Long-term Sleep Monitoring System and Long-term Sleep Parameters using Unconstrained Method

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***Abstract*—Sleep is a most important part of a human life. The Polysomnography (PSG) provides a standard method for diagnosing and estimating sleep diseases. But PSG requires heavy equiped and patient constrained method. This paper describes an up-to-date method to estimate long-term sleep quality and reliable sleep parameters through non-invasive and unconstrained method. We developed a bed type sensor which is implemented by a pneumatic pressure transducer, air-mattress and balancing tube. It provides a patient with comfort to sleep and bring a relief from psychological stress caused by PSG. We separates ballistocardiac signal, respiration signal, snoring signal and a body movement from pneumatic pressure signal. The ballistic heart rate (B-HR) and ballistic heart rate variability (B-HRV) extracted from ballistocardiac signal is compared with heart rate (HR) and heart rate variability (HRV) calculated from standard electrocardiography (ECG). And respiration signal and snoring signal is compared with flow-pressure meter. A body movement signal is compared with position sensor. We test two patients for three month and calculate the reliability of each signals. The correlation of HR and B-HR is 98.1%. HRV and B-HRV is 95.3%. And the correlation of respiration is 98.6%, snoring is 96.3%. These parameters are useful to sleep stage estimation and sleep apnea detection. And we introduce the indices which reflect a relation between daily psychophysiological stress and sleep patterns from the test of long-term monitoring. The indices are compared with list of questions that reflect a daily life stress with ten degrees. The proposed sleep monitoring system can not replace a standard PSG with Rechtschanffen and Kales (R-K) method as a sleep staging method. However, the new method is better of the fact that it does not physically restrain the patient and does not induce psychological stress. Also, it can be applied in long-term sleep monitoring system.**

I. Introduction

SLEEP occupies one third part of human life and the abnormal sleep patterns induce many daytime diseases. So it is a important work that monitoring and caring the sleep.

Respiration and heart beat movements present fundamental information for the monitoring of the human body in healthcare situations. Although many non-invasive methods of acquiring such physiological information have been studied, most of the methods are limited in that sensors or electrodes need to be attached to the body surface (Folke et al 2003)[1]. In the applications of sleep analysis and the monitoring of neonates, as well as home healthcare monitoring, an unconstrained measurement is highly preferred and in some cases absolutely required. In order to accomplish this requirement of unconstrained measurements, Hernandez et al (1995) [2], Chow et al (2000)[3] and Watanabe and Watanabe (2004)[4] used an air mattress system under the body as a sensor for respiration movements. The respiration and heart beat movements add pressure to the air mattress, which changes the air pressure in the mattress. Tanaka et al (2002) suggested a phonocardiogram sensor on the bed for either an air mattress or a water mattress, Jacobs et al (2004)[5] used PVDF (polyvynilidenefluoride) sensors in the form of a two-dimensional array, which is highly accurate but expensive, and Spillman et al (2004)[6] used multi-modal optic sensors. In spite of these trials, no leading method exists for the unconstrained measurement of respiration and heart beat movement that is considered to be stable, accurate, and economic..

In this study, Chee et al (2005)[7] method using balancing tube and air-mattress for unconstrained measurement of respiration and heart beat movement was applied.

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II. Subjects and Materials

A. Subjects

One healthy male volunteer (mean age ±standard deviation, 28.5±2.1) participated in this study. Subject participated after giving informed consent to the protocol that was reviewed and approved by the Ethics Committee.

B. Materials

First of all, MP150 the computer-based physiological signal acquisition system of BIOPAC Inc. and other transducers (ECG100C-Electrocariogram Amplifier, TDS201 -Respiratory Effort Transducer, TDS117-Airflow Transducer for Medium Flow Pneumotach) were used to obtaining the reference signals. As shown in figure 1(b), an air mattress system containing 19 air cells was constructed. Even though the patterns of waveforms change according to the contact cells where measurements were made, information on respiration and heart beats can be found from any pair of cells located between the neck and the waist of the subject. In this experiment, the two cells located on the backside of the chest and abdominal region were chosen

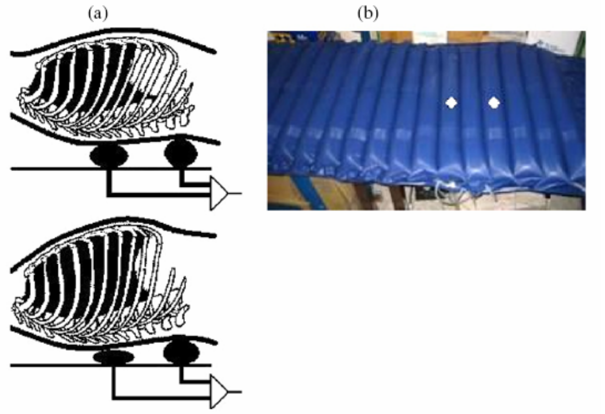

Fig. 1. (a) As a result of respiration and heart beat movements, the air cells under the body are depressed, causing the pressure inside to change. The upper figure is during the exhale movement, and the lower is during the inhale movement. (b) The air mattress sensor system implemented to perform measurements during sleep

(marked with white circles in figure 1(b)) where a large respiration signal, as compared with other locations, could be obtained. Each cell is in the form of a cylinder with a diameter of 110 mm and is made of a polyurethane sheet; the cell was filled with air at 10 kPa at the beginning of the experiment. In this range, when the subject lies on the mattress, the air pressure in the air compartment increases to the pressure imposed by the body weight while this system retains its elastic properties. The air cells and sensors that are assembled on the circuit board were connected with a silicone tube of an inner diameter of 4 mm, an outer diameter of 8 mm and a length of 1.0 m. The length of the balancing tube was 0.5 m. In the middle of the balancing tube, there was a rotational valve that could control the open area precisely. The air pressure sensor was a NPC-1210 (100 mV/5psi) from Luca NovaSensor R and the signal was amplified 3000–30,000 times in order to detect the small respiration and heart beat signals. The analog signal was sampled at 1000 Hz and digitized into 12 bits with a NI-DAQ from National Instrument Co.

III. Methods

A. Signal Separation

The aim of this processure is to analyze and comparing the air-mattress signals with reference signals. A further desired effect was elimination of the noise component from both air-mattress signals and reference signals.

Digitized air-mattress signals were separated by filtering for comparing with the reference signals. Figure 2 presents separated respiration and heart beat movements signals.

Figure 2(a) is a unseparated air-mattress signal.

Figure 2(b) is the ECG from ECG100c and the air-mattress ballistocardiogram (AM-BCG). BCG was filtered by FIR band-pass filter of pass band frequency 0.8~15Hz.

Figure 2(c) is the respiration signal from TDS201 and air-mattress respiration (AM-RES) effort signal from air-mattress..

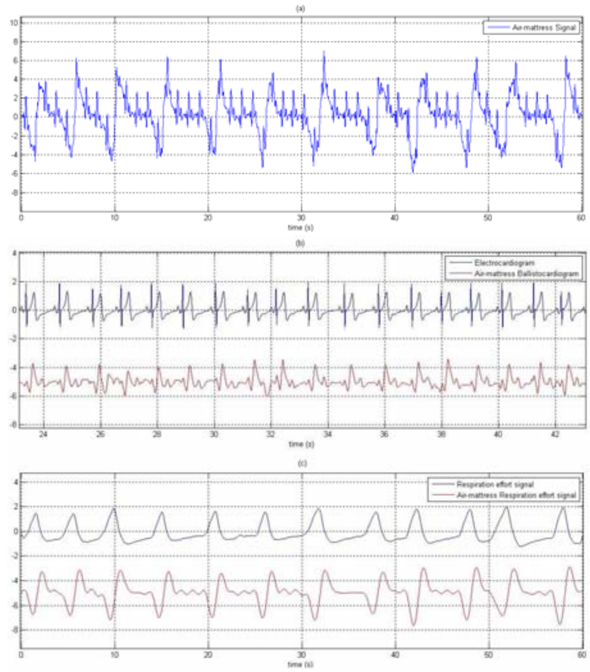

Fig. 2. (a) Unseparated air-mattress signal for 60 sec. (b) Cardiac signals for 20 sec. Upper : ECG, Lower : separated air-mattress ballistocardiogram (AM-BCG). (c) Respiration effort signals for 60 sec. Upper : respiration effort signal, Lower : separated air-mattress respiration signal (RM_RES).

The snoring event signal is separated by FIR band-pass filter of pass band frequency 50~100Hz.

B. Cardiac Signal Processing

The aim of this processure is to detect and measure a RR interval from the ECG waveform. In keeping with the near real-time requirements and low processing load, a necessary requirement is that the algorithm is simple but accurate. The approach taken is based on a simplification of the QRS detection scheme of Pan and Tompkins [8].

This processes the sampled data as follows:

1) Differentiation—obtain information about signal slope.
2) Squaring the derivative—intensify frequency response curve of derivative.
3) Integrate over moving window—obtain information about slope and width of QRS complex; also filters out some unwanted spikes.

For QRS detection, a threshold set is computed during an initial learning stage (lasting 8 s): the upper threshold is calculated from 0.4 times the average maximum on the integrated signal; from this, a lower threshold is calculated by another factor of 0.4. During the detection process, the current integrated moving window value is compared with the upper threshold. If this threshold is exceeded, an wave onset is assumed; QRS is confirmed by scanning backward (up to 100 ms) for a dip below the lower threshold. These threshold values are continually adjusted with each new QRS so as to compensate for variations in ECG baseline. Figure 3(a) is a detected R-peaks and ECG.

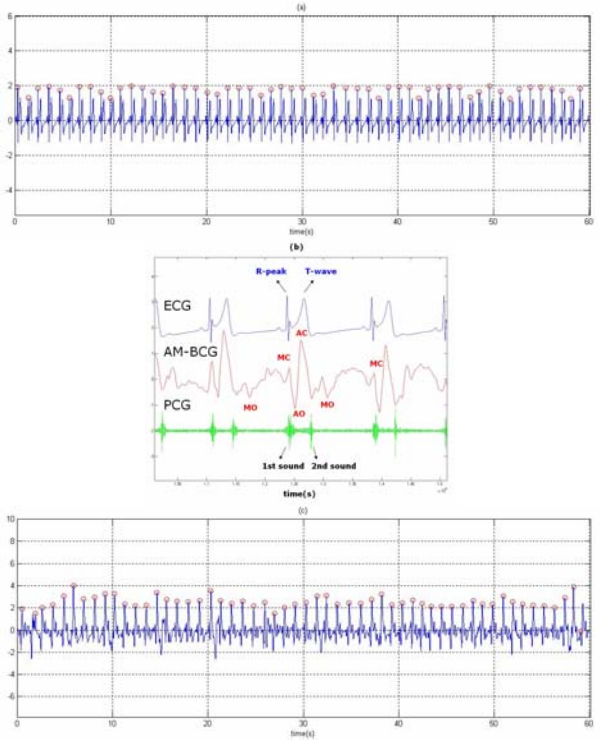

Fig. 3. (a) Detected R-peaks (red O) and ECG (blue line). (b) ECG (blue line), AM-BCG (red line) and phonocardiogram (PCG, green line) for varifying AC-peak, (c) Detected AC-peaks (red O) and AM-BCG (blue line).

For detecting cardiac event which is reflects the R-peak in AM-BCG, we simply find the max point between the R-R peaks. The peak was named AC-peak which means the peak is made by the aortic-valve closing (AC) action. Figure 3(b) is a ECG, AM-BCG and phonocardiogram (PCG) for varifying the peaks which is made by the cardiao ballistic actions from AM-BCG. And figure 3(c) is a detected AC-peaks and AM-BCG. These parameters, R-peaks and AC-peaks, were transformed to medical parameters, in particular, RR intervals(distance) and B-RR interval (Ballistocardiac RR intervals induced by AC-peaks).

C. Respiration Effort Signal Processing

To calculate the respiration rate in both reference respiration signal and a AM-RES, a dominant frequency detection algorithm based on short-time fourier transform (STFT) was applied. Fourier transform is a powerful tool to study the frequency content of signals but has the draw-back that it does not provide any localization in time. To obtain localization in time, STFT was introduced, where the signal is first multiplied by a window function and then the Fourier Transform of the windowed signal is taken. STFT, also called windowed Fourier transform, is the localization of the Fourier transform using an appropriate window function centered around a location of interest. Since the STFT is a localized Fourier transform, a rectangular window would have poor frequency localization, therefore smoother windows are preferred. Triangular window, Hanning window, Gaussian window and Hamming windows are some smoother windows

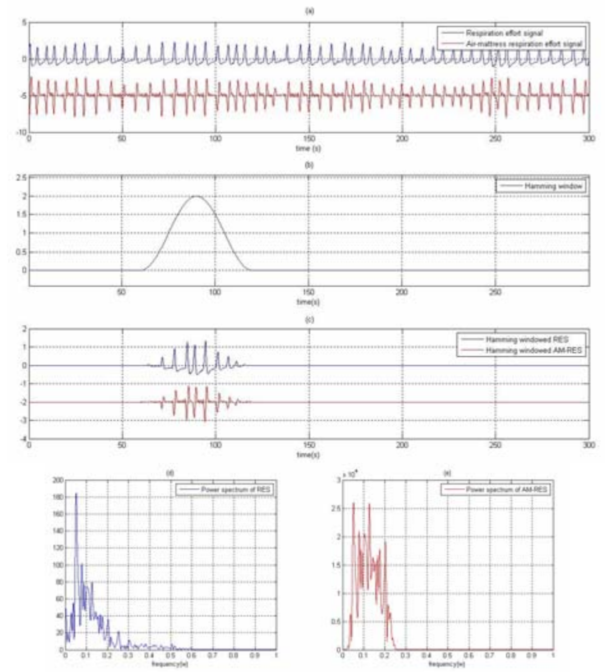

Fig. 4. (a) Respiration effort signals. Upper : respiration signal from Biopac, Lower : air-mattress respiration signal. (b) Hamming window (c) Upper : Hamming windowed Respiration effort signal, Lower : Hamming windowed Air-mattress Respiration effort signal. (d) Power spectrum of reference respiration effort signal. (e) Power spectrum of air-mattress respiration effort signal.

designed for data analysis[25]. In this study, the hamming window is used.

$$STFT(\omega,\tau)=\int w(t-\tau)f(t)e^{-jwt}dt$$

Because frequency components of the respiration signal is very low, under 2Hz, the window size is 60 seconds. And every 60 seconds, the hamming window is multiplied to respiration signals and transform to frequency domain using Fourier transform. The dominant frequency is detected in frequency domain respiration signals by finding the max amplitude of spectrums of they. When the dominant frequency components were founded then a inverse numbers were calculated to obtain respiration rate in 60 seconds.

Figure 4(a) is a respiration effort signals for 5min.

Figure 4(b) is a hamming window to localizing the respiration signals.

Figure 4(c) is a hamming windowed respiration signals.

Figure 4(d), (e) are power spectrums of respiration effort signals using STFT.

Using the respiration rate signals, respiration signal correlation was calculated.

D. Snoring Signal Processing

Snoring event in reference signal is detected manually for 60 seconds. Also the air-mattress snoring event is detected by manual description for 60 seconds. Both events were compared and correlation coefficient was calculated.

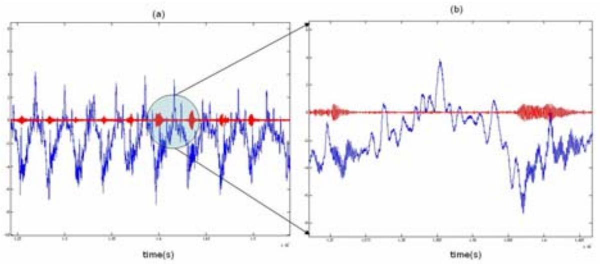

Fig. 5. (a) Snoring event signals. Red line is a reference snoring event signal using Biopac TDS117-Airflow Transducer and blue line is a air-mattress snoring event signal. (b) Time-scale zoomed graph of (a).

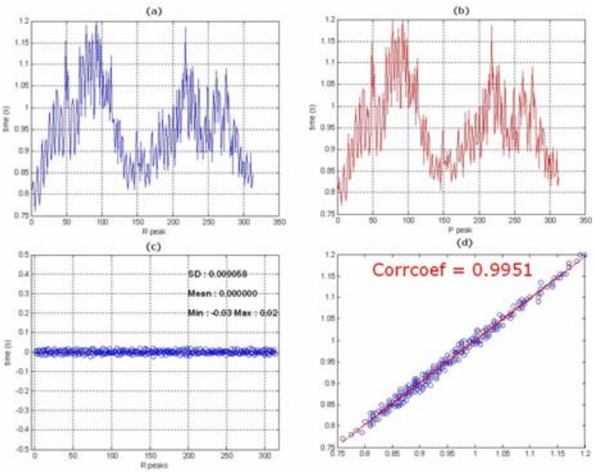

Fig. 6. (a) RR interval graph in every heart beats. (b) B-RR interval graph in every heart beats. (c) Difference error between RR interval and B-RR interval in every heart beats (d) Correlation coefficient of RR interval and B-RR interval.

Figure 5(a) is a snoring event signals. Red line is a reference snoring event signal using Biopac TDS117-Airflow Transducer and blue line is a air-mattress snoring event signal. Figure 5(b) is time-scale zoomed graph of figure 5(a).

IV. RESULT

A. RR vs. B-RR

Figure 6(a) is a RR interval graph in every heart beats.
Figure 6(b) is a B-RR interval graph in every heart beats.
Figure 6(c) is a difference error between RR interval and B-RR interval in every heart beats.

Figure 6(d) is a correlation coefficient of RR interval and B-RR interval. The correlation coefficient is 0.9951 which shows high reliability.

B. HRV vs. B-HRV

A HRV could be calculated using RR interval interpolation 2Hz and frequency domain analysis using Fourier transform. The low frequency component could be obtained by summation of the band power 0.04~0.18 Hz. The high frequency component could be obtained by summation of the band power 0.18~0.40 Hz. As the same way, the ballistocardiac heart-rate variability (B-HRV) could be calculated using B-RR interval signals.

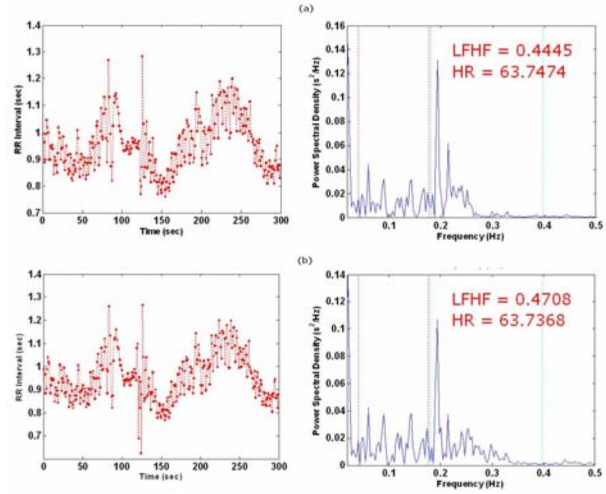

Fig. 7. (a) Reference signal, ECG : RR interval signal and interpolated power spectrum signal of it (b) AM-BCG : B-RR interval signal and its interpolated power spectrum signal.

Figure 7(a) shows HRV from ECG and power spectrum of RR intervals. The LF/HF ratio of ECG is 0.4445.

Figure 7(b) shows B-HRV from AM-BCG and power spectrum of B-RR intervals. The LF/HF ratio of AM-BCG is 0.4708.

The correlation coefficient of HRV and B-HRV is 0.9409 which is a good proof for our aims that air-mattress is a reliable sleep monitoring system offers clinically effective physiological parameters.

C. RR vs. B-RR

The reference respiration rate was calculated for 5 min using the dominant frequency algorithm. And the air-mattress respiration rate (B-RR) was calculated for 5 min using the same way.

The correlation coefficient of respiration rate between reference respiration signal from Biopac TDS201 and AM-RES is 98.6% which is reliable factor.

D. SNORING vs. B-SNORING

The snoring event was recorded for 5 min and manually described. The correlation coefficient of respiration rate between reference respiration signal from Biopac TDS201 and air-mattress snoring event signal is 98.6% which is reliable factor.

V. DISCUSSION AND CONCLUSION

The goal of this study was to develop a method of monitoring the cardio-respiratory signals in an unconstrained condition. Even though an air mattress system has many advantages, the fundamental difficulty is in detection of the very small periodic signals of respiration and heart beats from the large signals associated with body weight and postural changes. A balancing tube was used to set up a robust sensing device that could be used despite the existence of postural changes during the measurement period. Through the use of this balancing tube, the frequency of interest can be selected. A shortcoming of this balancing tube is the decrease in sensitivity of the measurement based on the properties of the passive filter. The shorter its transient time to equilibrium, the narrower the pass band of the band pass filter and the lower its gain, this means opening the valve in the balancing tube more. Because the frequency range of respiration is quite low (0.1-0.5 Hz in a normal adult), it decreases the sensitivity of the system. During the implementation, a decision between sensitivity and robustness should be made.

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