Sleep Onset Estimator: Evaluation of Parameters

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Abstract – **The electroencephalographic (EEG) alterations during the human sleep onset (falling asleep period) has been evaluated by several studies in the past. However, the analysis part has been limited due to standard signal processing methods. This paper has attempted to evaluate a number of advanced parameters for improved sleep onset estimation, such as EEG non-parametric coherence, power frequency and spectral band power. These parameters can be utilised in an on-line algorithm design for neurofeedback applications.**

I. INTRODUCTION

Falling asleep is a complex process. A common sleep disorder, insomnia, is often associated with disturbance of this complex process. There are electrophysiological, cognitive and behavioural changes that occur during this transition period. This paper will discuss the electrophysiological changes in the electreoencephalogram (EEG).

What is the moment of sleep onset? How can sleep onset be characterised from the sleep electroencephalographic (EEG) activity in human adults? These are some of the questions that have no clear and definite answer, except that some sleep researchers Dement and Kleitman [1] and Rechtschaffen and Kales [2] claim that sleep onset in stage 1 is based on criteria of alpha reduction. The sleep could be said to begin with the arrival of spindles. One report stated that stages 2, 3, 4 and rapid eye movement (REM) compose the 'true sleep'. Whereas, the 'true wake-to-sleep transition' is measured from stages 1 to 2 [3]. The study of the waking and sleeping transitions using EEG spectral analysis was mainly indicated by the delta and sigma frequencies [4]. Another study reported an increased variability in delta, theta and alpha EEG bands at the beginning of stage 1 sleep and continued 10 minutes into sleep [5]. Ogilvie's study reported theta power increase as the alpha and beta decreased during the sleep onset period and slow wave sleep [6]. Badia et al., investigated the 3 min EEG data sample of transition from wakefulness to stage 1 (1.5 min of wakefulness and 1.5 min of stage 1), 5 second epochs and single Hz bands [7]. The results revealed that the largest decrease in relative spectral power was between 9-11Hz at occipital region and the greatest increase at 3 and 4Hz, in the frontal and central sites. The changes in these bands were described as fluctuating, especially in the delta frequency range.

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An EEG power frequency variable can be derived from the spectral analysis of the EEG signal, such as spectral edge frequencies (SEF), for example SEF 95, SEF 90 and SEF 50 (median frequency). These spectral edge frequencies are referred as frequencies below which 95, 90 and 50% of the total EEG power is located [8]. For the EEG coherence analysis of sleep onset period, the investigation of spatiotemporal dynamics is also important. Earlier studies reported that delta band coherence between occipital and frontal, and central and parietal regions dropped sharply just before alpha disappeared. This finding was interpreted as the '*hypnagogic*' state which begins just before alpha vanishes [9]. Generally, the anterior synchronous activity, quantified by coherence, is related to sleep EEG while posterior region reflects waking EEG. The focus of this paper has been to objectively measure the wake-to-sleep transition, or drowsiness period from the sleep EEG activity. We present a new method for estimation of wake-sleep (sleep onset) depth, consisting of coherence, power frequency and band power estimations.

II. MATERIALS AND METHODS

A. Subjects

One human subject (age: 32, sex: male, weight: 96kg, height: 1.76m) was recruited for an overnight Polysomnographic (PSG) recording at St. Lukes Hospital (Sydney, NSW, Australia). The subject was diagnosed with the Apnoea Hypopnoea Index (AHI) of 5.1, representing a mild case of hypopnoea.

B. Experimental Protocol

The sleep PSG was recorded from 22:30 h until 05:00 the next day using Bio-Logic System and Adult Sleepcan Vision Analysis (Bio-Logic Corp., USA). The PSG data was recorded at 256 Hz sampling rate. Surface electrodes were placed on the skull (C3, C4 and O2; 10-20 system) and referenced to bridged left and right mastoid to record the EEG activity. Two channels were used to record the eye movements, with one electrode placed 1 cm above and slightly lateral to the outer canthus of one eye and the second electrode recording the potentials from an electrode 1 cm below and slightly lateral to the outer canthus of the other eye. Both electrodes were referenced to left-right mastoid. The other electrodes recorded the EMG from the muscle areas on and beneath the chin, ECG (using lead-II across the chest area), nasal and oral airflow, snoring sounds, breathing effort (measured at the chest and abdomen), oxymetry, actigraphy recording body positioning and leg movements (right and left anterior tibialis).

The recorded data was visually scored by the sleep technician according to Rechtschaffen and Kales [2] from 30

second epochs. The analysis reported 64.5% sleep efficiency and 91.8% in sleep maintenance with 132 min spent in wake (W) stage, 30 min in stage 1 (S1), 125 min in stage 2 (S2), 17 min in stage 3 (S3), 49 min in stage 4 (S4), 26 min in stage REM, 220 min non-REM and 3 min in movement time.

III. SIGNAL PROCESSING

The EEG and EOG data was processed and analysed using Matlab software (Mathworks, USA). The EEG-EOG correction was applied based on the regression analysis, together with source noise removal with the 50 Hz notch filter. The estimation of sleep onset transition was computed in EEG signals by applying various coherence, power frequency and band power methods.

A. EEG Inter-Hemispheric Coherence

The EEG inter-hemispheric coherence was computed to investigate the functional connectivity between the left and right brain hemispheres. There was 1 coherence parameter computed as a result of 2 EEG recorded channels (C3-C4). Coherence estimates the linear cross-correlation between two signals as a function of frequency. EEG coherence also estimates the degree of synchrony between the electrical activities of the two brain regions, concentrating on a certain frequency or EEG band.

$$\kappa^2{}_{xy}(f) = \frac{|S_{xy}(f)|^2}{|S_{xx}(f)||S_{yy}(f)|} \tag{1}$$

where $\kappa^2{}_{xy}$ is the estimated coherence range between 0 and 1.

In equation (1), S denotes the spectral estimate of two EEG signals x and y for a given frequency (f) bin. The numerator contains the cross-spectrum for x and y (S_{xy}), while the denominator contains the respective auto-spectra for x (S_{xx}) and y (S_{yy}). For each frequency (f) bin, the coherence value is obtained by squaring the magnitude of $\kappa_{xy}(f)$. The $\kappa^2{}_{xy}(f)=0$ indicates that the activities of the signals in this frequency are linearly independent, whereas a value of $\kappa^2{}_{xy}(f)=1$ gives the maximum linear correlation for this frequency.

A well known non-parametric spectral estimation algorithm is the Capon's approach, known as Minimum Variance Distortion-less Response (MVDR) [10]. Benesty has proposed a method to estimate the magnitude squared coherence function [11]. The MVDR spectrum is often considered as an output of a bank of filters with each filter centered at one of the analysis frequencies. The MVDR bandpass filters are both data and frequency dependent in comparison to parametric periodogram approach, which is both data and frequency independent.

In the proposed generalized MVDR method, the filter coefficients are chosen so as to minimize the variance of the filter output, subject to constraint:

$$g_k^H u_k = u_k^H g_k = 1 \tag{2}$$

where the unitary matrix is

$U = [\, u_0 \quad u_1 \quad \cdots \quad u_{K-1} \,]$ with $U^H U = U U^H$

and K filters of length K are

$g_k = [g_{k,0} \; g_{k,1} \; \cdots \; g_{k,K-1}]^T$, $k=0, 1, ..., K-1$.

The magnitude squared coherence function is defined between two signals $x(n)$ and $y(n)$ as

$$\gamma^2{}_{xy}(u_k) = \frac{|S_{xy}(u_k)|^2}{|S_{xx}(u_k)||S_{yy}(u_k)|} \tag{3}$$

By conducting extended mathematical calculation [10], the magnitude squared coherence becomes:

$$\gamma^2{}_{xy}(u_k) = \frac{|u^H{}_k R^{-1}{}_{xx} R_{xy} R^{-1}{}_{yy} u_k|^2}{[u^H{}_k R^{-1}{}_{xx} u_k][u^H{}_k R^{-1}{}_{yy} u_k]} \tag{4}$$

where (u_k) is a unitary matrix; and $0 \le \gamma^2{}_{xy}(u_k) \le 1$.

B. EEG Frequency Indicator "Brain-Rate"

The weighted mean frequency of the EEG spectrum has been defined as the 'brain-rate' [12] as a preliminary EEG frequency indicator of general mental activation (mental arousal) and wake-sleep transition. The 'brain-rate' (arousal indicator) is used to reveal the patterns of sensitivity/rigidity of EEG spectrum and is defined as a mean frequency of the brain oscillations weighted over all bands of the EEG power spectrum. The mental arousal level is based on the quantum transition probabilities and is derived [13]:

$$A = e^{\frac{f_e}{f} \ln 2} = 2^{\frac{f_e}{f}} \tag{5}$$

where f_e is the equilibrium frequency at which the entropy is maximal and also corresponds to the dominant frequency with eyes closed (6 and 10Hz for children and adults, respectively). The functional dependence of arousal (A) on the frequency (f) is stated as $A = A(f)$.

$$f_b = \sum_i f_i \frac{V_i}{V} \text{ with } V = \sum_i V_i \tag{6}$$

where index (i) denotes the frequency band (for delta $(i)=1$, for theta $(i)=2$, etc) and V_i is the corresponding mean spectrum band potential. The formula for calculation of the brain-rate (f_b) is based on the same algorithm used for determining the centre of gravity (mass). The brain-rate represents the mean EEG frequency weighted over the 'whole' brain potential (power) distribution spectrum.

C. EEG Frequency Bands

The spectral analysis of the sleep EEG signal was performed using the short-time Fourier transform (STFT), in which the signal is divided into small sequential data frames and FFT applied to each one. In STFT analysis, the signal is multiplied by a window function $w(t)$ and the spectrum of this signal frame is calculated using the Fourier transform. Thus

$$STFT(t, f) = \left| \int_{-\infty}^{+\infty} x(\tau) w(\tau - t) e^{-j2\pi f \tau} d\tau \right|^2 \tag{7}$$

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where $x(t)$ represents the analysed signal.

Prior to performing the STFT on the EEG signal, the band-pass filtering was undertaken for extracting absolute and relative (band/total) EEG frequency bands, such as: delta (0.75-4.5Hz), theta (4.75-7.75Hz), alpha (8-12.25Hz), sigma (12.5-15Hz), beta (15.25-24.75Hz) and gamma (25-44.75Hz).

IV. RESULTS

The scoring of sleep stages was undertaken by visual inspection of the sleep technical specialist. After the approximately 2 hours of subject laying in bed 'trying' to fall asleep, the first wake (W) (first 30 epochs) stage was identified as an 'ideal first half of sleep onset' period of 15 minutes (1 epoch = 30 seconds). The second part of the sleep onset was Stage 1 (S1), from 30-41 epochs (lasting 5.5 minutes); stage 2 (S2) from 41-70 epochs (14.5 minutes), stage 3 (S3) from 70-85 epochs (7.5 minutes); and stage 4 (S4) from 85-117 epochs (16 minutes). The whole EEG data segment was 120 epochs (60 minutes).

No statistical analysis was performed on the processed data for any significant evidence. Figure 1 shows the analysis structure of sleep depth estimation. The results indicated a sharp suppression in coherence at 4 Hz (narrow band) from around 0.9 to 0.4 from W to S1, followed by the recovery period where coherence is fairly consistent around 0.5 Hz (Fig. 1a,b). Figure 2 shows the contour representation of the coherence at multiple frequencies of 0.7-0.8 magnitude in delta band during W stage, 0.6 at 10 Hz (alpha band) and 0.1-0.4 from 10-25 Hz (alpha-sigma-beta band). However, at the actual sleep onset (S1 start), the coherence at both delta, alpha and higher EEG bands, disappeared. The coherence of 0.7 at low delta and of 0.4-0.6 was narrowed to theta-sigma band. In the Fig. 1b and Fig. 2, it seems obvious that coherence at epoch 20 was equal to 0. This event at epoch 20 also had a strong influence on most of the measures (Fig. 1c,e,f and g). This is an artefact which was caused by the movement of the human subject. Similar artefact was also present at epoch 55, as shown in Fig. 1 and 2. Therefore, further artefact removal is required to process the data and improve the accuracy.

The 'brain-rate' at W was 7-7.3 Hz and during S1 to S2 it was decreased to 5.5 Hz (Fig. 1a,c). In the middle of S2, it covered back to 7 Hz and progressively decreased to 4 Hz during S3 and S4 until it reached the next sleep stage/cycle. The EEG relative and absolute band power results revealed a clear suppression in alpha band from S1 onset and remained fairly constant (absolute) and continuously decreasing (relative) during S2-S4, with an exception of irregularity and possible movement artefact in the middle of S2 (Fig. 1d,e). In terms of relative delta band (Fig. 1g), there is an increase in delta relative power from W to S1-4 transition. However, not a substantial change was revealed in absolute delta band in W

to S1 transition (Fig. 1f). There was a noticeable increase in absolute delta band from the middle of S2.
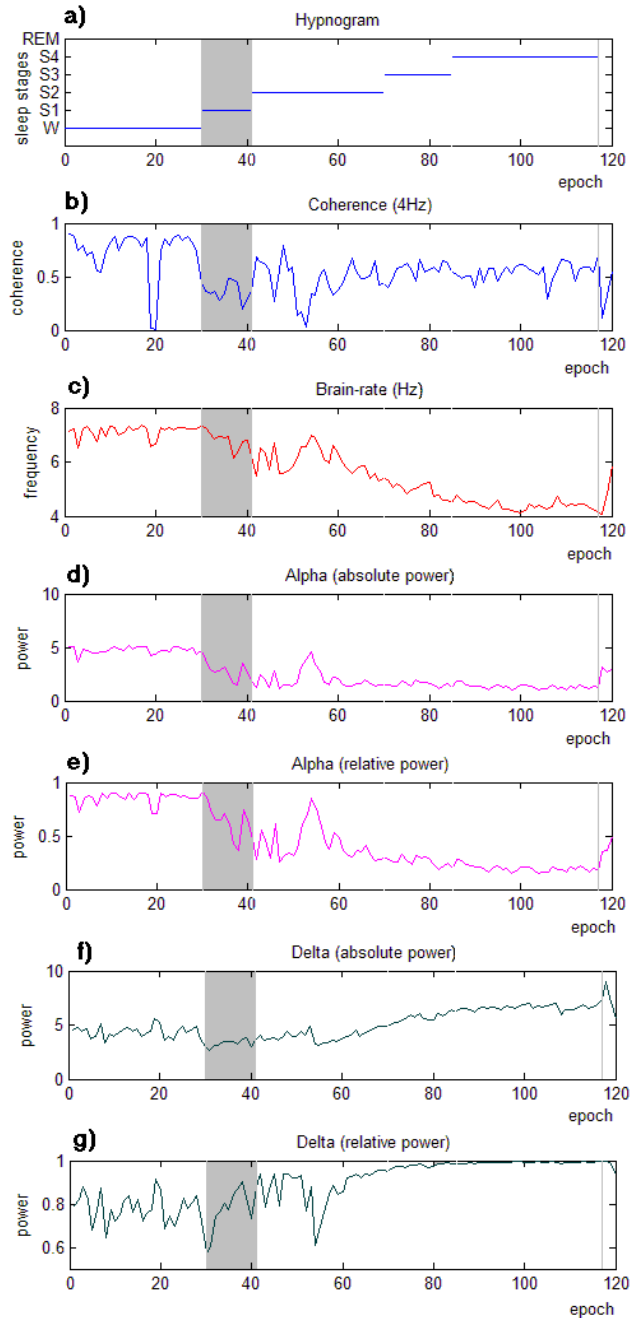

Fig 1 The analysis structure of sleep depth estimation, characterised by transitions between wake (W), stage 1 (S1), stage 2 (S2), stage 3 (S3), stage 4 (S4) and REM, is expressed by the hypnogram (a). The processed and analysed signals include delta 4Hz coherence with its intensity range from 0-1 (b); the brain-rate power frequency represented in Hz (c); the absolute (d) and relative (e) alpha power; and the absolute (f) and relative (g) delta power. The shaded area shown in all sub-plots indicate S1 or the sleep onset period.
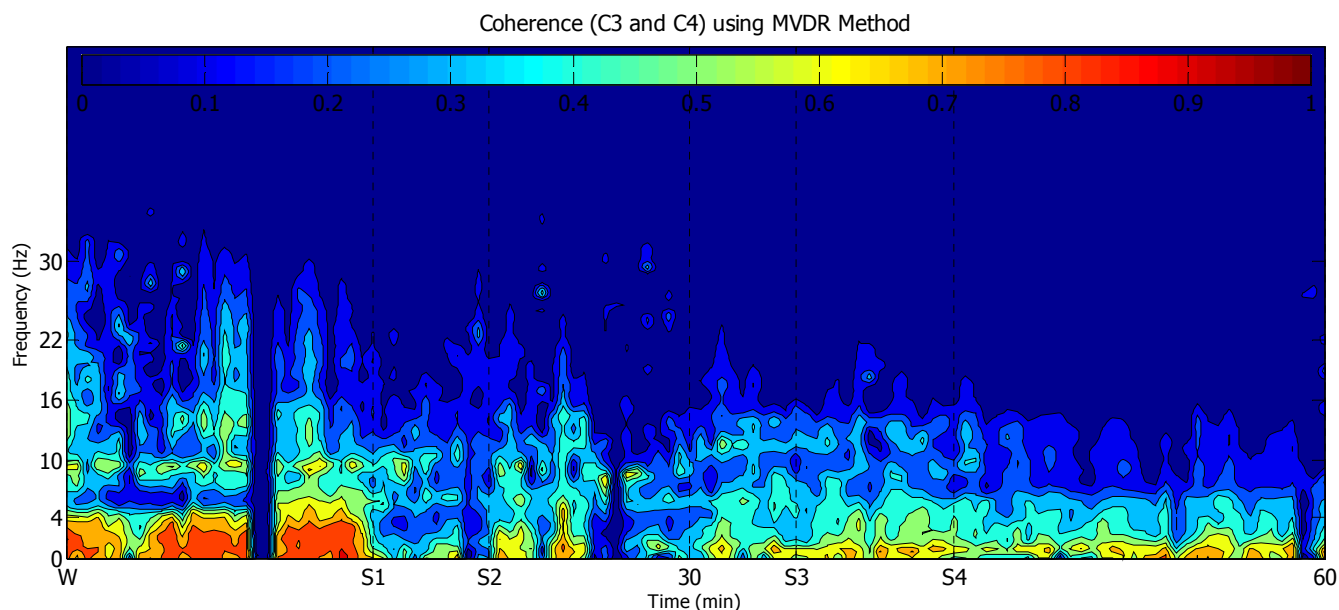
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Fig. 2 The inter-hemispheric MVDR coherence (C3-C4) was computed over the 60 min period (30 sec segments) consisting of W), S1, S2, S3 and S4. The legend represents the coherence intensity (0-1).

V. DISCUSSION AND CONCLUSION

It is possible to objectively estimate the wake-to-sleep transition, or drowsiness period from the EEG coherence, brain-rate and spectral delta/alpha band parameters [14]. There have been many studies, reporting on what is the 'right' moment of sleep onset [1-6]. We assume that this moment is from the onset of stage 1 (S1). All of our parameters evaluated have shown a potential as the solid indicators in sleep onset detection. However, further testing is needed on multiple human subjects, on-line algorithm design in neurofeedback applications, artefact removal techniques and investigation on EEG spatial and temporal dynamic changes during sleep onset [3]. For investigation of these spatiotemporal alterations, multiple EEG electrode sites are necessary to study the micro-structure.

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REFERENCES

[1] W.C. Dement and N. Kleitman, "The relation of eye movements during sleep to dream activity: an objective method for the study of dreaming, *J. Exp. Psychol.*, vol. 53, pp. 339-346, 1957.

[2] Rechtschaffen A, Kales A, "A manual of standardized terminology techniques and scoring system for sleep stages of human subjects", *Washington, DC: US Government Printing Office; Publication No. 204*, 1968.

[3] R.D. Ogilve, "Physiological Review, The process of falling asleep",*Sleep Medicine Reviews*, vol. 5, no. 3, pp. 247-270, 2001.

[4] A. Lubin, L.C. Johnson, M.T. Austin, "Discrimination among states of conciousness using EEG spectra", *Psychophysiol.*, vol. 6, pp. 122-132, 1969.

[5] T. Hori, "Spatiotemporal changes of EEG activity during waking-sleeping transition period", *Inter. J. Neurosci.*, vol. 27, pp. 101-114, 1985.

[6] R. Ogilvie, I.A. Simons, R.H. Kuderian, T. MacDonald, J. Rustenburg, "Behavioral, event-related potential (ERP), and EEG/FFT changes at sleep onset", *Psychophysiol.*, vol. 28, pp. 54-64, 1991.

[7] P. Badia, K.P. Wright Jr., A. Wauquier, "Fluctuations in single-hertz EEG activity during the transition to sleep. In: R.D. Ogilvie, J.R. Harsh (eds.), *Sleep Onset: Normal and Abnormal Processes. Washington: American Psychological Association*, 1994.

[8] D. Schwender, M. Daunderer, S. Mulzer, S. Klasing, U. Finsterer and K. Peter, "Spectral edge frequency of the electroencephalogram to monitor 'depth' of anaesthesia with isoflurane or propofol", *British Journal of Anaesthesia*, vol. 77, pp. 179-184, 1996.

[9] T. Morikawa, M. Hayashi, T. Hori, "Coherence analysis of the EEG in hypnagogic period", *Memoirs of the Faculty of Integrated Arts and Sciences III Hiroshima Univ.*, vol. 13, pp. 81-89, 1989 (Japanese; English abstract).

[10] J. Capon, "High Resolution Frequency-Wavenumber Spectrum Analysis", *Proc. IEEE*, vol. 57, no. 8, pp. 1408-1418, 1969.

[11] J. Benesty, J. Chen, Y. Huang, "A Generalized MVDR Spectrum", *IEEE Signal Processing Letters*, vol. 12, no. 12, pp. 827-830, 2005.

[12] N. Pop-Jordanova and J. Pop-Jordanov, "Spectrum-weighted EEG frequency ('brain-rate') as a quantitative indicator of mental arousal", *Contributions, Sec. Biol. Med. Sci.*, MASA, XXVI, 2; pp.35-42, 2005.

[13] J. Pop-Jordanov and N. Pop-Jordanova, "Quantum interpretation of mental arousal spectra", *Contributions, Sec. Biol. Med. Sci.*, MANU, XXV, 1-2; pp. 75-94, 2004.

[14] D. Cvetkovic, G. Holland and I. Cosic, "Evaluation of EEG Indexes for Estimation of Sleep Onset Depth: A Pilot Sleep Study", *Sleep and Biological Rhythms*, vol. 5, supp. 1, A42, 2007 (abstract published and presented at the WorldSleep07, 5th World Sleep Congress of the WFSRSMS, Cairns, Australia, Sep 2007).