Classification of Wakefulness and Anesthetic Sedation using Combination Feature of EEG and ECG

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***Abstract*—There have been lots of trials to classify a depth of anesthesia using diverse physiological indices. In this study, we classified wakefulness and propofol-induced sedation using combined electroencephalography (EEG) and electrocardiography (ECG) features for better classification performance. We extract each spectral band of EEG and very low frequency (VLF) of heart rate variability using spectrogram and low-pass filter, respectively. We used combined feature of EEG spectral bands and VLF and shrinkage-regularized linear discriminant analysis as a classifier. Our results show that combination of EEG spectral power and VLF can improve the classification performance between wakefulness and sedation from 95.1±5.3% to 96.4±4.2%.**

Keywords-Sedation; Propofol; Electroencephalography (EEG); Electrocardiography (ECG); Sigma Frequency Power; Very Low Frequency (VLF)

I. INTRODUCTION

In many studies, medical and surgical procedures have been performed under anesthetic states for reversibly changing consciousness states for various safety reasons [1]. Sedation is the one of the commonly used anesthesia states. There are also many medical and clinical procedures performed under the sedation (e.g., dental surgeries and endoscopy) [2]. Although this sedation is required as much as general anesthesia (GA), researches about sedation levels are insufficient.

For discrimination of sedation level, central nervous system (CNS) and autonomic nervous system (ANS) parameters have been used, which are represented by the brain and heart signal, respectively. Traditionally, ANS parameters have been used for monitoring indices in anesthetic condition, involving cardiovascular parameter such as heart rate (HR) and heart rate variability (HRV) [3]. In addition to ANS parameters, recent anesthesia researches focus on CNS parameters measured by electroencephalography (EEG). Bispectral index (BIS) which is most commonly used EEG-based index for depth of anesthesia shows a great performance accuracies under GA. However, the BIS has a limitation that the discrimination accuracy of sedation has very poor performance [4]. In other EEG-based analysis, even though various algorithms were performed for discrimination of depth of anesthesia, these studies have focused on only classification of wakefulness and GA.

Also, there are other approaches for depth of anesthesia, which consider not only one physiological parameter, but also various kinds of physiological parameters together. However, these studies have been conducted under only sleep or epilepsy situation [5,6]. Still, these methods combining features of different modality have not been tried for the discrimination of wakefulness and sedation.

In our study, we focused on the classification of wakefulness and sedation. We used patient-controlled sedation (PCS) to induce and maintain more obvious sedation. We also combined two kinds of physiological parameters together, one of which represents CNS and the other is ANS parameter. Unlike other studies using whole brain area, we only used five EEG frontal channels for classification of wakefulness and sedation. First, we extracted EEG time-frequency spectrum and also extracted VLF from HR calculated from electrocardiography (ECG). In addition, we observed changing pattern of EEG spectral power and VLF with sedation levels. Secondly, we classified wakefulness and sedation, and then compared classification accuracy when using only EEG features and when using EEG and ECG features together.

II. MATERIALS AND METHODS

A. Experimental Setup

The EEG signals were measured by the BrainAmp Amplifier and 64 Ag/AgCl EEG electrodes (Brainproducts GmbH, Germany) in this experiments. ECG was noninvasively and monitored by a standard patient monitoring system (solar 800M, GE, USA). PCS technique was performed by Perfusor Space (B Braun Melsungen AG, Germany) for maintaining sedation states. Additionally, we recorded BIS value using a Brain Monitor (BIS™, Covidien, USA). Two kinds of propofol doses were intravenously administered to subjects, 0.5mg/kg and 0.3mg/kg in the high dose case and middle dose case, respectively. Lock out time of high dose case was 3 min and middle dose case was 1 min. After injecting propofol, next injection was not administered during designated lock out time.

All subjects rested supine on a dental chair. All process was performed with closed eyes. Headset was provided to subjects for auditory cues “press the button” which were continuously given at average 10s inter-stimulus interval (ISI) during whole

experiment process. ISI is randomly assigned within 9-10s in each cue to avoid expectation of command timing. We set up loss of consciousness (LOC) and recovery of consciousness (ROC) points as the first points that subject did not press the button for consecutive five times and the first points that subject pressed the button for consecutive five times, respectively. Eighteen subjects participated in this experiments. Nine subjects participated in high dose and other nine subjects involved in middle dose propofol experiments.

This study was approved by the institutional review board (IRB) at Seoul National University Dental Hospital (CME15002) and written informed consent was obtained from all subjects. All experiments were performed under anesthesiologist's monitor.

B. Experimental Paradigm

This experiment was broadly divided into two sessions, resting session and main sedation session. During experiment process in resting session, importantly, propofol was not injected even though subjects pressed the button.

The next session called main sedation session, when subjects repeatedly lose and recover their consciousness, was divided by two parts. In the first part named injection part, propofol was injected when subject pressed the button. Infusion pump stopped working when subject went into the last unconsciousness states so ROC occurred without injection. The period after the last ROC was called as recovery part, where command and response (pressing the button) process lasted for more 5 minutes without any injection.

C. Data Acquisition

EEG signals were recorded from 62 channels (FP1, FP2, AF1-4, Fz, F1-4, F7, F8, FC1-6, FT7-10, Cz, C1-6, T7, T8, CPz, CP1-6, TP7-10, Pz, P1-8, POz, PO3, PO4, PO7-10, Oz, O1, and O2) of the modified International 10-20 systems with 1000 Hz sampling rates and down-sampled to 100 Hz. The Ag/AgCl sensors were mounted on a cap (actiCAP, Brainproducts GmbH, Germany). We used five frontal channels (AF1, AF2, F1, Fz and F2) for further analysis. ECG signals were recorded from ECG lead with 74 Hz sampling rate with diverse safe monitoring system.

D. Data Analysis

1) EEG spectral analysis : For EEG feature extraction, the spectral analysis of EEG was performed in time domain. Power spectral density (PSD) of five traditional spectral bands were computed in each five channels using the short-time fourier transform (STFT). The window length for spectral analysis was 2s with 1.9s overlap, and 1024 frequency points. The five spectral bands consisted of delta (0.5-4.5 Hz), theta (5-8.5 Hz), alpha (9-12.5 Hz), sigma (13-16.5 Hz) and beta (17-30 Hz) band.

To clarify how EEG spectral power changed over the sedation levels in time domain, we computed spectrogram using multitaper spectral analysis and observed the change of EEG spectral power. Multitaper spectrogram was computed with 2s window lengths and 1.9s overlap. We set time-bandwidth product and number of tapers as 2 and 3, respectively. We computed each EEG spectral power in the time domain to clarify the change of spectral power.

2) Heart rate analysis : To extract features from ECG, band-pass filtering was performed between 5-30 Hz for reducing noise in the preprocessing step, and the peaks of R-waves on the ECG were detected. Prior to detecting the R-wave peak times, the R-waves were interpolated at 1036 Hz for improving the time resolution of the detection. After R-wave detection, we used cubic spline interpolation methods for converting the non-equidistantly sampled RR interval time series (RRI) to equidistantly sampled new RRI having 100 Hz sampling rate. The resulting series of events were used for the HR computation. To extract VLF from HR, we applied low-pass filters at 0.04 to the HR [7].

3) Feature extraction : EEG spectral power of frontal five channels were collected and averaged in time domain. From this time domain, we extracted 100 trials with 2s per one trials. In resting session, sequential 100 trials were extracted from the starting point. Injection part of main sedation session was divided into several unconsciousness cycles. We extracted 100 trials from each cycle. As a result, we have different length of data according to subjects because each subject has different cycles. VLF trials were also extracted on the same time points of EEG spectral power trials after moving average of 100s.

4) Classification : Our data has some characteristics that the dimension of data is a quite high and the number of data is relatively small, which can result in errors from biased covariance. For compensating these systematic errors, we used regularized LDA with shrinkage for classification [8]. For comparison with commonly used previous methods for discriminating sedation levels, we also classified wakefulness and sedation using only EEG spectral power with same classification methods.

III. Results

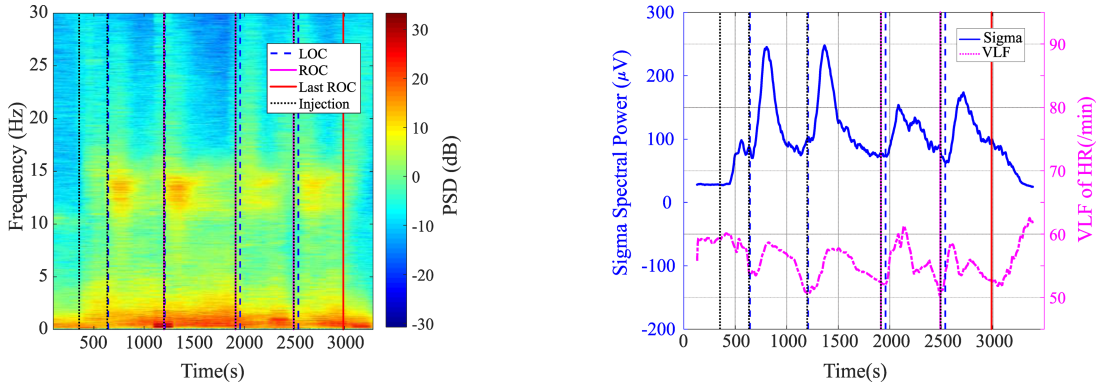
A. EEG Spectral Analysis and Heart Rate Analysis

EEG spectral frequency power showed specific changing trend at LOC and ROC as shown in the multitaper spectrogram. For instance, sigma power has a particular pattern over propofol injection and LOC (Figure. 1a).

Sigma power starts to increase at the point of propofol injection and at a certain point in unconscious states, it starts to decrease until ROC point. Repeatedly, the sigma power increases at the next injection of propofol. VLF repeatedly increases and decreases in the period between the injection of propofol and ROC points. Although decreasing points are subject-dependent, the increasing points are the same as the injection or ROC points among almost all subjects (Figure. 1b).

B. Classification with Shrinkage-RLDA

When we used combined EEG spectral power and VLF, the grand average of classification accuracy was 96.4±4.2%. However, when we used only EEG spectral power as features without VLF features, the grand average of classification accuracy was 95.1±5.3% (Table 1). We identified that classification accuracies of combined EEG and ECG features is



(a) (b)

Figure 1. (a) Multitaper spectrogram and (b) spectral power of sigma and VLF (red block for sigma band, black dot-line for injection, blue dash-line for LOC, magenta line for ROC with injection, and red line for pure ROC).

significantly higher than that of only EEG features using paired t-test ($p<0.05$).

IV. Discussion and Conclusion

In this paper, we could find several physiological changes over the sedation levels. Sigma power and VLF show constant trend over sedation levels, from which we can conclude that sedation effects on the heart and brain. This observation gave the potential for the usefulness of HRV for discriminating wakefulness and sedation. Secondly, we used combined EEG and ECG features for classification between wakefulness and sedation. The features used in this experiment are the EEG spectral power and VLF. Shrinkage-RLDA was used for a classifier due to lack of data with high dimensionality. It showed better accuracies than when using only EEG spectral power as features. This demonstrates that EEG signals combined with VLF can be a better index for discriminating sedation. This results show that we can use only five EEG frontal channels for discriminating sedation if with ECG channel. This can be a very useful and practical tool for monitoring sedation.

In this work, we only classified two class, one of which is wakefulness and the other is sedation. For real applications, it will be our further research issue to develop a method for discriminating more sophisticated sedation levels.

TABLE I. Comparison of Classification Accuracies between the Cases using EEG and Combining EEG and ECG

Sub.	Accuracies		Sub.	Accuracies	
	EEG (%)	*EEG + ECG (%)*		*EEG (%)*	*EEG + ECG (%)*
1	95.1	96.4	**10**	99.8	100.0
2	96.8	97.4	**11**	98.7	98.7
3	86.7	91.6	**12**	99.0	99.0
4	88.5	92.6	**13**	100.0	100.0
5	88.2	93.1	**14**	90.9	93.3
6	84.7	86.3	**15**	98.9	98.9
7	93.4	99.6	**16**	100.0	100.0
8	98.2	98.4	**17**	99.5	99.5
9	89.8	89.3	**18**	98.7	100.0
			Ave.	**95.1±5.3**	**96.4±4.2**

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