304 1.793

4. CONCLUSION

Most of studies avoid the issue of VCUS effects while studying the difference in coherence function estimated for various neurological and cognitive disorders. However, results in this study showed significant effects of VCUS on statistics of EEG coherence methods based on difference in coherences between alcoholics of non-alcoholic's EEG activities. These results provided substantial evidence that VCUS effects are not additive and therefore cannot be ignored in the study of examining coherence difference between different brain states.

REFERENCERS:

Andrew, C.N.C., P. Rappelsberger, O. FILZ (1998) Topology of EEG Coherence Changes May reflect differential Neural Network Activation in Cold and Pain Perception. Brain Topography, (**11**): 125-135.

Brazier, M.A.B (1972) Spread of seizure discharges in epilepsy: anatomical and electrophysiological consideration, (**36**): 263-264.

Bahcivan, H., Ning Zhang, M.A. Mirski and D. Sherman (2001) Cross-correlation analysis of epileptiform propagation using wavelets (2001) Conference Proceedings of the 23rd Annual International Conference of the IEEE Engineering and Medicine and Biology Society, **(1):** 1804-1807.

Bendat, J.S. and A. G. Piersol (2001) Random data analysis and measurement procedures, 3rd edition, Wiley Press, New York, U.S.A.

Hjorth, B., (1975) An on-online transformation of EEG scalp potential into orthogonal source derivation. Electroencephalography and Clinical Neurophysiology, (**39**): 526-530.

Gotman, J (1983) Measurement of small time differences between EEG channels: method and application to epileptic seizure propagation. Electroencephalography and clinical neurophysiology, 56 **(5):** 501-514.

Gevins, A.S., L. E. Brickett, P. Reutter and J. Desmond (1991) Seeing through the skull: advanced EEGs use MIRs to accurately measure cortical activity from the scalp. Brain Topography, 4**(2):** 125-131.

Mormann, F., K. Lehnertz, P. David and C.E. Elger (2000) Mean phase coherence as a measure for phase synchronization and its application to the EEG of epilepsy patients. Physica D, 144**(3-4):** 358-369.

Moselhy, H.F., G. Georgiou and A. Kahn (2001) Frontal lobe changes in alcoholism: A review of the literature. Alcohol and Alcoholism, (**36**): 357-368

Marsdlek, T., V. Matousek, P. Mautner, M. Merta and R. Moucek (2006) Coherence of EEG signals and biometric signals of handwriting under influence of nicotine, alcohol and light drugs. Neural Network World, 16**(1):** 41-60.

Nunez, P (1981) Electric fields of the brain, 1st edition, Oxford Press, New York, U.S.A.

Nunez, P.L and K.L. Pilgreen (1991) The spline-Laplacian in clinical neurophysiology: a method to improve EEG spatial resolution. Journal of Clinical Neurophysiology, (**2**): 397-413.

Nelson, A.V and P.L. Nunez (1993) Comparison of the Laplacian with classical recording methods in coherence analysis of EEG. Proceedings of the 15th Annual International Conference of the IEEE

Engineering in Medicine and Biology Society. Published by IEEE, Piscataway, NJ, USA 450-451.

Nunez, P. L and A. F. Westdorp (1994) The Surface Laplacian, High Resolution EEG and Controversies. Brain Topography, 6 **(3):** 221-226.

Nunez, P.L., R. Srinivasan, A.F. Westdorp, R.S. Wijesinghe, D.M. Tucker, R.B. Silberstein and P.J. Cadusch (1997) EEG coherency. I. Statistics, reference electrode, volume conduction, Laplacians, cortical imaging and interpretation at multiple scales. Electroencephalography and clinical neurophysiology, 103 **(5):** 499-515.

Nunez, P. L and R. Srinivasan (2005) Electric fields of the brain. Oxford university press.

Oostendorp, T.F and Van Oosterom, A (1996) Surface Laplacian of the potential: theory and application. IEEE Transactions on Biomedical Engineering, 43 **(4):** 394-405.

Perrin, F., O. Bertrand and J. Pernier (1987) Scalp current density mapping: value and estimation from potential data. IEEE Transactions on Biomedical Engineering, 34 **(4):** 283-288.

Pernier, J., F. Perrin and O. Bertrand (1988) Scalp current density fields: Concepts and properties. Electroencephalography and Clinical Neurophysiology, (**69)**: 385-389.

Rangaswamy, M., B. Porjesz, D.B. Chorlian, K. Wang, K. A. Jones, L.O. Bauer, J. Rohrbaugh, S.J. O'connor, S. Kuperman, T. Reich and H. Begleiter (2002) Beta power in the EEG of alcoholics. Biological Psychiatry, 52 **(8):** 831-842.

Snodgrss, J.G and M. Vanderwart (1980) A standardized set of 260 pictures: norms for the naming agreement, familiarity, and visual complexity. Journal of Experimental Psychology: Human Learning and Memory, (**6)**: 174-215.

Srinivasan, R., P.L. Nunez, D.M. Tucker, R.B. Silberstein and P.J. Cadusch (1996) Spatial sampling and filtering of EEG with spline Laplacians to estimate cortical potentials. Brain Topography, (**8)**: 355-366.

Sherman, D.L., Y.C. Tsai, L.A. Rossell, M.A. Mirski and N.V. Thakor (1997) Spectral analysis of a thalamus-to-cortex seizure pathway. IEEE Transactions on Biomedical Engineering, 44 **(8):** 657-664.

Srinivasan, R., P.L. Nunez, and R.B. Silberstein (1998) Spatial filtering and neocortical dynamics: estimates of EEG coherence. IEEE Transactions on Biomedical Engineering, 45 **(7):** 814-826.

Srinivasan, R (1999) Methods to improve the Spatial Resolution of EEG. International Journal of bioelectromagnetism, 1**(1):** 102-111

Winterer, G., M.A. Enoch, K.V. White, M. Saylan, R. Coppola and D. Goldman (2003) EEG phenotype in alcoholism: increased coherence in the depressive subtype. Acta Psychiatrica Scandinavica, 108 **(1):** 51-60.

Xiaoli l., Y. Xin., J.R.G. Jefferys, and J. Fox (2006) Computational neuronal oscillations using Morlet wavelet transform (2005). 27th Annual International Conference of the IEEE Engineering in Medicine and Biology Society, 31 Aug.-3 Sept, 4-7.

Zhang, X.L., H. Begleiter., B. Porjesz W. Wang., and A. Litke (1995) Event related potentials during object recognition tasks. Brain Research Bulletin **(38):** 531-538.