PROCESSING OF EEG SIGNALS FOR REDUCING THE EFFECTS OF ADJACENT ACTIVITIES

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ABSTRACT

The primary objective of recording EEG signal at a particular location on the scalp is to know the electrical activity of the brain right below the electrode position. However, signals of cerebral origin from areas beneath adjacent electrodes also gets superimposed on the original signal of interest. In this paper, we are proposing a least squares technique for estimating the effect of adjacent activities and subsequently subtracting it from the recorded EEG so as to get a better estimate of the activity beneath the electrode of interest. The results show that the processed signal better reflects the activities below the electrode location than the one originally recorded.

INTRODUCTION

The synchronous discharge of individual neurons of the brain generates rhythmical electrical potentials, which appear as a waveform on the surface of the scalp and is called the electroencephalogram (EEG). These signals are picked up by placing electrodes on the intact skull. The signal contains a number of components. The most obvious activity is a rhythmic activity called the alpha activity and is by definition within the frequency range of 8-12Hz. The rhythmic activity in the higher frequency range of 14-30Hz. is called beta activity and its spatial distribution is different from that of alpha activity. Lower frequency activities are called delta activity in the frequency range 0.5-3Hz. and theta activity in the frequency range 4-7Hz. In addition to these activities, the EEG also contains transients like spikes, spindles etc. [1].

The nature of the brain electrical activities vary over different parts of the scalp and is dependent upon various factors such as age and mental state of the subject, location of the electrodes on the scalp, external stimuli, injuries to the brain, functional disturbances and diseases of the brain [2,3]. Hence EEG has become a research tool in neurophysiology, psychophysiology and psychopharmacology and its use is well established in monitoring sleep, surgical operations which influence the brain functions [4], depth of anesthesia and cessation of brain functions. In all these applications, one is interested to know from the recorded EEG signals, the activity of the brain right below the electrode position on the scalp. However signals of cerebral origin from areas beneath the adjacent electrodes also gets superimposed on the original signal of interest. The signals from other locations may be considered as artefacts (noise) and it is desirable to reduce its effects so as to get the real EEG signal due to the area of the brain below the electrode. The EEG signals are recorded from different locations on the scalp for a duration of 20 to 30 minutes. The 10-20 electrode placement system of Jasper adopted by the International Federation of Soceities for Electroencephalography and Clinical Neurophysiology is shown in Fig.1 [5]. The

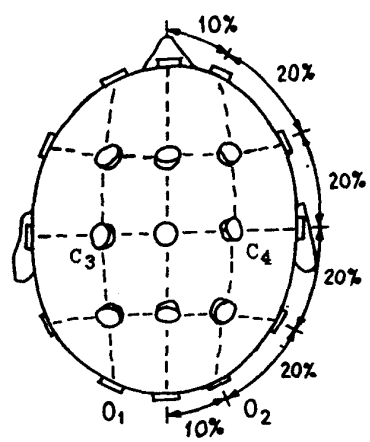

Fig. 1 Placement of the 21 electrodes used in the International 10-20 system.

recording can be done either using bipolar or

monopolar (unipolar) techniques. In the bipolar technique, the difference in potential between two adjacent electrodes is measured, whereas in the monopolar technique the potential of each electrode is measured with reference to an indifferent electrode on the chin, ear or back of the neck. The indifferent electrode can also be created by connecting all the other electrodes through equal resistances. The visual interpretation of such long records are highly subjective and time consuming for a clinician. For this reason, computers are being increasingly used for extracting useful clinical features [6]-[11].

In this paper, we are proposing a least squares technique [12] for estimating the effect of adjacent activities and subsequently subtracting it from the EEG record so as to get a better estimate of the activity beneath the electrode of interest. EEG signals from adjacent electrodes are used to determine the ratios in which these signals have contributed to the artefact in the corresponding EEG signal. The criterion for determining these ratios is to minimize a least squares performance measure defined on the artefact subtracted EEG signal. These ratios are calculated for every EEG derivation and the artefacts are subtracted accordingly. It is assumed that the net effect of the adjacent channels at the recording location of interest may be expressed as a weighted linear combination of EEGs from adjacent locations. A comparison of time plots as also the continuity index (CI) [13] shows that the processed signal better reflects the activities below the electrode locations than the one originally recorded.

ESTIMATION TECHNIQUE

The procedure for estimating the EEG activity beneath the electrode of interest, making use of the activities recorded from the adjacent electrodes along with the activity at the electrode of interest, is as follows.

This method is based on the principle that the EEG activities due to adjacent electrodes are additive to the EEG activity of interest, and is linearly related to the adjacent activities.

$$EEG_m(i) = EEG_t(i) + \sum_{\substack{j=1 \\ j\neq m}}^{P} \theta_j EEG_j(i) + \text{noise}$$

$$= \sum_{\substack{j=1 \\ j\neq m}}^{P} \theta_j EEG_j(i) + e(i) \quad ; i= 1,..M \quad ...(1)$$

where $EEG_m(i)$ is the measured EEG and $EEG_t(i)$ is the true EEG at the point of interest, θ is a vector of correlation coefficients or transmission coefficients representing the fraction of the adjacent activities measured at the point of interest along with the true EEG. The noise represents the measurement noise whose effects are assumed to be negligible. The error term e(i) represents the combined effects of $EEG_t(i)$ and other artefacts. M is the total length of the data available and P is the number of channels of EEG recorded and available for the processing.

In this work, an estimate $\hat{\theta}$ of θ is obtained. An estimate of the true EEG is then calculated by subtracting the estimated artefact, due to the adjacent activities, from the measured EEG.

Ordinary Least Squares method

The model may be conveniently written as in Eqn.2 below

$$x_m(i) = \theta_1 x_1(i)+...+\theta_{m-1}x_{m-1}(i) +\theta_{m+1}x_{m+1}(i)+...+\theta_P x_P(i)+e(i)$$

$$= x^{m^T}(i)\theta^m + e(i) \quad ;i= 1,...M \quad ...(2)$$

where

$$x^m(i) = [x_1(i)...x_{m-1}(i),x_{m+1}(i).. .. x_P(i)]^T \quad ...(3)$$

and

$$\theta^m = [\theta_1,...\theta_{m-1},\theta_{m+1},...\theta_P]^T \quad ...(4)$$

are (P-1) dimensional vectors.

$x_m(i)$ is the ith sample of the raw EEG at m^{th} electrode, $x^m(i)$ is a vector of the ith samples of the adjacent (P-1) EEGs used in the model. The T indicates transposition. Here the problem is to obtain an estimate of θ^m given $x^m(i)$ and $x_m(i)$, i=1,2,... M, and then to use this estimate to obtain an estimate of the EEG at an electrode of interest.

Estimate of θ^m can be obtained, using the ordinary least squares method, by minimizing the sum of squares of the residuals which is the estimate of the true EEG.

To estimate the value of θ^m, say $\hat{\theta}^m$, Eqn.2 can be written more compactly in matrix form as

$$X_m = X^m \theta^m + E \qquad ...(5)$$

where

$$X_m = [x_m(1), x_m(2), ... x_m(M)]^T \qquad ...(6)$$

$$\theta^m = [\theta_1, \theta_2, ... \theta_{m-1}, \theta_{m+1}, ... \theta_P]^T \qquad ...(7)$$

$$E = [e(1), e(2), ... e(M)]^T \qquad ...(8)$$

are vectors of dimensions M, (P-1) and M respectively, and

$$X^m = [x^m(1), x^m(2), ... x^m(M)]^T \qquad ...(9)$$

is an M x (P-1) matrix.

Ordinary least squares estimate of θ^m is obtained by minimizing the sum of the squares of the residuals.

$$\text{i.e., } \min J = \sum_{i=1}^{M} \hat{e}^2(i)$$

$$= \hat{E}^T \hat{E}$$

$$= (X_m - X^m \hat{\theta}^m)^T (X_m - X^m \hat{\theta}^m) ...(10)$$

Differentiating J with respect to $\hat{\theta}^m{}_j$ and simplifying we get

$$\hat{\theta}^m = (X^{m^T} X^m)^{-1} X^{m^T} X_m \qquad ...(11)$$

where $\hat{\theta}^m$ is the vector of parameter estimates, X^m is the EEG (adjacent) matrix and X_m is a vector of EEG samples. Having obtained $\hat{\theta}^m$, the EEG activity at the electrode of interest can be obtained as

$$\hat{e}(i) = x_m(i) - x^{m^T}(i) \hat{\theta}^m$$

$$i=1, ... M \qquad ...(12)$$

The implicit assumption in the ordinary least squares method is that the EEGs due to adjacent activities are not perfectly collinear. If they are collinear, the inverse of the matrix $(X^{m^T} X^m)$ will not exist and hence $\hat{\theta}^m$ cannot be obtained.

RESULTS AND DISCUSSION

Monopolar recording of the normal EEG is made at four locations C_3, C_4, O_1 and O_2 on the scalp. These signals are digitized at 256Hz. sampling rate and low pass filtered at 32Hz. using a linear phase finite impulse response digital filter. By considering O_2 as the desired location and others (viz C_3, C_4 and O_1) as adjacent locations (the activities due to these acts as the artefact at O_2), the least squares technique is applied. The effect of adjacent EEG activities on the location O_2 is considered as a weighted linear combination of the signal values at C_3, C_4 and O_1. The weights are calculated by the minimization procedure and then the estimate of the signal contributions from C_3, C_4 and O_1 at location O_2 is subtracted from the signal recorded at O_2 so as to get the true signal at O_2. This procedure is applied to other locations, viz C_3, C_4, O_1 and considering the effect of activities from remaining locations as artefacts, to get a better estimate of the activities at the electrodes of interest.

Fig 2(a), (b), (c) and (d) shows the recorded electrical activities at locations O_1, O_2, C_3 and C_4 respectively on the scalp. Fig 3a is the least squares estimate of the electrical activity at O_1 taking the activity at O_2, C_3 and C_4 as artefacts. It may be observed that it reflects a better estimate of the actual activity at O_1 on the brain. Fig 3b is the estimate of the EEG activity at O_2 position on the scalp keeping the other activities as artefacts which are linearly related to the activity at O_2 position. This also shows a change in the pattern as compared with the original recorded signal indicating a better estimation. Fig 3c is the EEG estimated at C_3 position keeping the other activities as artefacts. The high amplitude low frequency activity is being reduced from the original recording at C_3 position which shows a better activity at C_3 position. Fig 3d is the illustration of the least squares estimate of the electrical activity at C_4 electrode on the scalp by taking the EEG activities at other electrodes as artefacts.

As a measure of comparison between the original and processed signals at different electrodes, we have calculated the continuity

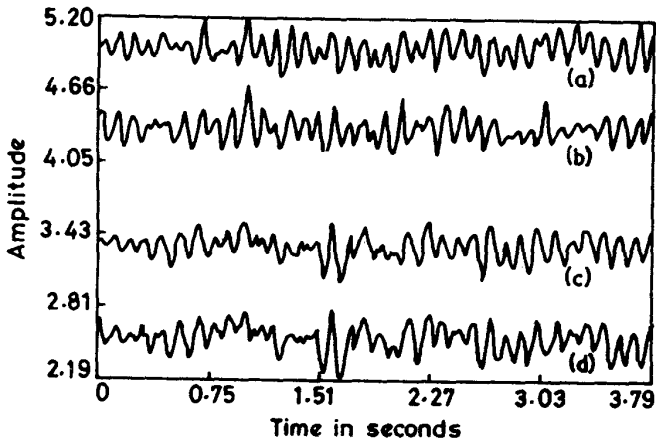

Fig. 2 Unprocessed signals in time domain

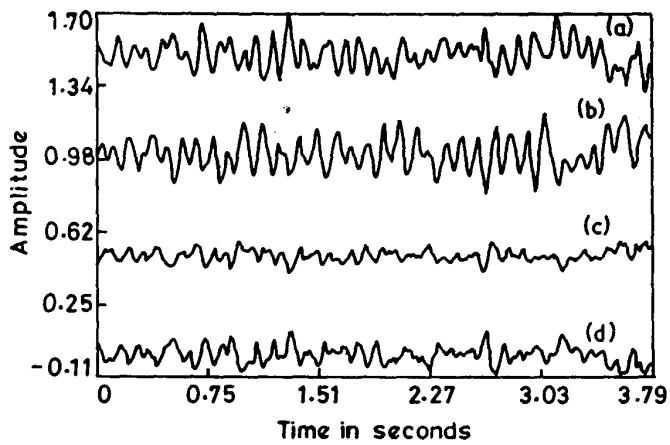

Fig. 3 Processed signals in time domain

Table 1. The continuity index (CI) of signals before and after processing.

Electrode Position	CI of original signals	CI of estimated signals
O_1	0.33333333334	0.444444444445
O_2	0.33333333334	0.555555555558
C_1	0.44444444445	0.444444444445
C_2	0.33333333334	0.555555555558

index (CI)(see appendix) which is a measure of continuity of the signals. The table shows the CIs of these signals before and after processing. The high value of the CIs for the processed signals reflects a better estimate of the EEG activities at the electrodes of interest.

In conclusion, we have proposed a least squares technique for estimating the effect of adjacent activities and subsequently subtracting it from the EEG record so as to get a better estimate of the activity beneath the electrode of interest. The technique may be applied to EEG signal at any location on the scalp provided simultaneously recorded adjacent signals are available, which is normally the case with routine EEG recording.

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APPENDIX

The continuity index (CI) of the signals can be calculated as follows :

Divide the total length of the data into segments of equal data length (say N). If M is the total length of the data, then the number of segments is given by,

$$R = \frac{M}{N}$$

The CI is calculated using the formula given by

$$CI = \frac{\text{Number of segments having average value} \geq a_T}{\text{Total number of segments}}$$

where ,

a_T is the average value of the original data of length M.