Monitoring Respiration and Cardiac Activity During Sleep Using Microbend Fiber Sensor: A Clinical Study and New Algorithm

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***Abstract*— We report results from a clinical trial for monitoring respiration and cardiac activity of patients during sleep using microbend fiber sensor. This sensor is used to acquire respiratory and heart beat information. We have collected reference data from standard Polysomnography and data from microbend fiber sensor on 22 patients. A new algorithm is developed to calculate breathing rate and heart rate simultaneously. The Bland-Altman analysis demonstrates the measurements have good accuracy for monitoring purposes compared with the standard Polysomnography. An accuracy of 1.06bpm for breathing rate and 3.32bpm for heart rate has been validated for 30s averaging time although there were significant signal distortions under sleep conditions.**

I. Introduction

Sleep is very important to our health. There are many different sleep disorders, the most common of which is obstructive sleep apnea (OSA). This condition is associated with other significant medical diseases like ischemic heart disease, hypertension, stroke, and poor psychological well being. Currently, the gold standard for objective sleep assessment is an overnight in-laboratory Polysomnography (PSG) recording, which involves continuous recording of multiple physiological parameters such as respiratory effort, air flow, heart rate and oxygen saturation as a patient sleeps. Unfortunately, a typical PSG involves multiple cables and leads being attached to the patient for the duration of the study. This causes significant discomfort and is intrusive to the patient's sleep. Such disturbance alters the patient's natural sleep structure and could consequently lead to inaccurate results. In addition, PSG test is very expensive and its equipment is unlikely to be used at patient's home. To overcome the above problems, many "contactless" and cableless monitoring sensors and methods have been studied and developed for sleep application. Breathing rate (also a proxy for breathing effort) and heart rate are two important clinical parameters. They are widely used for monitoring and diagnosis of diseases for various patients. In this paper, we focus only on monitoring of respiration and heart rate during sleep. Generally speaking, current sensors and methods for monitoring respiration and heart rate may be classified as: electronic, optoelectronic and fiber optic sensor method. In electronic method, Emfit sensor [1], air mattress sensor [2], liquid pressure sensor [3], electrocardiography (ECG) sensor [4], Doppler radar sensor [5], and acoustic sensor [6] were used in the research. Optoelectronic method used photoplethysmography (PPG) sensor [7-8]. This type of PPG sensors need skin contact. In fiber optic sensor method, many of such sensors have been proposed for monitoring of breathing rate, heart rate and body movement in the past [9-11]. Fibre Bragg grating (FBG) based sensors are used for the simultaneous measurement of breathing rate and heart rate [9-10]. However, its wavelength detection technology is too complex and expensive for sensor fabrication and instrumentation. A fibre optic statistical mode (STM) sensor and a high order mode excitation (HOME) sensor have also been proposed for breathing rate and heart rate measurement [11]. These sensors required highly coherent light source and a bulky high order mode generator. The interferometric fibre optic sensor [12] has very high sensor sensitivity, but its sensing system and interrogation is complex and expensive. In our group, we proposed and developed a simple and low cost microbend fiber sensor for respiratory monitoring and MRI imaging gating during MRI [13]. This microbend fiber sensor has many distinctions over other fiber optic sensors, e.g., simple system configuration and low cost. A BOM cost of USD50 for this sensing system (simple version) is possible based on retailed prices of materials. The common disadvantages of microbend fiber sensor are fluctuations of light source and variations of transmissivity of optical fiber path, which affect sensor's accuracy and stability greatly. However, our microbend fiber sensor for this application doesn't share the common disadvantage of conventional microbend fiber sensor as intensity modulation based sensor. Compared with electronic sensors, our sensor has another advantage: our sensor is not point sensor, each location of fiber can sense, and its sensing area is scalable with little cost increase while electronic sensors are point sensors and many sensors are needed to cover large sensing area. In Reference [13], the duration of monitoring is about 45min and the patients were required to lie down still. Although the study demonstrated that the microbend fiber sensor was excellent for monitoring of respiration and imaging gating in MRI environment, the patients were required not to move in the MRI bed and the movement of patient's body was under control. This is not practical for sleep monitoring application. For sleep monitoring, the sensor should be enable long-term continuous monitoring of cardiopulmonary parameters

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without constraining the patient's body movement in the bed and the need for any prior patient preparation or skin contact. This raises a critical question whether our fiber sensor is feasible and practical to monitor respiration and cardiac activity of patients during sleep. In the publications [9-13], the measurement conditions were too ideal and not realistic, compared with the random body movements associated with natural sleep. To our knowledge, this is the first study to address this question. The main objective of this study is to evaluate the accuracy of our microbend fiber sensor as medical device for simultaneous measurement of breathing rate and heart rate during sleep using newly developed algorithms. In doing so, we built highly sensitive microbend fiber sensor, developed new algorithms and conducted a clinical trial in the sleep lab, National University Hospital, Singapore. The results are reported below.

II. System and Data Collection

A. Microbend Fiber Sensor

Fig.1 shows schematic diagram of microbend fiber sensor system. Our sensing system consists of a microbend fiber sensor mat, a transceiver, a NI USB 6216 data acquisition (DAQ) card, and a computer. Transceiver has an emitter at 1300nm, a light detector, amplifiers, filters, and other simple circuits. A low pass filter at 250Hz is built into the transceiver to remove high frequency noise. The output light from transceiver is input into one end of the sensor mat. The light come out from another end of the sensor mat is detected by the light detector and then converted into electrical signals in the transceiver. These electrical signals are acquired by the DAQ card at a sample rate of 50Hz and 16 bit resolution. The computer is used to record the data for whole night during testing. The microbend fiber sensor mat consists of a section of graded multimode fiber clamped between a pair of microbenders as showed in Fig.2. As the pressure is applied to the sensor mat, the displacement between two microbenders changes, the light intensity of the clamped multimode fiber changes with patient's body vibrations caused by respiration and heart beating. The light intensity in the microbending fiber is then modulated by the body vibrations. The modulated signals are extracted as respiratory signals and Ballistocardiogram (BCG) signals.

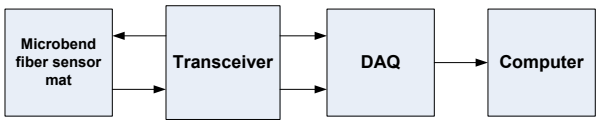


Figure 1. Schematic diagram of the microbend fiber sensor system

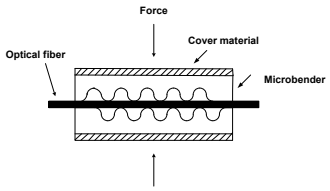


Figure 2. Microbend fiber sensor mat

Figure 3. Photo of sleep lab: we were testing our system

Fig.3 shows a photo that we were testing our sensing system in the sleep lab. The sensor mat (size: 300x150x2mm) was put under bed covers in the pillow (head) and shoulder area for clinical trial. This system was delivered to hospital team to solely run clinical trial. Technical team was not involved during the period of data collection.

B. Data collection

This clinical trial was approved by the ethics committee of National University Hospital (NUH) and the recorded data were obtained with patient consent. During the sleep study, recordings were simultaneously obtained using both microbend fiber sensor and PSG's standard thoracic belts and ECG moniters.. This study included 22 patients who were clinically suspected to have Obstructive Sleep Apnea from the Ear, Nose and Throat clinic and the Respiratory clinic of NUH. These subjects were scheduled for an overnight PSG at the NUH Sleep Lab. The data from our microbend fiber sensor were digitized with 50Hz sampling rates and 16 bit resolution using NI USB 6216 DAQ card and stored using an acquisition program developed using Labview 8.2. The PSG data were recorded in the synchronized time.

III. Signal Processing of Microbend Fiber Sensor

Motion noises are common problem to all "contactless" sensors for vital signs measurement. Because the nature of sensor types is different, the signal processing methods maybe different to best fit the specific sensor type. We developed a new algorithm for simultaneous measurement of breathing rate and heart rate for our sensor.

A. Motion noise removal

This step is to remove noisy signals caused by big movements. Signals from the sensor mat were passed into a moving window. The difference of the maximum and minimum in the moving window was compared with a preset threshold. The noisy window segment was removed from original signal if the difference was larger than the threshold. The remaining segments were detrended by straight-line linear fitting and then concatenated. This is a very effective method to remove big body movement. As for small motion noises induced by talking, muscle tremor, etc, averaging, smoothing and histogram methods will be implemented to remove such noises.

B. Respiration Waveform Extraction

This step is to extract respiration waveform for respiration rate calculation. After movement noise removal, signals were smoothed by a moving average window of 100 points to remove unwanted spikes, followed by a moving average

window of 1000 points to obtain the baseline. The signals after subtraction of the signals and baseline were further smoothed using Savitzky–Golay method with a span of 60 points.

C. Respiration Rate Calculation

To calculate the respiration rate, local maxima points of respiration waveform were found. Intervals on x-axis between each pair of adjacent maxima points were calculated, converted to bpm (beats per minute) unit and stored in an array in order. Finally, each element in the array represented the instantaneous respiration rate in corresponding time interval.

D. BCG Waveform Extraction

In this step, BCG waveform was extracted from signals after movement noise removal with 5 filters as follow in order: Butterworth band pass filter of order 2 with 3dB frequencies of 5 and 25 Hz, Butterworth band pass filter of order 2 with 3dB frequencies of 0.8 and 3 Hz, Butterworth low pass filter of order 4 with 3dB frequency of 5 Hz, Butterworth high pass filter of order 8 with 3dB frequency of 0.1 Hz and Butterworth band stop filter of order 2 with 3dB frequencies of 0.1 and 0.5 Hz. Fig.4 shows a typical BCG waveform before and after filtering.

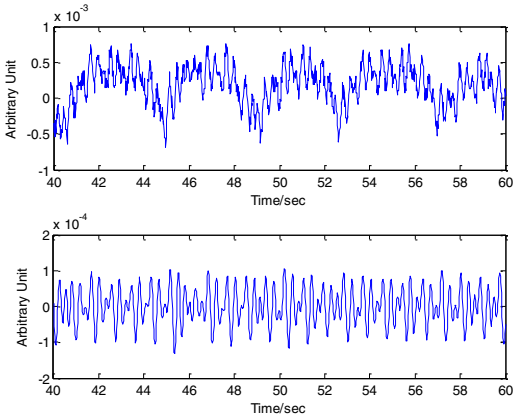


Figure 4. BCG waveform extractions. Upper: BCG signal after movement noise removal, Lower: BCG waveform after filtering.

E. J peak detection

A typical BCG waveform has IJK complex, which corresponds to the QRS complex of ECG waveform. Of all components of BCG waveform, the J peak is usually the easiest one to be identified. In this step, all local maxima points of BCG signals after filtering were found. The peak points that match our J peak feature requirements were selected to calculate peak to peak intervals. The IJK complex may be upside down occasionally. Therefore, the opposite peak values of signals were also considered for J peak detection.

F. Heart Rate Calculation

In this step, histogram method was used to eliminate the effect of false J peaks found in previous step, and calculate the heart rate of BCG signals. Peaks in the upside and downside were detected. The time intervals of adjacent J peaks were calculated, converted to bpm and stored into arrays. Heart rate values out of preset measurement range were removed. The elements in the arrays were divided into 10 equally spaced groups based on the maximum and minimum of heart rate values in the arrays. Figure 5 shows a typical histogram. The highest occurrence of that group was selected and reported as final heart rate result. The accuracy is dependent on averaging time.

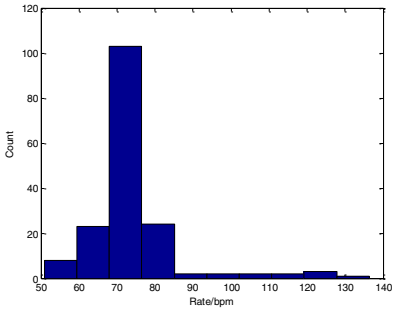


Figure 5. Histogram of the heart rate

IV. Result

For analysis, the clinical team randomly selected 10 minutes of recording sequences (comprising of 5 individual blocks of 2 minute recordings) from PSG and microbend fiber sensor recordings for each patient while they were in stage I-III sleep. An example of PSG data (2 minute block recording) is shown in Fig.6. The calculations of breathing rate and heart rate from PSG recordings are based on the eye inspection of the breathing waveforms and ECG waveform by two blinded qualified staffs. These PSG results are considered as gold standard results.

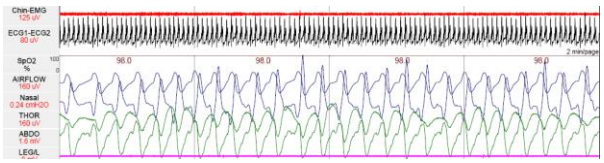
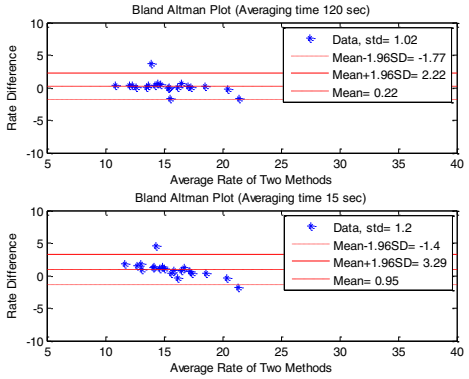
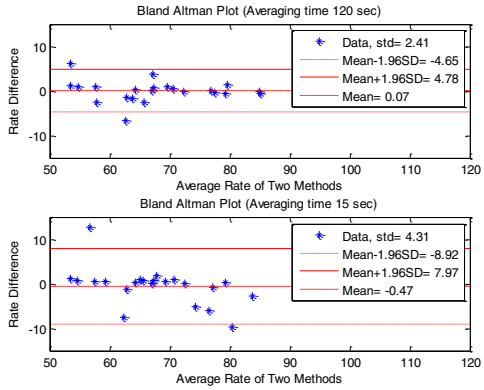


Figure 6. Example of PSG data

Calculations of breathing rate and heart rate from our sensor are based on our algorithm. To compare the differences between the values of our calculation and gold standard PSG results, we use the Bland-Altman plot. Fig.7 shows the Bland-Altman plots for two methods. Our method used two averaging time, 15s and 120s.



(a) Breathing rate results



(b) Heart rate results

Figure 7. Statistical analysis using the Bland-Altman plot

The breathing rate mean difference as measured by two methods, is 0.95bpm and 0.22bpm for averaging time of 15s and 120s, respectively. 95% CI is [0.43bpm, 1.47bpm] for 15s averaging time and [-0.23bpm, 0.67bpm] for 120s averaging time. This suggests that the bias is small and is unlikely to differ by 1bpm for averaging time of 120s. Even for 15s averaging time, the bias is less than 1.5bpm. The averaging time has little effect on the measurement accuracy of breathing rate. The limits of agreement [-1.4bpm, 3.29bpm] and [-1.77, 2.22bpm] correspond to 15s and 120s averaging time, respectively, which are small enough to fairly confident that the new sensor is as good as PSG in the breathing rate measurement. The heart rate mean difference is -0.47bpm and 0.07bpm for averaging time of 15s and 120s. 95% CI is [-2.27bpm, 1.33bpm] for 15s averaging time and [0.94bpm, 1.08bpm] for 120s averaging time. This suggests the bias in the heart rate measurement is small. However, averaging time has significant effect on the limits of agreement when averaging time is 15s. The limits of agreement for heart rate measurement is [-8.92, 7.97bpm]. When the averaging time is 120s, the limits of agreement is [-4.65, 4.78bpm]. Fig.8 shows curve of heart rate standard deviation between two methods as a function of averaging time. Because of significant signal distoration, averaging time of 15s and 30s is not long enough to average out motion noises. However, averaging time of 1 minute gives the similar results as two minutes. This means the one minute averaging time is a fair value for this algorithm.

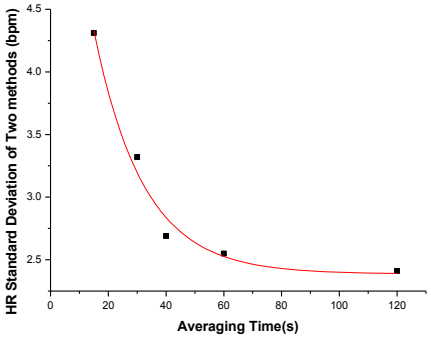


Figure 8. Heart rate standard deviation of two methods as a function of averaging time

V. Conclusion

A new microbend fiber sensor has been proposed and demonstrated for monitoring breathing rate and heart rate during sleep. Our sensor was validated to have an accuracy of 1.06bpm (standard deviation) for breathing rate and 3.32bpm (standard deviation) for heart rate for 30s averaging time based on a 22-patient clinical trial. This level of accuracy is very clinically acceptable in our opinion. Beat to beat evaluation and evaluation on respiration effort can also be done using our sensor although they are not shown in this paper due to limited space. Generally speaking, the current technique requires subject not to move his or her hand, foot and head during measurement. Another limitation of our technique is that our sensor cannot be put under very thick mattress, with a thickness of 28-33cm for example. However, this limitation could be overcome by improving sensor's sensitivity and signal processing algorithm.

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