

# Development of a Condition Monitoring System Using an Air-pack Type Pressure Sensor for Bedridden Patients in a Supine Position

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**Abstract**—This paper proposes an online biological signal measurement system with an air-pack-type pressure sensor (APS) to monitor the condition of supine patients from pulse activity. The system enables estimation of three physiological signals corresponding to pulse pressure waves, pulse beats and respiratory waves from measurements taken and filtered using the APS, which has mechanical characteristics similar to the impedance characteristics of human muscle and can also be fitted with a mattress pad to prevent decubitus. In this study, the physiological significance of signals obtained through the APS was determined by comparing them with signals measured using commercial medical devices. The results showed that the correlations of pulse pressure waves, pulse beats and power spectral densities for respiratory waves between signals measured using the APS and those measured using commercial sensors were 0.84, 0.84 and 0.98, respectively. It was therefore concluded that the proposed system is suitable for outputting pulse pressure waves, pulse beats and respiratory waves.

**Keywords** - Air-pack sensor, Biological signal, Noninvasive monitoring, The cervical spinal cord injured

## I. INTRODUCTION

Patients involved in traffic accidents and those with general paralysis from injuries such as cervical spinal cord damage are often prescribed complete bed rest in Japan [1]. In particular, spinal cord injuries are accompanied by the potential risk of

complications such as bradycardia, sleep apnea syndrome, arteriosclerosis and decubitus. Many patients with such injuries receive treatment at home, and those suffering from related quadriplegia need life maintenance and round-the-clock monitoring from a care worker, a healthcare professional or a family member. However, as it is difficult for such observers to constantly monitor the patient's condition, a system that can perform this task in place of an observer is required.

Such a monitoring system needs to fulfill the following requirements: (1) it must detect urgent dangers such as arrhythmia and the cessation of breathing and alert a healthcare professional; (2) it must detect chronic dangers to the circulatory system such as abnormal blood pressure; (3) it must have prophylactic functions to prevent the death of quadriplegic patients; (4) it must be able to monitor the patient's biological signals in a nonbinding manner over a long period; and (5) it must be easy to use. In previous studies, condition monitoring systems that measured and evaluated the biological signals of patients in a nonbinding manner were developed using a poly-vinylidene fluoride sensor [2], a camera system [3] and a microwave radar [4]. A system that can monitor arrhythmia, sleeping times and sleep apnea syndrome as well as measuring and evaluating biological signals was also developed using measurement sensors installed under a mat [5]. However, although these systems can

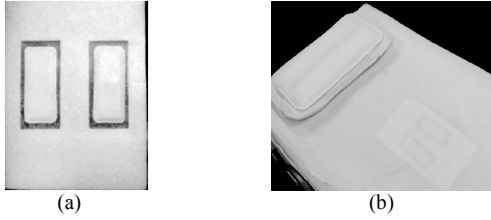


Fig. 1 Experimental equipment for measuring biological signals: (a) air-pack sensor, (b) mattress

detect urgent dangers such as abnormal cardiac rhythm and sleep apnea, they cannot measure pulse pressure waves. Accordingly, they are unable to detect chronic dangers to the circulatory system, and consequently cannot be used for prophylaxis.

This paper proposes a monitoring system that can be used to examine variations in vascular mechanical characteristics using an air-pack sensor that constantly measures pulse pressure waves of patients with cervical spinal cord injuries over a long period. The system can be used to measure pulse pressure waves, pulse beats, and respiratory waves. It can also be used as a mattress to reduce the constriction of blood flow in order to prevent decubitus.

## II. A VITAL-SIGN MONITORING SYSTEM WITH AN AIR-PACK SENSOR

### A. Air-pack sensor

Figure 1 (a) shows the air-pack-type pulse pressure sensor [7] used in the proposed system. The sensor was originally developed for a car seat, and was modified for this system. It consists of a pair of air bags made of urethane elastomer sheets, a capacitor-type microphone sensor and a 3D polyester net to act as a spring and damper. The 3D net flexes when pressure is applied to a small area of it. In contrast, when pressure is applied to a wider area, a large reactive force from the net acts on the contact surface, meaning that the 3D net reflects the characteristics of pressure dispersion from the human body. The 3D net facilitates the transmission of pressure fluctuations from the human body [7]. Air-pack sensor elasticity depends on the air compressibility and restoring force of the 3D net. The spring constant of the air space in the air pack  $K_a$  is expressed as follows [8]:

$$K_a = \gamma P_0 \frac{l_p a}{h_0}, \quad (1)$$

where  $\gamma$  is the specific heat ratio of air,  $P_0$  is atmospheric pressure,  $a$  is the width of the air pack,  $l_p$  is its length, and  $h_0$  is its initial thickness. Then, a combined spring constant  $K$  between that of the air pack  $K_a$  and that of the 3D net  $K_c$  is expressed as

$$K = K_a + K_c. \quad (2)$$

However, the air current in the air pack is seen as an incompressible viscous flow. The thickness of the air pack  $h$ , its vibration velocity  $\dot{h}$ , and its vibration acceleration  $\ddot{h}$  are expressed as follows:

$$h = \hat{h} - A \sin \omega t, \quad (3)$$

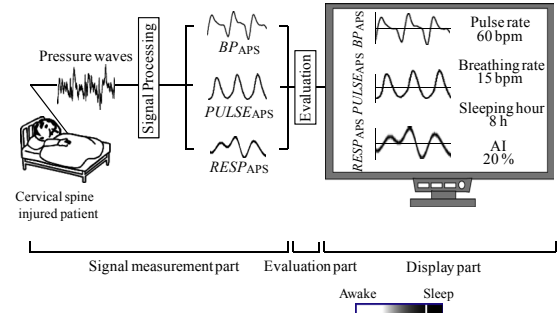


Fig. 2 Overview of proposed system

$$\dot{h} = -A\omega \cos \omega t, \quad (4)$$

$$\ddot{h} = A\omega^2 \sin \omega t, \quad (5)$$

where  $\omega$  is the vibration frequency of the air pack  $\hat{h}$  is the equilibrium position, and  $A$  is the amplitude. The air pack's internal pressure  $P$  is expressed as follows:

$$\frac{P}{\rho} = -\frac{l_p^2}{2} \frac{A\omega^2 \sin \omega t}{\hat{h}} + \frac{\beta l_p^2 A\omega \cos \omega t}{2\hat{h}^3}, \quad (6)$$

where  $\rho$  is the density and  $\beta$  is a constant. Then, following Equation (3), the transfer function of pressure variation  $G(s)$  from an air pack displacement to a capacitor-type microphone sensor is expressed as follows [8]:

$$G(s) = -\frac{\rho l_p^2 \frac{1}{h_0} s^2 + \rho \beta l_p^2 \frac{1}{h_0^3} s + \frac{2K}{al_p}}{m_c s^2 + k_c}. \quad (7)$$

Output waves measured using the air-pack sensor are then filtered using a capacitor-type microphone sensor, which can be installed in various places such as inside a car seat and enables simple measurement of biological signals [7].

Figure 1 (b) shows a mattress consisting of 3D nets that can be used to prevent decubitus [9]. The mattress also has mechanical characteristics similar to the impedance characteristics of human muscle. Thus, its shape readily transforms in the vicinity of bone projections when a patient lies on it; the compression of peripheral circulation is mitigated, and blood flow constriction is reduced.

### B. Health monitoring system

Figure 2 shows the proposed health monitoring system, which consists of three parts. (1) The measurement part is used to measure pressure waves via the air-pack sensor; waves corresponding to pulse pressure waves, pulse beats and breathing waveforms are extracted from the filtered output pressure waves of the sensor. (2) The evaluation part is used to calculate indices pertaining to the condition of the circulatory system using the above-mentioned filtered waves. (3) The display part is used to show the calculated indices.

In the measurement part, the pressure waves extracted from the subject's dorsal region are filtered using three fourth-order infinite impulse response (IIR) low-pass and high-pass filters. The first set of filters includes a 0.3-Hz IIR high-pass filter and a 10-Hz low-pass filter; the second set includes a 0.5-Hz IIR high-pass filter and a 2.0-Hz low-pass filter; and the third set includes a 0.1-Hz IIR high-pass filter and a 0.5-Hz low-pass filter. Signals corresponding to pulse pressure waves ( $BP_{APS}$ )

are obtained using the first set of filters, those corresponding to pulse beats ( $PULSE_{APS}$ ) obtained using the second set, and those corresponding to breathing waveforms ( $RESP_{APS}$ ) obtained using the third set, respectively, when the patient is sleeping so that the influences of body motion and changes in autonomic nervous activity are negligible.

In the evaluation part,  $PULSE_{int_{APS}}$  is defined as the time interval of  $PULSE_{APS}$  between the maximum value of the current wave and the maximum value of the next cycle. The heart rate  $HR_{APS}$  is expressed as

$$HR_{APS} = \frac{1}{PULSE_{int_{APS}}} \times 60. \quad (8)$$

$RESP_{int_{APS}}$  is defined as the time interval of  $RESP_{APS}$  between the maximum value of the current wave and the maximum value of the following cycle. The breathing rate  $BR_{APS}$  is expressed as

$$BR_{APS} = \frac{1}{RESP_{int_{APS}}} \times 60. \quad (9)$$

The time-series waveforms and the results of the raw data for  $BP_{APS}$ ,  $PULSE_{APS}$  and  $RESP_{APS}$  are shown in the display part. The mean values of  $HR_{APS}$  and  $BR_{APS}$  over 15 seconds are displayed.

### III. EXPERIMENT

To verify the physiological significance of signals measured using the air-pack sensor in the proposed system, signals obtained in this way and others measured using a commercial physiological monitor were compared in an experiment involving five healthy young male subjects (mean age  $\pm$  S.D.:  $23.8 \pm 0.5$  years), who were asked to assume a supine position in a resting state on a mattress containing an air-pack sensor.  $BP_{APS}$ ,  $PULSE_{APS}$  and  $RESP_{APS}$  (as described in Section II) were measured for 3 minutes using the sensor. Simultaneously, carotid artery blood pressure ( $BP_c$ ) and radial artery blood pressure ( $BP_r$ ) were measured from the right common carotid artery using a pressure pulse wave transducer (TK-701T, Nihon Kohden) and the physiological monitor (BP-608E, Omron Colin) used for the comparison of  $BP_{APS}$ .

Electrocardiogram (ECG) signals were also measured using the same physiological monitor to enable the comparison of  $PULSE_{APS}$  values, and respiratory waves (RESP) were measured using a multi-channel telemeter system (WEB-7000, Nihon Kohden) to enable comparison of  $RESP_{APS}$  values. Each signal was measured at 1,000 Hz using an analog-digital card (CSI-3601169, Interface), and the values were entered into a PC. The measured  $BP_{APS}$ ,  $BP_c$  and  $BP_r$  values for each beat were isolated using  $PULSE_{int_{APS}}$ . The direct-current component of each signal was then removed using the low-pass and high-pass filters discussed in Section II, and the coefficient of correlation between  $BP_{APS}$  and  $BP_r$  and that between  $BP_{APS}$  and  $BP_c$  were calculated. The  $PULSE_{int_{APS}}$  value discussed in Section II and the RR interval of the measured subject's ECG were extracted, and the coefficient of correlation between the two was calculated. The fast Fourier transforms of the measured  $RESP_{APS}$  and  $RESP$  values were performed, then the coefficient of correlation between the calculated power spectrum densities of  $RESP_{APS}$  and  $RESP$  was ascertained.

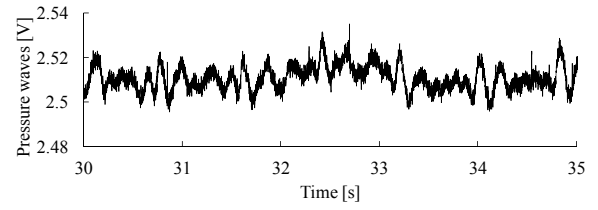


Fig. 3 Measured pressure waves (Subject A)

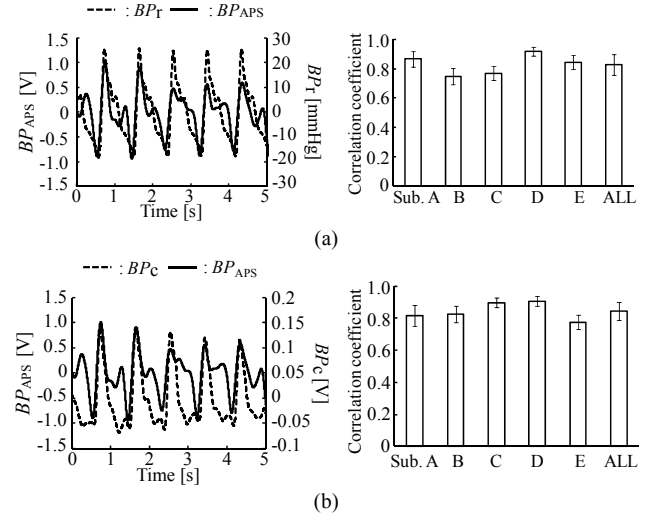


Fig. 4 Comparison among  $BP_r$ ,  $BP_c$ , and  $BP_{APS}$ : (a) correlation coefficients in one-beat signals between  $BP_{APS}$  and  $BP_r$ , (b) correlation coefficients in one-beat signals between  $BP_{APS}$  and  $BP_c$ .

Next, to verify the proposed system's ability to monitor a bedridden patient, patient's biological signals were measured while a patient was sleeping. One 57-year-old patient who had suffered an injury at the fifth cervical vertebra C5 required total assistance for activities of daily living (ADL) due to quadriplegia and had arteriosclerosis was asked to maintain a supine position in a resting state on a mattress containing an air-pack sensor, and  $BP_{APS}$ ,  $PULSE_{APS}$  and  $RESP_{APS}$  were measured during sleep over a period of one week. Each signal was measured at 1,000 Hz using an analog-digital card, and the values were entered into a PC. In this experiment, the period of sleep was defined as the time between the patient pressing a PC key before falling asleep and pressing it again after waking up. Informed consent was obtained from all study subjects before the experiments were performed based on the Declaration of Helsinki.

### IV. RESULTS AND DISCUSSION

Figure 3 shows pressure waves measured at rest using the proposed system. The waves were non-periodic and included electrical noise, and were used to calculate the values of  $BP_{APS}$ ,  $PULSE_{APS}$  and  $RESP_{APS}$ .

Figure 4 shows the  $BP_{APS}$ ,  $BP_r$  and  $BP_c$  waveforms measured at rest with the direct-current component removed. The mean values of the coefficients of correlation between  $BP_{APS}$  and  $BP_r$  and between  $BP_{APS}$  and  $BP_c$  were calculated for all subjects as  $0.83 \pm 0.07$  ( $p < 0.01$ ) and  $0.84 \pm 0.06$  ( $p < 0.01$ ), respectively, indicating a high correlation.

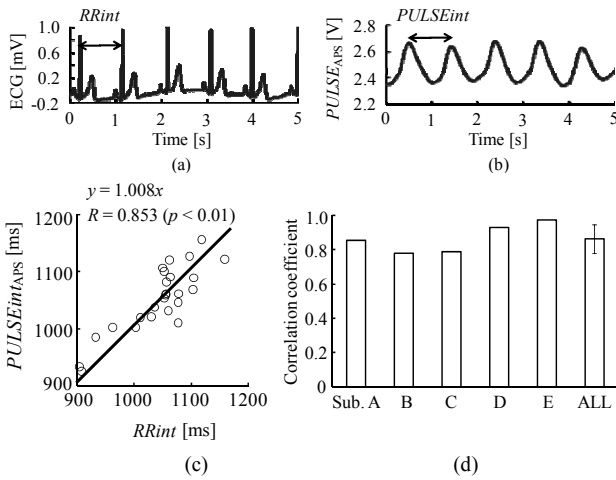


Fig. 5 Comparison between  $PULSE_{APS}$  interval and  $RR_{int}$  measured from ECG signals: (a) example of measured ECG signals (Subject A); (b) example of measured  $PULSE_{APS}$  values (Subject A); (c) point diagram showing  $PULSE_{APS}$  interval vs.  $RR_{int}$  (Subject A); (d) all subjects

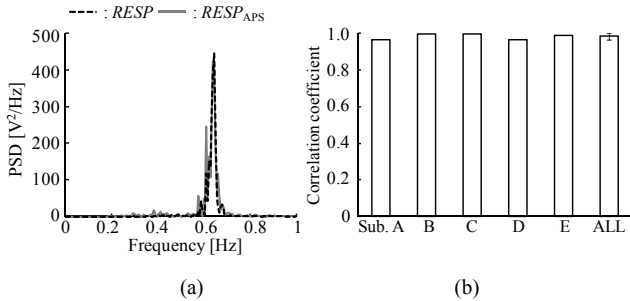


Fig. 6 Comparison of frequency components between  $RESP_{APS}$  and  $RESP$ : (a) subject D, (b) all subjects

Figure 5 shows the calculated coefficients of correlation between  $PULSE_{int_{APS}}$  and  $RR_{int}$  for all subjects, and indicates a mean value of  $0.84 \pm 0.09$  ( $p < 0.01$ ). Here, it is understood that  $HR_{APS}$  is almost equal to the heart rate measured using a commercially available device because the inclination ratio of the single regression line is almost 1. In addition, the average percentage error between  $PULSE_{int_{APS}}$  and  $RR_{int}$  for all subjects was  $2.90 \pm 1.20$ .

Figure 6 shows the PSD values found using FFT for  $RESP_{APS}$  and  $RESP$  and the mean value of the coefficients of correlation between these PSD values for all subjects. The PSD shows peaks near the same bandwidth, and the mean coefficient of correlation was  $0.98 \pm 0.01$  ( $p < 0.01$ ). It was confirmed that each signal obtained from the proposed system exhibited good measurement accuracy.

Figure 7 (a) shows the  $BP_{APS}$ ,  $PULSE_{APS}$  and  $RESP_{APS}$  waveforms of the patient measured using the proposed system 30 minutes after the patient fell asleep. Each signal shows rhythmicity in a manner similar to that of signals measured from the healthy subjects.

## V. CONCLUSION

This study investigated a condition monitoring system for

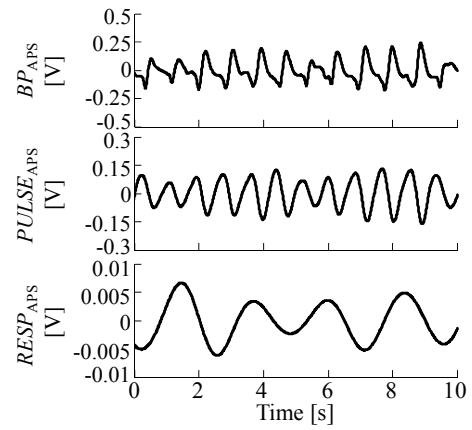


Fig. 7 Experimental results for patient with a cervical spinal cord injury during sleep: (a) measured waveform of each signal

cervical cord injury patients that includes an air-pack sensor to measure pulse pressure waves. A system capable of extracting biological signals from such patients was constructed, and the signals of five healthy subjects and a patient with cervical cord injury were measured at rest. It was confirmed that each signal was accurately obtained using the proposed system. The results further indicated that the system could be used to detect arteriosclerosis at an early stage. In the future, we intend to improve the proposed system to enable judgment of sleep conditions using the measured output signals of the proposed sensor and to allow simultaneous monitoring for sleep apnea and abnormal cardiac rhythm.

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