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The Modeling of the Differential Measurement of Air Pressure for Non-intrusive Sleep Monitoring Sensor System

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Abstract: The respiratory and heart beat signals are the fundamental physiological signals for sleep monitoring in the home. Using the air mattress sensor system, the respiration and heart beat movements can be measured without any harness or sensor on the subject's body which makes long term measurement difficult and troublesome. The differential measurement technique between two air cells is adopted to enhance the sensitivity. The concept of the balancing tube between two air cells is suggested to increase the robustness against postural changes during the measurement period. With this balancing tube, the meaningful frequency range could be selected by the pneumatic filter method. The mathematical model for the air mattress and balancing tube was suggested and the validation experiments were performed for step and sinusoidal input. The results show that the balancing tube can eliminate the low frequency component between two cells effectively. This technique was applied to measure the respiration and heart beat on the bed, which shows the potential applications for sleep monitoring device in home. With the analysis of the waveform, respiration intervals and heart beat intervals were calculated and compared with the signal from conventional methods. The results show that the measurement from air mattress with balancing tube can be used for monitoring respiration and heart beat in various situations.

Key words: Respiration, Heart beat, Air mattress, Pneumatic frequency filter, Sleep monitoring, Non-intrusive

INTRODUCTION

As the obese population is growing rapidly, the need of sleep monitoring device is also increasing. Even though polysomnography (PSG) is the standard sleep analysis method in hospitals, it is too complex to be used in the home. American Sleep Disorder Association summarized the systemic classification of devices for sleep analysis in the home into four categories and reviewed its sensitivity and specificity [1]. In this review, level III and level IV might be used as portable devices, but these measurement systems need electrodes or sensors on the body surface [2]. This constrained measurement makes inconvenience for daily monitoring. For a certain monitoring device for daily life, there should be the non-intrusive measurement technology not to make any inconvenience for one's daily life.

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The respiration, heart beat, and the body movements are the fundamental physiological signals that are needed for sleep analysis. For the purpose of non-intrusive monitoring, there have been several trials until now. Tanaka [3] suggested a phonocardiogram sensor on the bed and Jacobs [4] used PolyVyniliDeneFluoride (PVDF) sensors in the form of a 2-dimensional array, which might be expensive but accurate. Spillman [5] used multi-modal optic sensors. Static Charge Sensitive Bed (SCSB) was also introduced by measuring the changes of static charge by the movement of respiration [6]. Even with these trials, there is no leading method that is stable, accurate, and economic until now. Hernandez [7], Chow [8] and Watanabe [9] used an air mattress system under the body as a sensor for the respiration movements and heart beat. The respiration and heart beat movement add pressure to the air cell, which changes the pressure in the air mattress. In the measurement using air mattress, the most difficult point is to pick up the small signals of the respiration and heart beat movements from the large signals those are associated with motion artifacts and body weight. Normal air pressure sensor has its limited operating range and the sensitivity. The accuracy is defined by the percentage of its full dynamic range of the pressure sensor. In order to solve this problem, Hernandez [7] added jugs and balloons between the air mattress and sensor to block direct current (DC) component. Chow [8] measured the differential pressure between the two air cells. Chow connected these two cells with tube and he opened the valve between the two compartments for 5 seconds to equalize the pressures in each cell when the sensor detected an excessive pressure difference. Watanabe [9] used super-sensitive pressure sensor with condenser microphone. As stated above, to measure the respiratory and heart beat signals using air mattress, there should be stable and robust measurement mechanism against the postural changes during sleep. To accomplish the non-intrusive monitoring of respiratory and heart beat, a novel air mattress sensor system using balancing tube was suggested by the authors [10]. In this paper, the mathematical model was suggested and validation experiments were performed. Also recordings from human body were done and showed its potentials to the application of sleep monitoring system in the home.

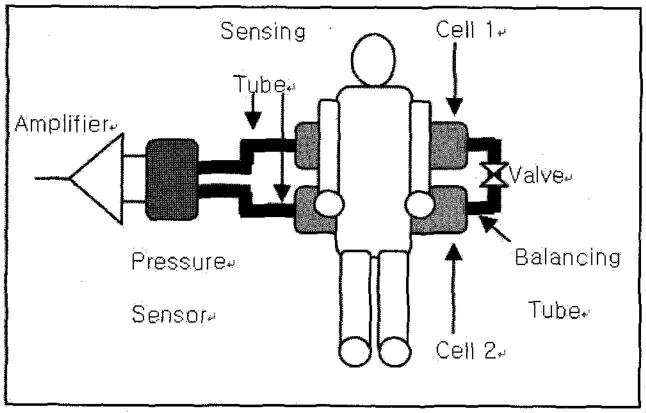


Fig. 1. The air mattress sensor system with balancing tube

METHODS

Measurement Principle

As shown in Fig. 1, when a subject is lying on the air mattress, any movement including respiratory effort, heart beating and postural changes make change the air pressure inside air cell. To enhance the sensitivity, the differential measurement technique was used as shown in Fig. 1. The pressure by the body weight can be regarded as the static pressure during the measurement. By measuring the differential pressure in remote device with tubes (sensing tubes in Fig. 1), we can record the respiratory movement in pneumatic method. One of the advantages of the differential measurement is to reject the common mode artifacts that are not related with respiratory and heart beat movement like body weight and environmental noise.

The respiratory and heart beat movement can be described as the linear combination of sinusoidal waves with small amplitudes. And the postural changes during sleep can be modeled as combination of step function if we neglect the transient time. The amplitude of the step function is quite large compared to the heart beat or respiratory signals. Before the air pressure arrives to the limit (dynamic range) of the pressure sensor, it is needed to filter out these step functions. Previous studies [7-9] suggest focused on solving this problem. In this paper, I suggest the balancing tube connection between two cells to solve this problem. The balancing tube has high resistance and small capacitance. It is able to block the air flow for high frequency components between two cells. For the DC component (constant component for a certain time of period), it passes in both directions with time constant that can be controlled by the controlling of opening area. With this balancing tube, after the postural changes, the operating point of differential pressure returns around zero level automatically.

Modeling of Two Air Cells

Fig. 2 shows a schematic diagram of the sensor system with air pressure (p), flow (q), mass (m), air resistance(R), and air capacitance(C) of the sensing tubes and the balancing tube. To analyze this system step by step, at first let's assume that there is no balancing tube (black) between the two cells in Fig. 2. Using pneumatic system modeling [11] for the air resistance and the air capacitance of the sensing tube, Eq. (1) can be used to represent the pneumatic measurement system.

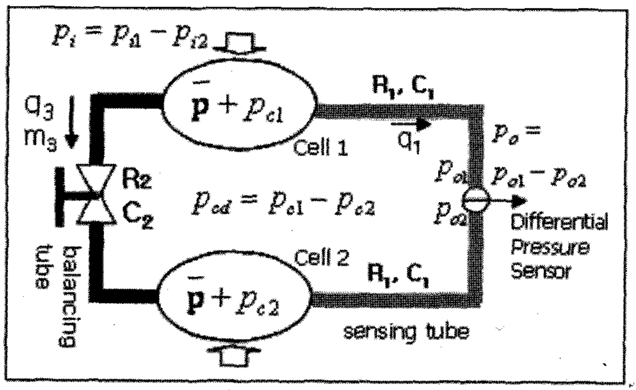


Fig. 2. Schematic illustration of the modeling of the two cells and sensing tubes with the balancing tube. R_1, C_1 are the resistance and capacitance of the sensing tube and R_2, C_2 are the resistance and capacitance of the balancing tube.

$$R_1 = \frac{p_{i1} - p_{o1}}{q_1},\ C_1 = \frac{dm_1}{dp_{o1}} = \frac{q_1 dt}{dp_{o1}} = \frac{(p_{i1} - p_{o1})dt}{R_1 dp_{o1}}$$

$$R_1 C_1 \frac{dp_{o1}}{dt} + p_{o1} = p_{i1} = p_{c1} \tag{1}$$

where

q_1: mass flow rate through the upper and lower tube

m_1: mass of air passing through the sensing tube and balancing tube

p_{i1}, p_{i2}: pressure that is forced to the air cell 1 and 2, here $p_{i1} = p_{c1}, p_{i2} = p_{c2}$

p_{c1}, p_{c2}: pressure inside air cell 1 and 2, where $p_{cd} = p_{c1} - p_{c2}$

p_{o1}, p_{o2}: pressure that is delivered through the upper and lower sensing tube to the pressure sensor

The initial pressure in each cell is eliminated since only the differential pressure was measured. Eq. (1) indicates that p_{o1} is the output of the low pass filter, with a time constant of R_1 and C_1 for the input of p_{c1}. For the other sensing tube, the equation can be written as shown in Eq. (2). By subtracting Eq. (2) from Eq. (1), the transfer relationship between the difference in pressure of p_{cd} and p_o is defined in Eq. (3). The output signal(p_o) is the low pass filtered signal of p_{cd} if the two tubes are of the same size and material.

$$R_1 C_1 \frac{dp_{o2}}{dt} + p_{o2} = p_{c2} \tag{2}$$

$$R_1 C_1 \frac{d(p_{o1} - p_{o2})}{dt} + (p_{o1} - p_{o2}) = p_{c1} - p_{c2}$$

$$R_1 C_1 \frac{dp_o}{dt} + p_o = p_{cd} \tag{3}$$

By connecting the balancing tube, represented by the black line in Fig. 2, the flow based on the pressure difference between air cells 1 and 2 is considered as Eq. (4).

$$R_2 = \frac{p_{c1} - p_{c2}}{q_3},\ C_2 = \frac{dm_3}{dp_{c2}} = \frac{q_3 dt}{dp_{c2}} = \frac{(p_{c1} - p_{c2})dt}{R_2 dp_{c2}}$$

$$R_2 C_2 \frac{dp_{c2}}{dt} + p_{c2} = p_{c1} \tag{4}$$

where

q3: mass flow rate through the balancing tube

m3: mass flow rate through the balancing tube

By applying Eq. (4) and Eq. (3) sequentially, the final output of P_0 can be obtained for any input of pi1 and pi2. The summarized procedure determining the response signal from any input pressure is as following:

STEP 1: For any outside pressure input [pi1, pi2], let pc1=pi1' = pi1 - pi2, pi2'=0

STEP 2: Using the Eq. (4), solve for pc2 and get pcd (= pc2 - pi1)

STEP 3: Solve for P_0 through Eq. (3) with the pcd

By the differential measurement, the effect of low pass filter in balancing tube makes the effect of high pass filter in P_0.

Response for Step Function Input

The postural change during sleep can be regarded as a combination of step functions $u(t)$. For the step input, by following the above procedures, the response can be calculated as Eq. (5).

$$p_{i1} = u(t),\quad p_{i2} = 0,$$
$$p_{c1} = u(t),$$
$$p_o = \tau_2 \frac{e^{-\frac{t}{\tau 2}} - e^{-\frac{t}{\tau 1}}}{\tau_2 - \tau_1} \tag{5}$$
$$where\ \tau_1 = R_1 C_1 \quad and \quad \tau_2 = R_2 C_2$$

In Eq. (5), τ_1 is the time constant of the low pass filter effect by the sensing tube. When we choose proper diameter and length for sensing tube not to distort the signal, τ_1 goes to zero. In Eq. (5), τ_2 is the time constant by the high pass filter effect by the balancing tube. By changing the opening area of the valve in balancing tube, we can adjust R_2 in Eq. (5).

Fig. 3 shows its plot when τ_2 = 1.0, τ_1= 0.01. Because the air resistance in the tube is proportional to $(1/\text{radius})^4$, air resistance in balancing tube of R_2 is higher than 100 times of the air resistance of sensing tube R_1 (i.e. $\tau_2 >> \tau_1$). By neglecting τ_1, Eq. (5) becomes

$$p_o \cong u(t) e^{-t/\tau_2} \tag{6}$$

This equation is the step response of 1st order RC high pass filter in electrical circuit with capacitor and resistor. Based on the property of 1st order RC high pass filter, time constant τ defines its frequency response as shown in Fig. 4. When air resistance increased, i.e., time constant is increased, the cut off frequency of the high pass filter is decreasing. The system transfer function on frequency is following.

$$G(jw)=\frac{jwRC}{1+jwRC} \tag{7}$$

$$f_c=\frac{1}{2\pi}\frac{1}{RC}\ [Hz] \tag{8}$$

By applying the step input to the system, and measuring the response, we can measure time constant R_2C_2 value i.e., τ_2. Using the above equation, we can define the cut off frequency of the high pass filter of air mattress sensor system. The experimental results are described in next section. To summarize, we can control the opening area of the balancing tube, i.e. air resistance. With adjusting $\tau_2(R_2C_2)$, we can control the cut off frequency of the pneumatic high pass filter.

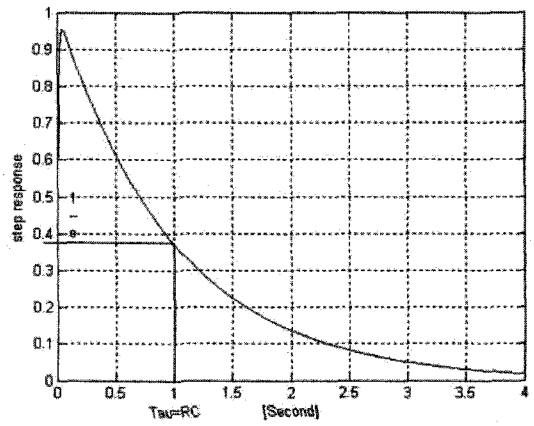


Fig. 3. Plot of step response

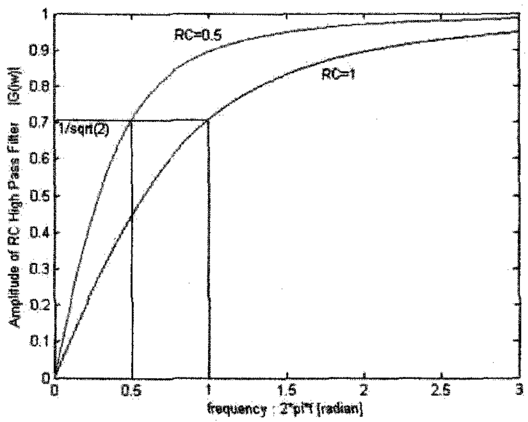


Fig. 4. Amplitude of 1st order RC High Pass Filter according to frequency and time constant.

Response for Sinusoidal Function Input

The response for the sinusoidal input of $A_0\sin(\omega t)$ can be calculated by following the procedures in section 2.2.

$$p_o=\frac{A_0\tau_2\omega((1-\tau_1\tau_2\omega^2)\cos(\omega t)+(\tau_1+\tau_2)\omega\sin(\omega t))}{(1+\tau_1{}^2\omega^2)(1+\tau_2{}^2\omega^2)} \tag{9}$$

The resultant amplitude of p_o is plotted in Fig. 5 (b). It shows again the high pass filter property that was implemented with balancing tube and differential measurement. In order to produce the sinusoidal input force on the air cell, the mechanism in Fig. 5 (a) which converts the rotation movement into sinusoidal movement along the horizontal axis was constructed. This stimulator provides the sinusoidal pressure to one of the air cells. The plot of Eq. (9) is shown in Fig. 5(b), and the experimental results are shown in Fig. 5(b) with '*' marks.

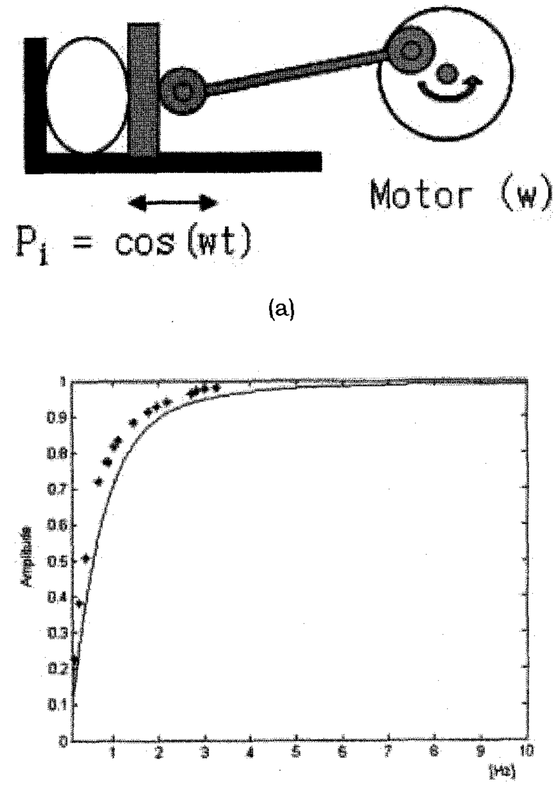


Fig. 5. plot of frequency response and experimental result
(a) The schematic diagram of the sinusoidal stimulator.
(b) The plot simulation where τ_1=0.01, τ_2=1.0 and A_0=1 ('-') and experimental results('*')

Analysis Algorithm

To extract the quantitative information from air mattress system, we have to calculate the heart beat interval and respiration interval. In this study, Pan&Tompkins algorithm [12] to detect QRS complex from ECG was used with modifications. The respiration and heart beat signals from air mattress have quite different properties from those of ECG. So modification is required to detect certain feature points from the measured waveform. Fig. 6 shows the algorithm in flow chart.

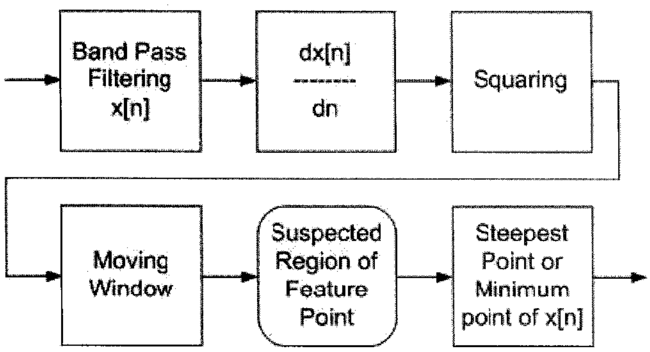
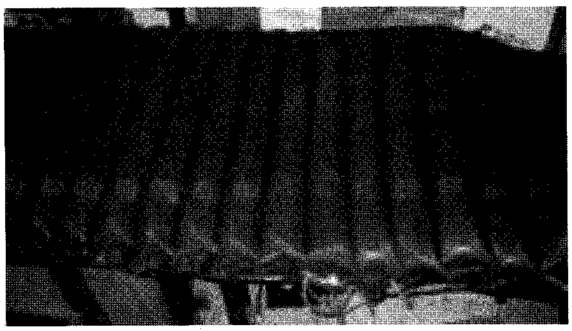


Fig. 6. The flow chart to localize the feature points in semi-periodic waveform

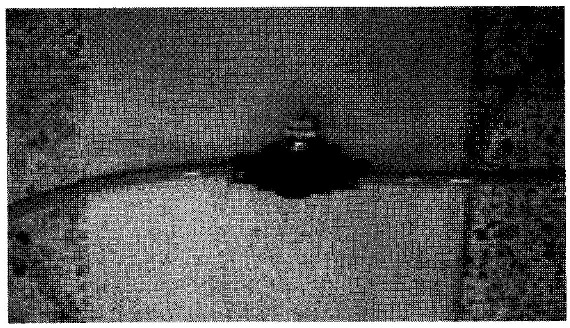
The power spectrum of respiration signal from air mattress is mainly distributed in lower frequency band (< 1Hz). And the largest peak in one heart beat complex has the frequency from 1 to 15 Hz. As a pre-processing, there should be band pass filter for each purpose. In this study, 4^{th} order Butterworth filter was used. Then the filtered signal is derived and squared to enhance respiration effort and heart beat movement. The moving average window technique was used to produce the smooth shape for each cycle. For respiration, the width of moving window is 200ms, and width of heart beat is 100 ms. By applying threshold, the suspected interval where the biggest peaks are located can be found. In next chapter, the analysis results were shown for this algorithm.

Implementation Into Measurement System

As shown in Fig. 7, the air mattress system with 20 air cells was constructed. Even though the patterns of waveforms are changing according to the cells where the body is contacting, we can find the information of respiration and heart beating from any pair of cells if the cells are located between the neck and the waist of the subject. In this experiment, the two cells that are located on the backside of chest and abdominal region were chosen where we could obtain the large respiration signal than the other locations. The other air cells are to support the body in comfortable way. The cells were in the form of a cylinder with the diameter of 110mm, which was made from a polyurethane sheet, and air was filled at 10 kPa in the beginning. We connected the air cells and sensors that are assembled on the circuit board with the silicone tube with the inner diameter of 4mm, an outer diameter of 8mm, 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 is a rotational valve that can control the open area precisely (Fig. 7(b)). NPC-1210 (100mVolt/5psi) from Luca NovaSensor® is used for pressure transducer and the signal was amplified 30,000 times in order to detect the small signals from respiration and heart beat. The analog signal was sampled at 200 Hz and digitized into 16 bits with Biopac Inc. The signals were measured from a human subject who lied on the air mattress. Together with the air mattress sensor, the traditional electrocardiography (Biopac Inc.) and nasal airflow (Biopac Inc.) and respiration effort (resistive type, Biopac Inc.) were measured simultaneously to compare with standard method.



(a)



(b)

Fig. 7. The sensor system using air mattress
(a) Air cells and air mattress (b) Balancing tube

RESULTS

Validation of Modeling

To record the step response, object of 1 kg was used. The static pressure was 2.14 kPa and amplifier gets 12 Pa/1volt. By opening the valve in balancing tube step by step, the responses in Fig. 9 were measured. As stated in this section, the more when the valve opened, the faster it arrives to zero level, i.e. the shorter the time constant (τ) becomes.

To find out τ in each step from the waves in Fig. 8, the peak level was used in approximation. The table 1 shows the τ in each trial and cut off frequency by Eq. (8).

Table 1. Time constant of step response

opening area	τ(RC) [second]	cut off freq. [Hz]
0%		N.A.
10%	> 10	< 0.01
20%	3.910	0.041
30%	1.625	0.098
40%	1.315	0.121
50%	0.985	0.162
60%	0.720	0.221
70%	0.500	0.318
80%	0.375	0.424
90%	0.305	0.522

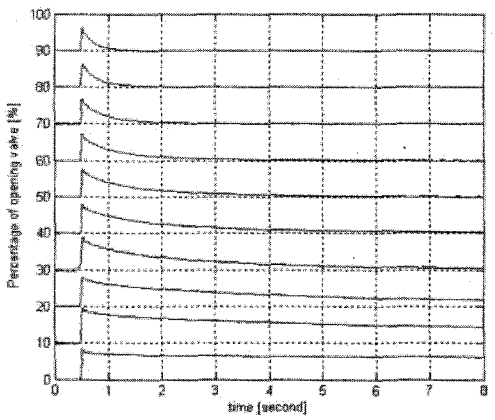


Fig. 8. The step response experimental result

0% opening means the closing of the valve. It has the same effect that there is no balancing tube. So it delivers step function input into step function output. We can confirm this phenomenon in the lowest waveform in Fig. 8 and first row of table 1. When we open the valve 30%, its corresponding time constant of high pass filter is 1.625, and the cut off frequency is about 0.1 Hz.

For the sinusoidal input as described in section 2.4, the amplitude of output signal for each frequency were plotted in Fig. 5(b) with '*'. As we see in the graph, the experimental results follow the properties of 1st order high pass filter.

Measurement Results from Human Body

Fig. 9 shows the measurement results from human body with the air mattress sensor system. The subject is 36 years old man with 67 Kg of body weight. The subject has no disease on sleep or cardiovascular system. Fig. 9(a) is the signal from the pressure sensor, which measures the pressure change in the air mattress system with a balancing tube. This figure shows clearly the respiration movement, which can be confirmed by comparing it with Fig. 9(b) that is the direct measurement of nasal air flow using a thermistor below the nose hole. Fig. 9(c) is the band pass filtered waveform in the range from 1 to 15 Hz. In order to extract the heart beat movement from the respiration movement, a digital filter (4th order Butterworth filter) was applied to the signal (a). The heart beat movement could be extracted from the air mattress system. The heart rate can be detected using the waveform of Fig. 9(d) which is the first derivative of Fig. 9(c). Fig. 9(e) is the conventional ECG recording. As a result, Fig. 9 shows that the suggested air mattress sensor system with a balancing tube can be used as one of the unconstrained monitoring of cardio-respiratory signals for sleep analysis.

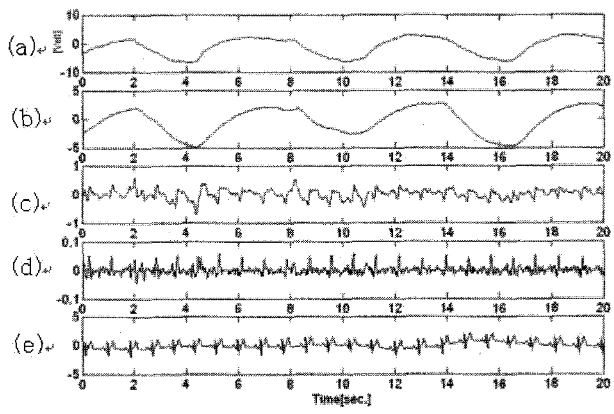


Fig. 9. The waveforms from air mattress and the conventional methods. All of the horizontal axes are the time in second, and all of the vertical axes are the amplified voltage outputs from each electrode and sensor. The meaning of each waveform is described in the text

Fig. 10 shows the respiration movement signal under various conditions. Fig. 10(a) is the signal when there was a small movement. From 120 to 140 second, we can see the drift of signal caused by the slight movement of subject. Through the effect of the balancing tube, the operating point of the sensor system returned to the zero position automatically. In Fig. 10(b), from 80 to 90 second, the subject changed his posture by 90° from the supine to lateral posture. The respiration cannot be measured during the movement. However, the signal also returned to a zero voltage automatically after the postural change had stabilized due to the effect of balancing tube. Fig. 10(c) is acquired while the subject stopped his breathing to simulate the apnea situation from 945 to 985 second. The heart beat information can be extracted from these waveforms whenever needed as shown in Fig. 9.

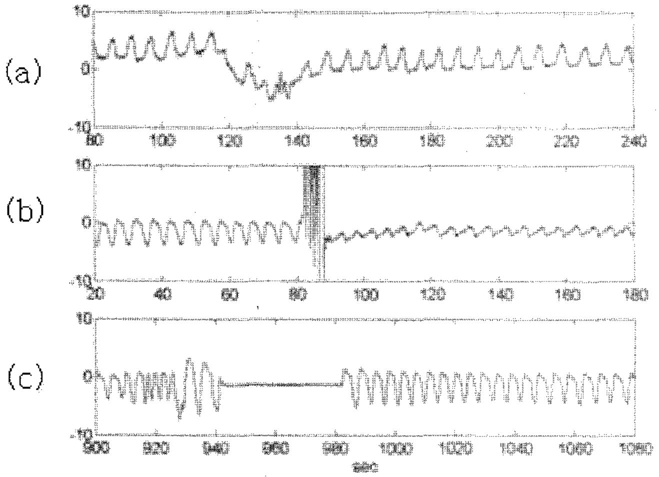


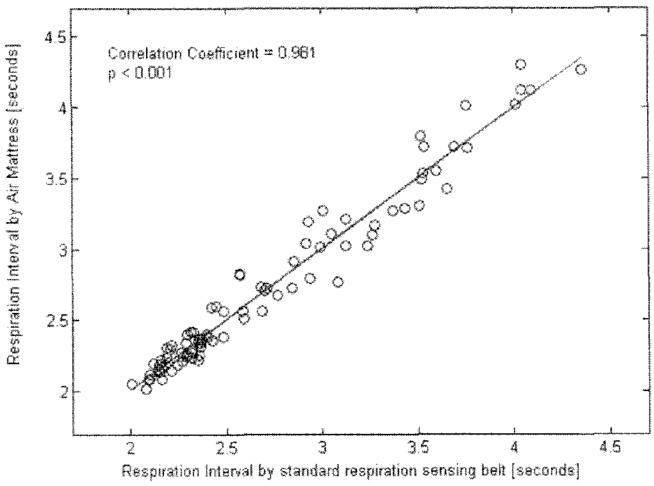
Fig. 10. Respiration signal under various conditions.

Accuracy and Error Analysis of the Measurement System

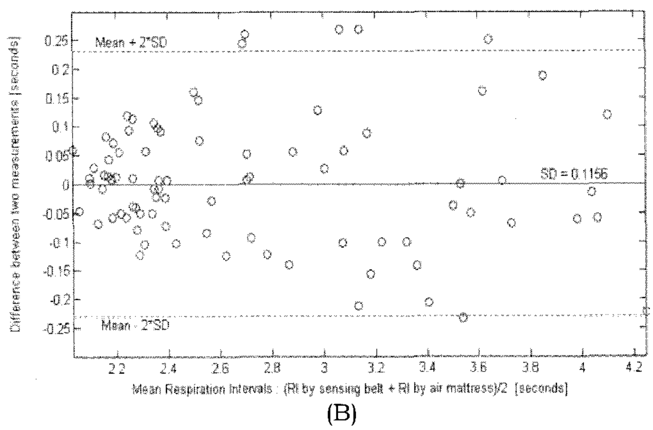
To analyze the measurement error between the standard method and air mattress method, the subject did dynamic exercise for 10 minutes and started the recording with air mattress for 5 minutes. While measuring with air mattress system, ECG and conventional respiration belt (plethysmography with chest belt) were measured. The R-R intervals from ECG data are compared with the heart beat interval from air mattress system by the method of Fig. 6. The Inter-Breathing Intervals (IBI) with conventional respiration sensor (Biopac Inc.) and air mattress were also measured and calculated to analyze the error of air mattress system.

Fig. 11 is the comparison data of the IBI that were calculated from standard respiration sensor and from air mattress. The respiration intervals from air mattress has high correlation (correlation coefficient = 0.981) with the signal from standard sensor. The Standard Deviation (SD) is 0.1156 second that is sufficiently small to monitor the breathing intervals. In Fig. 11(b), Bland Altman analysis was shown. The mean of the difference is lower than 0.005 second, the sampling interval.

Fig. 12 shows the comparison data of the heart beat intervals those were calculated from standard ECG and from air mattress. The heart beat intervals from air mattress and R-R intervals from ECG has also high correlation coefficient of 0.8516. The SD of the difference between two methods is 10 ms. The differences come from the methods to pick up the feature points in waveform to calculate rate. In Fig. 12, Bland Altman analysis was shown also.



(A)



(B)

Fig. 11. Comparison inter-breathing intervals from respiration effort sensor belt and air mattress
(a) Standard method (respiration belt) versus air mattress
(b) Bland Altman Analysis

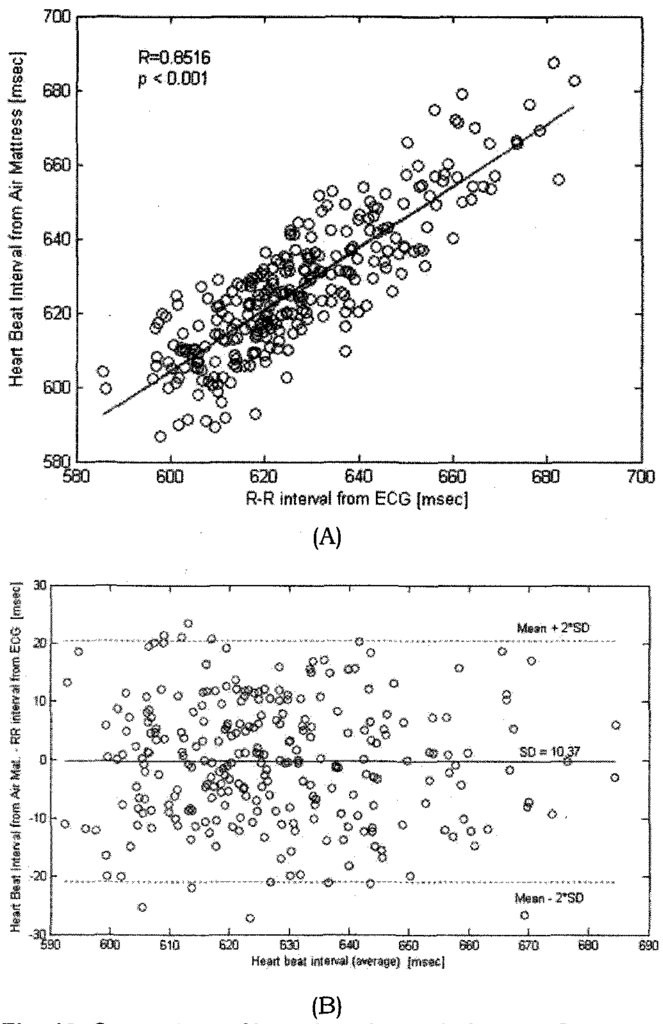


Fig. 12. Comparison of heart beat intervals from ECG and air mattress
(a) Standard method (ECG) versus air mattress
(b) Bland Altman Analysis

DISCUSSION

The modeling of the air mattress system with the balancing tube was done as in Eq. (3) and (4) as the role of band pass filter. Especially the balancing tube has the properties of high pass filter. Even though the band pass filter can be implemented with electrical components, there are few advantages of implementation it in pneumatic method. When we implement this filter with electrical components, because the dynamic range of pressures due to postural changes is so big compared with the pressure changes due to the respiration and heart beat movements, the measurement circuit should work in wide dynamic range with high sensitivity. Because we implemented the filter in pneumatic way in front of pressure sensor, the electric signal varies only the small range around zero voltage.

To find out the required cut off frequency of high pass filter in pneumatic way, i.e. time constant of 1st order RC high pass filter, the method using step function response was suggested and performed in section 2.3. For any subject from obese people to neonate, we can easily control the needed time constant by rotating the knob of valve in balancing tube. It changes the air resistance of balancing tube.

The waveforms have different shapes according to the contact locations of the subject with air cells because the respiration effort and movement is different in each part of human body. Even with these variances, calculating the respiration rate and heart beat rate was possible based on the robustness as a result of the differential measurement between the two cells.

For the unconstrained monitoring, the absence of respiration, respiration interval, heart beat interval, and level of activity has much information on his sleep. To see the accuracy of interval measurement with air mattress signals, the comparison with ECG and respiration belt is performed. The results shown in Fig. 12 for respiration interval shows enough accuracy (correlation coefficient is 0.981) for 5 minutes measurement. If the subject's movement in the similar frequency of respiration, it may makes error for measurement. For this kind of long term monitoring device for sleep analysis, the robustness is more important. The correlation between heart beat interval and RR interval from ECG is not so high enough to analysis of heart rate variability. (Correlation coefficient is 0.851, standard deviation is 10 ms.) This error comes mainly from localization of feature points and the limitation of sampling. With the current level of accuracy, we'd better to use the heart rate in time domain analysis.

CONCLUSION

The goal of this study is to develop the non-intrusive monitoring device for home healthcare. That was the basic reason to consider air mattress sensor because lying on the bed is inevitable position for daily sleep. With this measurement system, the physiological signal was detected without any electric contact with subject. A balancing tube was used to implement the robust sensing device against the postural changes during the measurement period. Using this balancing tube, the frequency of interest can be selected. The application for neonate and infant monitoring is one of the most promising areas for air mattress sensor system. Also to develop this system to effective monitoring device, intelligent and robust algorithm to calculate many parameters like time in bed, sleeping time, respiration interval during sleep and heart beat interval during sleep should be investigated.

REFERENCES

[1] W. Flenons, et al., *"Home Diagnosis of sleep Apnea: A Systemic Review of the Literature"*, CHEST, Vol.124, pp.1543-1579, 2003.

[2] D. White, et al., *"Assessment of Accuracy and Analysis Time of a Novel Device to Monitor Sleep and Breathing in the Home"*, Sleep, Vol. 18, No.2, pp. 115-126, 1995.

[3] S. Tanaka., Y. Matsumoto, and K. Wakimoto, *"Unconstrained and non-invasive measurement of heart-beat and respiration periods using a phonocardiographic sensor"*, Med. Biol. Eng. Comput., Vol. 40, pp.246-252, 2002.

[4] J.L. Jacobs, P. Embree, M. Glei, S. Christensen, and P.K. Sullivan, *"Characterization of a novel heart and respiratory rate sensor"*, Proc. Of 26th Ann. International Conference of IEEE EMBS, pp.2223-2226, 2004.

[5] W.B. Spillman Jr., M. Mayer, J. Bennett, J. Gong, K.E. Meissner, B. Davis, R.O. Clauss, A.A. Muelenaer Jr., and X. Xu, *"A smart bed for non-intrusive monitoring of patient physiological factors"*, Meas. Sci. Technol., Vol. 15, pp.1614-1620, 2004.

[6] J. Alihanka, A. Vaahtoranta, and K. Saarikivi, *"A new method for long term monitoring of the ballistocardiogram, heart rate, and respiration"*, Am. J. Physiol., Vol. 240, pp.384-392, 1981.

[7] L. Hernandez, B. Waag, H. Hsiao, and V. Neelon, *"A new non-invasive approach for monitoring respiratory movement of sleeping subjects"*, Physiol. Meas., Vol. 16, pp.161-167, 1995.

[8] P. Chow, G. Nagendra, and J. Abischeganaden, Y.T. Wang, *"Respiratory monitoring using an air-mattress system"*, Physiol. Meas., Vol. 21, pp.345-354, 2000.

[9] T. Watanabe, and K. Watanabe, *"Noncontact Method for Sleep Stage Estimation"*, IEEE Trans. On Biomedical Engineering, Vol. 51, No.10, 1735-1748, 2004.

[10] Y.J. Chee, J.M. Han, J.W. Youn, and K.S. Park, *"Air mattress sensor system with balancing tube for unconstrained measurement of respiration and heart beat movements"*, Physiol. Meas., Vol. 26 pp. 413-422, 2005.

[11] M. Ogata, System Dynamics, New Jersey, Prentice-Hall, pp.258-264, 1978.

[12] W. Tompkins, Biomedical Digital Signal Processing, New Jersey, Prentice-Hall, pp.236-254, 1993.