Low-cost plastic optical fiber sensor embedded in mattress for sleep performance monitoring

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Abstract: In this study, we investigated plastic optical fiber (POF) pressure sensors embedded in mattresses to measure respiration and heart rate for sleep performance monitoring. The signal is amplified in the circuit using a two-stage amplification scheme to collect weak respiration and heart rate signals while an algorithm was designed to obtain respiration and heart rate. We also propose a good reliability cutting-POF technology which can be used to improve pressure sensitivity. The experimental results indicate that the mattress can distinguish four behavioral states related to sleep (on bed, lying, moving and leaving bed) and can detect respiration and heart rate values in different positions and postures. Validation experiments on 10 participants showed that absolute error was less than one breath per minute and two beats per minute, making our approach suitable for household sleeping monitoring.

Keywords: Plastic optical fiber, Mattress sensor, Respiration rate, Heart rate, Sleep behavior monitoring

1. Introduction

Physiological signals can be obtained with fewer limitations on duration and distance via wireless equipment and remote monitoring technology [1] which makes such technology suitable for monitoring health problems of elderly people living alone. Respiration and heart rate are important physiological signals used for monitoring human health [2]. Long-term and continuous measurement of respiration and heart rate is particularly important for the evaluation of some diseases [3] such as coronary heart disease [4], arrhythmia [5], fatigue [6] and apnea [7]. At present, several methods have been proposed to measure respiration and heart rate, such as electrocardiogram (EEG) [8] and photoelectric volume pulse wave (PPG) [9], which have already enjoyed commercial success. However, both methods require wires to be connected to the human body, which are less suitable for long-term scenarios. To solve this problem, several non-inductive measurement approaches have been proposed, such as piezoelectric thin film [10], electromagnetic wave reflection [11], acoustic sensor [12] and other electrical sensors. However, these types of sensors are easily affected by electromagnetic interference and are not suitable for measuring respiration and heart rate in the magnetic resonance imaging (MRI) environment.

Optical fiber sensors have the advantages of immunity to electromagnetic interference [13], biocompatibility and chemical stability [14], and are widely used in various fields to replace electrical sensors. Nowadays, sensors based on silica optical fibers are under extensive research, some of which have already gone into commercial applications. Plastic optical fiber (POF) has attracted the attention of medical and health researchers due to higher flexibility, sensitivity [15, 16] and lower cost [17] in sensor applications. The POF sensors have already been applied to structural health monitoring [19], rehabilitation treatment [20] groundwater level monitoring [21], temperature monitoring [22], pulse wave monitoring [23], respiration and heart rate monitoring [24].

Several approaches have been proposed to measure respiration and heart rate using optical fiber sensors, such as fiber Bragg grating (FBG) sensors [25-29], Mach-Zehnder interferometer (MZI) sensors [2] and light intensity modulation POF sensors [30 -32]. FBG and MZI require expensive production machinery and testing instruments and are not suitable for family guardianship due to high cost. However, POF sensors based on light intensity modulation are suitable for large-scale production [33] and health monitoring at low cost.

Chen, et al proposed a kind of multimode optical fiber sensor based on micro-bending, which can be implanted in mattresses and cushions [31]. However, serious light loss is suffered during transmission which leads to high power requirements for light sources. *Koyama, et al* designed a single-mode heterogeneous core optical fiber to measure respiration and heart rate, embedded in the textile and placed in the left side of the chest, close to the heart [32]. A low-cost plastic optical fiber sensor embedded in a mattress is reported in [30], which describes a pressure rectangular array designed to detect pressure. Furthermore, one kind of wearable device has been designed based on sensitive elastic bands with plastic optical fiber [3] although it is uncomfortable to be worn for long periods and not suitable for usage overnight.

In this paper, we present a plastic optical fiber sensor embedded in mattress for sleep monitoring, which measures respiration (breathing) rate (BR) and heart rate (HR) non-invasively and non-inductively. The mattress can distinguish four behavioral states (not in bed, lying, moving, and leaving bed), and also measure respiration and heart rate in different positions and postures, making it suitable for sleep monitoring. This paper is divided into the following sections after introduction. The second section introduces the design and implementation of the sensor, including plastic optical fiber processing, circuit design and algorithm design. In the third section, we describe experiments which verify the performance of the smart mattress, the measurement of behavioral state, respiration and heart rate in different positions and different postures. Finally, the last section presents conclusions and avenues for future work.

2. Sensor design and implementation

2.1. Overall system design

The overall system design is based on light intensity modulation technology as describes the block diagram shown in Figure 1. The system can be divided into three parts: optical fiber mattress design, circuit design and data processing.

The light-emitting diode (LED) driving circuit is controlled by the microcomputer (MCU). The LED output light is launched into one end of the optical fiber embedded in the mattress, and the other end of the optical fiber is connected to the photodiode (PD). At the same time, the signal conditioning circuit collect the change of voltage due to light intensity in the optical fiber, and sends the values of the voltage to the host computer via Bluetooth for data analysis. Finally, the measurement of respiration and heart rate is completed by an algorithm running on the host computer.

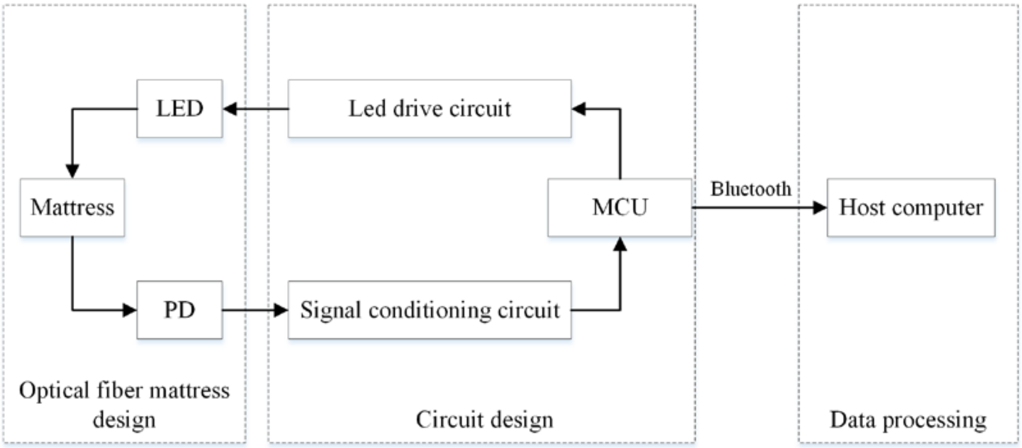

Fig. 1. System structure block diagram.

2.2. Sensor design

Common plastic optical fiber, with cladding diameter 1000 µm, core diameter 980 µm and outer diameter 2200µm, is embedded in the mattress. Due to the pressure of the body, the optical fiber will bend and deform slightly, thus the intensity of light traveling through the fiber varies, and the respiration and heart rate signals can be measured by detecting the change of light intensity. To enhance sensitivity [3], we cut the fiber cladding and part of the core (with a blade) to increase the leakage (the cutting accuracy around 0.02 mm), in order to enhance the light coupling loss between the two faces of the POF under pressure as shown in Figure 2.

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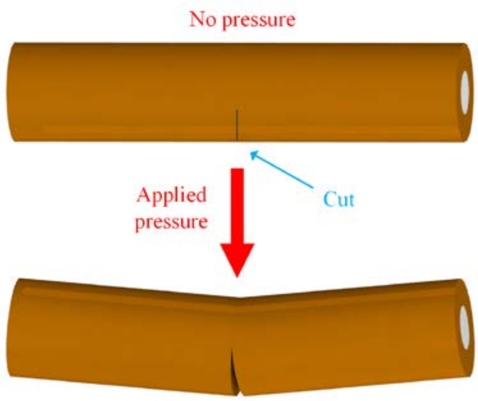
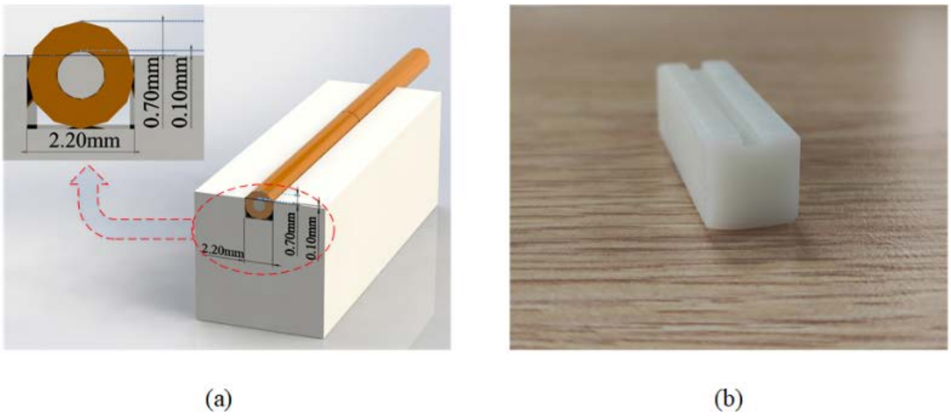


Fig. 2. Schematic diagram of optical fiber cutting.



(a) (b)

Fig. 3. (a) Model of the 3D cutting device, inset is partially enlarged; (b) Photo of cutting device.

To ensure uniform depth of cutting (0.7 mm), a 3D printed mold is designed as shown in Figure 3. Then the plastic optical fiber sensor embedded in the mattress as shown in Figure 4. The distance between cut points is 10 cm and the total length is 90 cm.

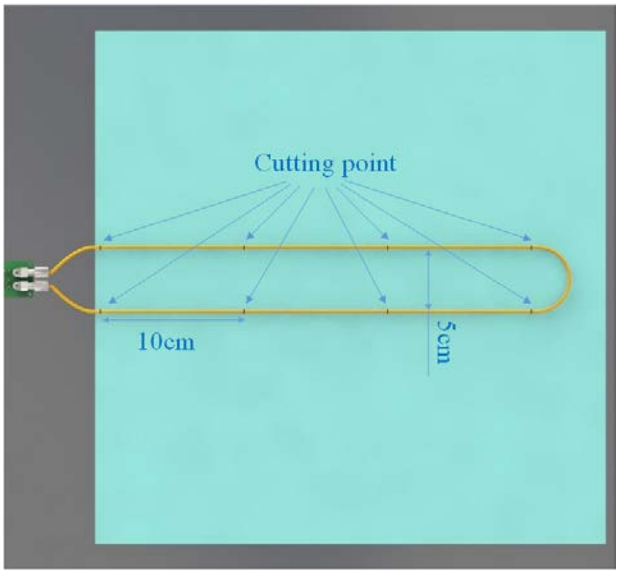


Fig. 4. Schematic diagram of plastic optical fiber, with cut points, embedded in the mattress.

2.3. Circuit design

The hardware circuit is divided into three modules: optical emission, optical reception, and signal processing. The main function of the optical emission module is to transmit light of fixed intensity. The light source is an IF E96E (Industrial Fiber Optics, USA), a visible red LED emitting at 645nm and housed in a "connector-less" style plastic fiber optical package. As shown in Figure 5, we use a low-dropout regulator (LDO) to drive the LED at constant current, the output current calculation equation is:

$$I_A = \frac{U_{FB}}{R_1} \quad (1)$$

Here, U_{FB} = 1.2246 V (obtained from the data sheet of TPS79101DBVR), if I_A = 20 mA, then R_1 = 62 Ω .

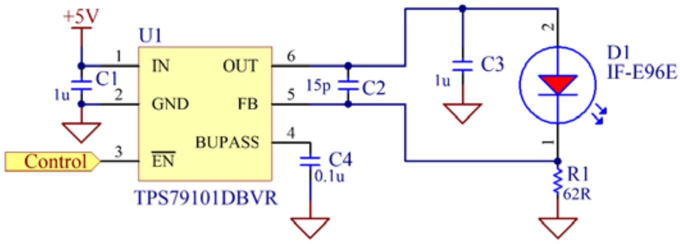


Fig. 5. LED drive circuit.

An optical reception module, containing an IF-91B PD (Industrial Fiber Optics, USA), converts the change in light intensity into a change in current. Since signals obtained from POF are weak, the collected signal is processed in a two-stage amplification. The front stage of the circuit includes a trans-impedance amplifier (TIA) and converts the current into voltage, the output voltage is U_1 =- I_{PD}×R_2, C8 and R5 form a high-pass filter with a cutoff frequency of 0.1 Hz. The TIA is followed by an instrument amplifier (AD8422ARZ-R7, ADI, USA), the magnification is G = 1+19.8K/R4 = 100 (the circuit diagram is shown in Figure 6)

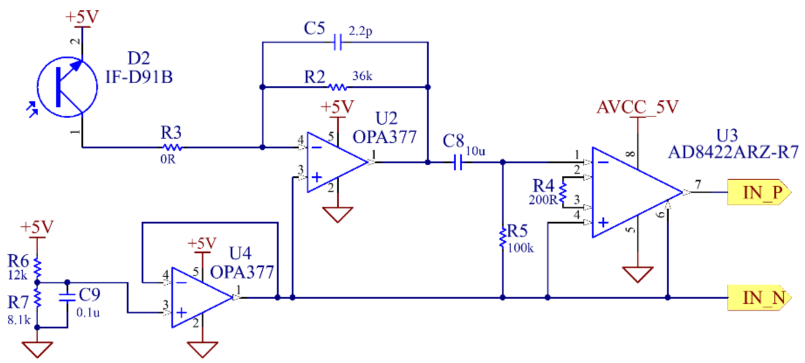


Fig. 6. Photoelectric signal amplification circuit.

The signal processing module collects the analog signal and converts it into digital data by a digital-to-analog converter (DAC) before a microcomputer (MCU) sends the data to the host computer via the Bluetooth protocol. Respiratory rate and heart rate are calculated by the algorithm running on the host computer, as described below.

2.4. Data processing

Normally, people breathe 12 to 20 times per minute (0.20 to 0.33 Hz), and normal heart rate is 50 to 100 times per minute, (0.83 to 1.67 Hz). Since the frequency bands of these two signals do not overlap, we can look for peaks in the spectrum at their corresponding bands. The processing steps for respiration and heart rate is shown in Figure 7.

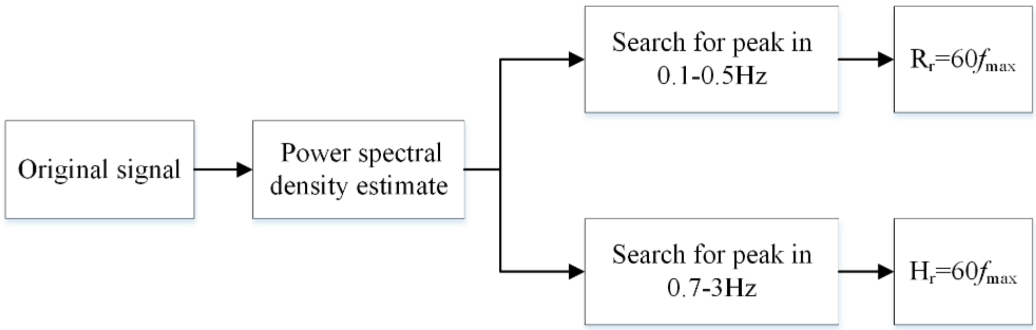


Fig. 7. The process of calculating respiration and HR.

We use the Welch method to estimate the power spectral density, add windows to the data, segment the data according to the length of the window, and then calculate the power spectrum for each part of the data, finally stack and average the data. The power spectrum density estimated at every 30 seconds with the sampling frequency is 1000 Hz, 30000 points used in this sample. In other words, the window length of each data segment M set to 3000, there are L = 10 segments of data, and the power spectrum of each data segment can be estimated by equations 2 and 3:

$$\mathrm{P}_i(\mathrm{f}) = \frac{1}{MU}\left|\sum_{n=0}^{M-1} x_i(n)w(n)e^{-i2\pi fn}\right| \tag{2}$$

$$U = \frac{1}{M}\sum_{\mathrm{n}=0}^{M-1} w^2(\mathrm{n}) \tag{3}$$

Here, U is the normalization factor, and $w(n)$ is the Hamming window function, $x_i(n)$ is the signal of each segment, and the power spectrum of the L segments averaged to obtain the power spectrum estimation of each package of the data as:

$$\mathrm{P}(\mathrm{f}) = \frac{1}{L}\sum_{i=1}^{L} P_i(f) = \frac{1}{MUL}\sum_{i=1}^{L}\left|\sum_{n=0}^{M-1} x_i(n)w(n)e^{-i2\pi fn}\right| \tag{4}$$

The number of breaths and heartbeats are calculated by:

$$R_r = 60 \times \max[PSD(\text{BR})] \tag{5}$$

$$H_r = 60 \times \max[PSD(\text{HR})] \tag{6}$$

Where max [PSD(BR)] is the frequency corresponding to the peak in the power spectrum in the frequency band of 0.1-0.5Hz, and max [PSD(HR)] corresponds to the peak in the power spectrum in the frequency band of 0.7-3 Hz.

3. Experiments and results

3.1. Respiration and heart rate signal processing

Figure 8(a) shows 30 seconds of data time series collected using our mattress near the heart position. Since body motion caused by respiration is much larger than that from heartbeat, the heartbeat signal is submerged in the respiratory signal. Looking at the power spectral density (Figure 8(b)), there is an obvious peak in the respiratory and heartbeat bands. The separated respiratory and heartbeat signals are shown in figure 8(c) and 8(d), respectively. Note the filter has an initial setting time of ~2 seconds.

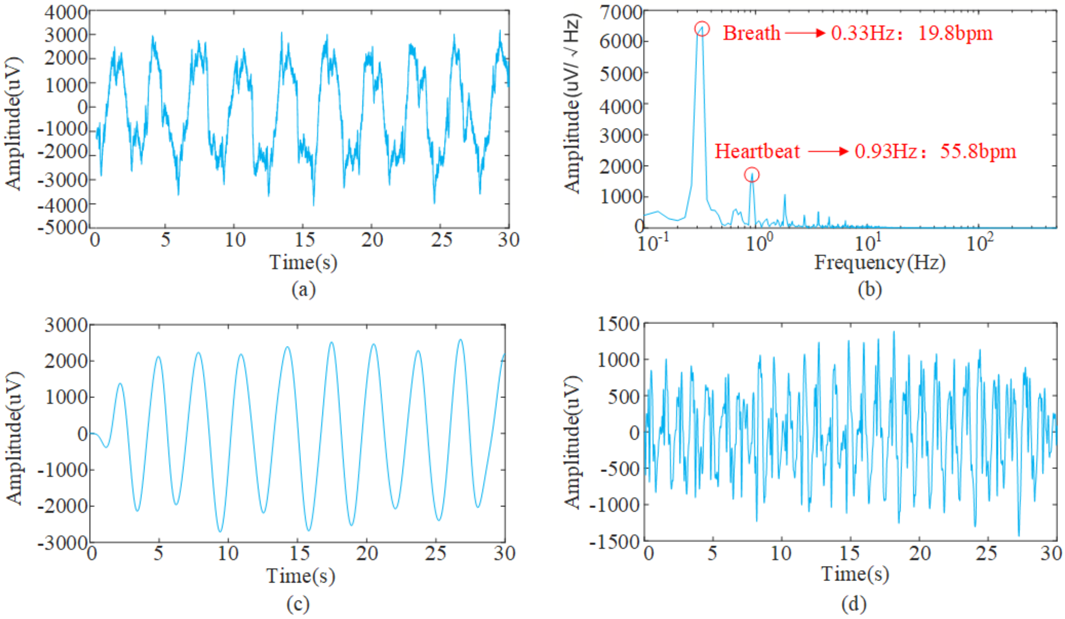


Fig. 8. (a) Original signal; (b) Power spectral density of the original signal; (c) Respiratory waveform; (d) Heartbeat waveform.

3.2. Sleep behavioral monitoring

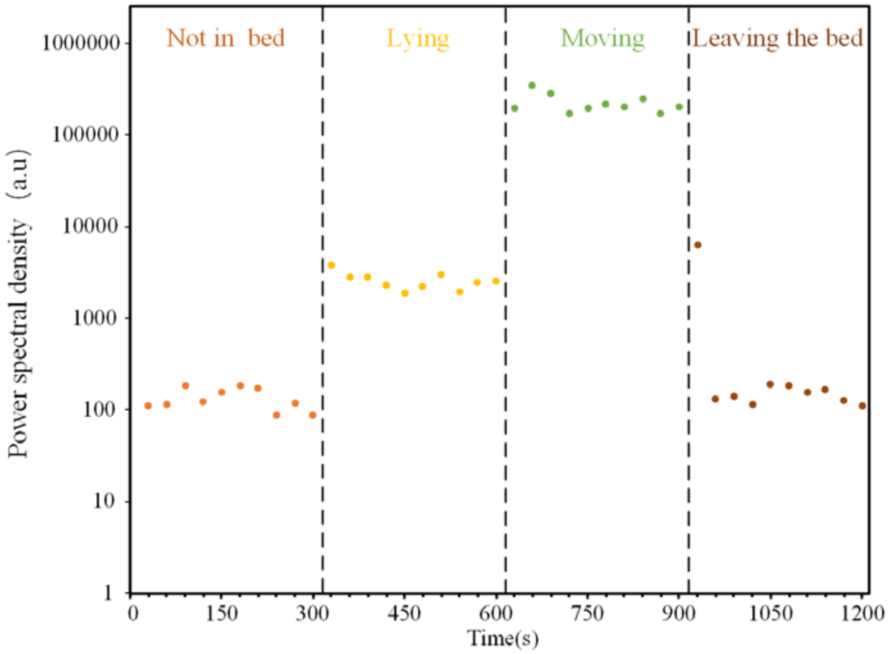


Fig. 9. Power spectral density of four behavioral stages for sleep monitoring.

We classify the behaviors of people in bed into four stages: not in bed, lying, moving and leaving the bed. These behaviors are distinguished according to the energy of the power spectral density. In our experiments, the participant was required to stay for five minutes in each state. We measure the highest point of energy spectrum every thirty seconds. Data for one example participant is shown in Figure 9. During not lying on the mattress, the spectral energy is low, and most of the energy is generated by power line frequency coupling and noise in the circuit. When the person lays still on the mattress, the spectral energy is increased by an order of magnitude, mainly caused by respiration and heartbeat. When the person moves in bed, the spectral energy is one order of magnitude higher than lying on the bed, due to gross bodily movements. When the person leaves the mattress, the spectral energy decreases rapidly and returns to the same level as not in bed, which indicate the situation of not in bed similar as lying on the bed, we also record one situation as the process of leaving bed, as shown in Figure 9.

3.3. Accuracy analysis

To confirm the accuracy of respiration and heart rate measurement, the same mattress was used to collect data for 30 s, during which respiration was manually counted and HR was measured with a commercial oximeter (YH303, Yuwell, China). A total of 10 participants were recruited and data was collected from each participant three times (30 s trials) (Table 1). The maximum absolute errors of BR and HR are less than 1 and 2 events per minute, respectively, and the maximum relative errors are 6.7% and 2.4%, respectively. These error values are acceptable when compared with physiological testing equipment [2].

Table 1. Comparison of measurements between optical fiber mattress and manual observation/commercial oximeter

Participant	Gender	Body masst (kg)	POF Sensor		Reference	
Number			BR±abs error	HR±abs error	BR±abs error	HR±abs error
1	Male	50	18.0±1.6	59.3±1.9	18.0±1.6	59.7±2.1
2	Male	69	14.0±1.6	65.3±3.4	15.0±0.8	67.0±1.6
3	Male	75	13.3±0.9	87.3±1.0	13.7±0.5	87.0±2.2
4	Male	70	12.0±0	68.0±0	12.7±0.9	69.0±0.8
5	Male	59	10.7±0.9	72.7±0.9	10.7±0.5	72.3±1.9
6	Male	67	31.3±0.9	61.3±2.5	32.0±0	62.3±0.9
7	Male	68	11.3±0.9	84.0±1.6	12.0±0.8	84.3±0.5
8	Female	54	25.3±1.9	69.3±1.0	25.3±1.9	69.7±0.5
9	Female	57	15.3±0.9	78.3±1.3	15.3±0.9	78.3±1.3
10	Female	55	12.0±1.6	42.7±0.9	12.7±0.9	42.3±2.1

3.4. Reliability analysis

When a person is sleeping, the position and posture in bed are different each time, so two more experiments were done to verify the performance of the mattress. The first experiment aimed to verify the influence of different positions on the mattress always lying on his/her back.

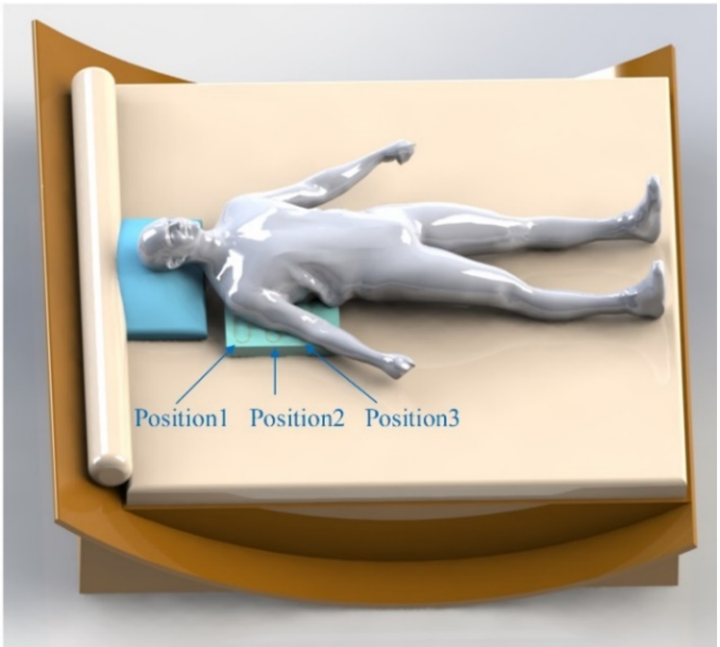


Fig. 10. Schematic diagram of measuring position.

A participant was asked to lie in different positions on the mattress, the heart is usually located next to the fifth rib, so fibers are placed on the second, fifth, and seventh ribs, as shown in Figure 10. Data from each position was collected for 30 s, as shown in Figures 11(a) and (b), respiratory signals are clearly seen in each position. For heartbeat, position 2 and position 3 are closest to the heart, and data from those two positions are more obvious. Position 1 is far away from the heart, and therefore, in this case, heart rate signals are weaker. The BR and HR measured at these locations (position 1, position 2, position 3) are close to normal values.

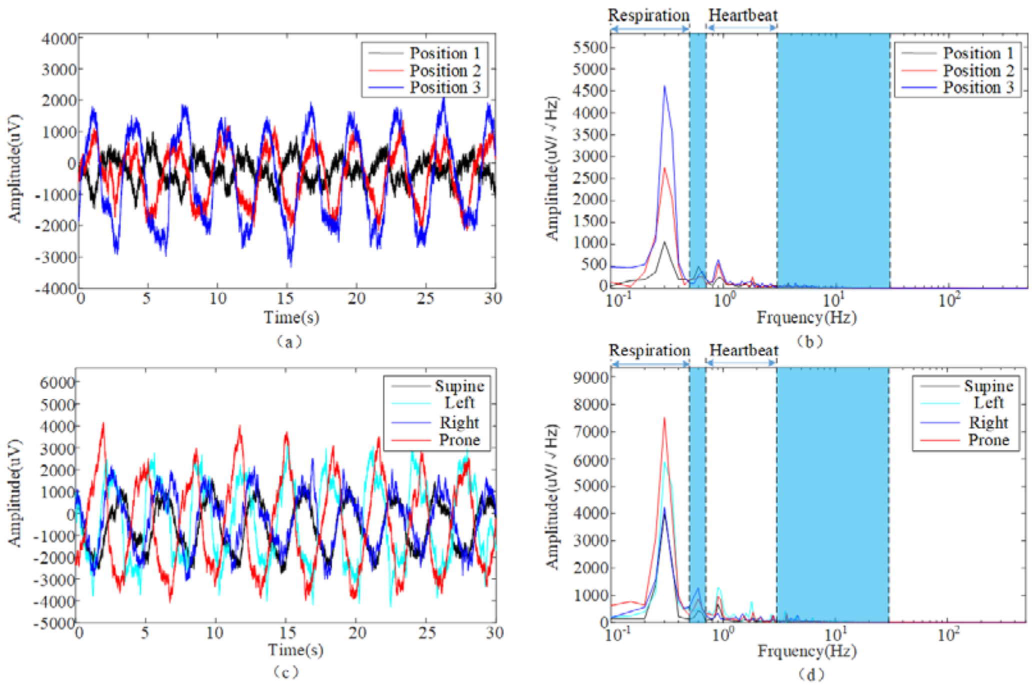


Fig. 11. (a) Time series of different recording locations; (b) Power spectral density at different recording locations; (c) Time series of different postures; (d) Power spectral density of different postures.

The second experiment aimed to verify the effect of different sleeping postures on the sensor. A participant was asked to lie on the mattress in a supine, left, right or prone posture. As shown in Figure 11(c) and (d), We collected data for 30 s from each posture, the respiratory signal is obvious in every position. As for the heartbeat signal, due to asymmetry of the heart location in the human body, the signal was weakest for the lying on the right-side (contact) position, and the supine and left posture shown more obvious heartbeat signal performance. These two experiments indicate that the smart mattress presented in this work can be used for sleep performance monitoring with different sleeping postures, which also have some applications for monitoring diseases such as apnea and hyperventilation.

4. Conclusion

In this study, we propose a smart mattress based on plastic optical fiber sensors that can measure respiration and heart rate by sensing slight motions of the body. The system uses a commercial, low-cost plastic optical fiber, which is embedded in the mattress. Due to the simplicity of the sensor, the system is suitable for large-scale production and deployment as an overnight sleep monitoring method. The sensor does not only distinguish four behavioral states related to sleep, but also measures respiration and heart rate in different positions and postures. Our validation experiment with ten participants showed that the absolute errors in respiration and heart rate are less than one breath per minute and two beats per minute, respectively, which is acceptable for overnight sleep monitoring and provides a feasible approach for in-home monitoring as well as in-hospital monitoring. In future work, the algorithm can be improved to mitigate the influence of movement. Our approach can be extended to detect respiration and heart rate of people sitting in chairs, providing monitoring for long-duration sitting.

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

Acknowledgements

National Natural Science Foundation of China (62003046); National Defense Basic Scientific Research Program of China (JCKY2018110B011); The Spanish Ministerio de Ciencia, Innovación y Universidades RTI2018-101658-B-I00 FOCAL Project; Guangdong Recruitment Program of Foreign Experts (2020A1414010393); C. Marques acknowledges Fundação para a Ciência e a Tecnologia (FCT) through the CEECIND/00034/2018 (iFish project) and this work was developed within the scope of the project i3N, UIDB/50025/2020 &UIDP/50025/2020, financed by national funds through the FCT/MEC.

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