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Fundamental study of measurement of cardiorespiratory signals in a sitting position using piezoelectric sensors

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Abstract— **In this paper, we propose a measurement system to extract the heartbeat and respiration signals from piezoelectric sensors located in a chair seat. Eight healthy male subjects (22–26 years old) sat on a chair with piezoelectric sensors. The experiment consisted of five acquisitions (300 s recording with 30 s resting). In the 1st acquisition, the subject was instructed to maintain a relax state. During the next three acquisitions, the subject maintained respiration rates of 3, 4, and 5 s. The order was randomly given for each subject. Finally, the subject was instructed to stay in a resting state for 300 s. Band-pass filters were used to separate the heartbeat and respiration signals from the output signal of the piezoelectric sensors (heartbeat: 0.7–7 Hz, respiration: 0.1–0.4 Hz). Then, the standard heartbeat interval was calculated using the autocorrelation function to generate a template. Heartbeat and respiration signals were extracted using a two-dimensional cross-correlation pattern matching method. For validation, the error ratio between the piezoelectric sensor data and the reference data was determined. The error ratios of the heartbeat interval and respiration interval were 2.89±4.41% and 5.57±4.42%, respectively. Therefore, it is suggested that heartbeat and respiration signals extracted from piezoelectric sensors in a sitting position can be used as an alternative method for extracting biological signals.**

I. INTRODUCTION

In recent years, the aging population has grown around the world. In addition, increasing public health problems such as heart disease have raised concerns over health risks. Biological information must be measured to monitor health on a daily basis. Health care monitoring installed at home is one solution for collecting such biological information.

Conventional devices such as scales, thermometers, and sphygmomanometers are now widely used in homes to determine basic information about our physical conditions. The advantages of such devices are their low initial/running cost, ease of use, and compact size/low weight. To obtain the detailed information on patients' physical conditions, it is necessary to visit a clinic or hospital. However, most of us usually only go to the hospital or clinic when we are ill or injured, and not for health maintenance. In addition, the medical equipment in clinics and hospitals requires high initial/running cost, specialized training to operate, or a special environment for placement. Such difficulties obstruct home healthcare monitoring.

It is well known that the heart rate and respiration rate contain much information for understanding our physical condition. Notably, it is also known that heart-rate variability (HRV) represents the activity of the autonomic nervous system. Hence, it can be used to recognize not only heart failure but also diseases that affect the autonomic nervous system. Consequently, it is quite effective for measuring cardiorespiratory information at home.

Recently, various methods to record cardiorespiratory signals instead of conventional medical equipment have been developed. Gramse et al.[1] developed pajamas with embedded electrocardiograph (ECG) electrodes and strain gauges (abdominal respiration sensors). Meanwhile, piezoelectric sensors[2][3][4], phonocardiographs[5], and bioradar[6] have been employed to detect the vibrations or sounds generated by the cardiorespiratory organs.

However, most reports have focused mainly on monitoring during sleep. That is, the examinations have been performed in the supine position. Although, it may be necessary to measure cardiorespiratory signals when subjects are awake, few reports have investigated the properties of HRV indices with subjects in a sitting position. Therefore, in this study, we proposed a method that measures cardiorespiratory signals with subjects in a sitting position using piezoelectric sensors and that extracts each component from the recorded signals. We then confirmed the validity of the proposed method and the differences in HRV properties from ECG or abdominal respiration signals.

II. METHODS

A. Subjects

Eight healthy males, 22–26 years old, participated in the following experiment. Before the experiment, informed consent was obtained from each subject. Subjects were instructed to sit on a chair and maintain a relaxed state. Then, 300 s worth of cardiorespiratory signals were acquired five times at 30 s intervals. From the 2nd to the 4th acquisition, the subject was directed to breathe with at 3, 4, or 5 s intervals (randomly selected), whereas, for the 1st and the 5th acquisitions, the subject breathed at his own pace.

B. Recordings

Two piezoelectric sensors (DT4-028K/L, Tokyo Sensor Co., Ltd., Japan) were placed on the seat of the chair (Fig. 1). The location of each sensor was adjusted to correspond to a middle point of the short side of the sensor and the ischial bone of the subject. At the same time that the left/right piezoelectric signals were measured, ECGs from standard limb lead II and abdominal respiration were measured as reference signals (Fig. 2). All of the signals were filtered

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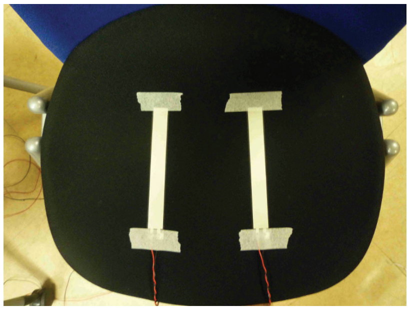

Fig. 1. Piezoelectric sensors placed on the seat of the chair.

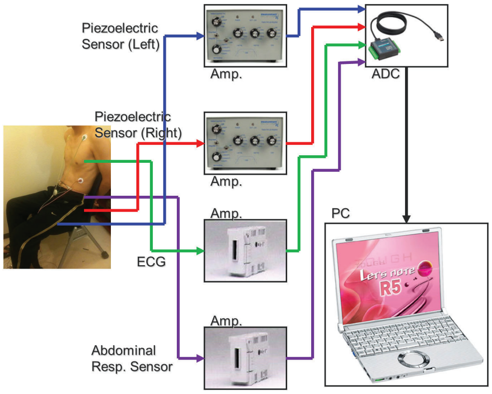

Fig. 2. Experimental environment.

(0.1–10 Hz band-pass filter (BPF) for piezoelectric signals, 40 Hz low-pass filter (LPF) for ECG, 30 Hz LPF for abdominal respiration) and amplified, and then stored in a PC with a 1 kHz sampling frequency.

C. Analyses

1) Cardiac component extraction: As a cardiac component, the heartbeat interval was extracted from the left/right piezoelectric signals using the following procedure:

1) Considering that the heart rate of a healthy subject in the resting state ranges from 50 to 100 bpm, we applied a 0.7 Hz high-pass filter (HPF) to the recorded piezoelectric signals as a primary filter.
2) We performed half-wave rectification to the primary-filtered signals and to those with polarity-inversion.
3) We applied a 0.7–7 Hz BPF filter to the four half-wave rectified signals as a secondary filter.
4) From the beginning 5 s of each secondary-filtered signal, we calculate the standard heartbeat interval using an autocorrelation function.
5) We divided the initial 5 s of each secondary-filtered signal of every standard heartbeat interval and then generated a template by averaging the divided signals.
6) We choose two secondary-filtered signals out of four and executed template matching using a two-dimensional cross-correlation function.
7) We searched every local maximum of the correlation coefficient that exceeds the preset threshold level defined for "heartbeat." If there were two or more candidates, we selected the candidate that is closest to the standard heartbeat interval by computing the time difference from the previous heartbeat.
8) We obtained a time series of the heartbeat interval.

2) Respiratory component extraction: As a respiratory component, the breath interval was extracted from the left/right piezoelectric signals by the following procedure:

1) Considering that the respiratory rate of the healthy subject in the resting state reanges from 12 to 20 bpm, we applied a 0.4 Hz LPF to the recorded piezoelectric signals as a primary filter.
2) We perform half-wave rectification to the primary-filtered signals and to those with polarity inversion.
3) We applied a 0.1–0.4 Hz BPF to the four half-wave rectified signals as a secondary filter.
4) From the beginning 30 s of each secondary-filtered signal, we calculated the standard breath interval using an autocorrelation function.
5) We divided the initial 30 s of each secondary-filtered signal of every standard breath interval and then generated a template by averaging the divided signals.
6) We chose two secondary-filtered signals out of four and executed template matching using a two-dimensional cross-correlation function.
7) We searched every local maximum of the correlation coefficient that exceeded the preset threshold level defined as "breath." When two or more candidates were present, we selected the one that was closest to the standard breath interval by computing the time difference from the previous breath.
8) We obtained a time series of the breath interval.

3) Validation: To validate the proposed method, the time series of the heartbeat/breath interval, which was obtained from the two piezoelectric sensors, was compared to that obtained from ECG/abdominal respiration. For the time series of the heartbeat interval, HRV properties were also examined. The investigated indices were

- Time domain analysis
  - standard deviation of the beat-to-beat/R-to-R interval (SDNN)
  - coefficient of variance of the beat-to-beat/R-to-R interval (CVNN)
  - rootmean-square successive difference (RMSSD)
- Frequency domain analysis
  - low-frequency (0.04–0.15 Hz) power (LF)
  - high-frequency (0.15–0.4 Hz) power (HF)
  - ratio of LF to HF (LF/HF)

For all comparisons, the error ratio was used for the validation.

III. RESULTS

Fig. 3 shows the representative example of the recorded signals from the two piezoelectric sensors (subject #1, controlled respiration every 5 s, beginning at 30 s). Two periodical features were observed: in one, spike-like components were repeated approximately every 0.7 s; in the other,

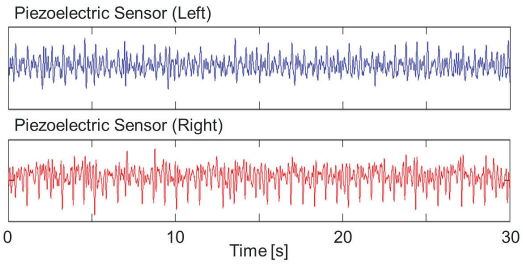

Fig. 3. Representative example of recorded signals from two piezoelectric sensors (subject #1, controlled respiration every 5 s, beginning at 30 s).

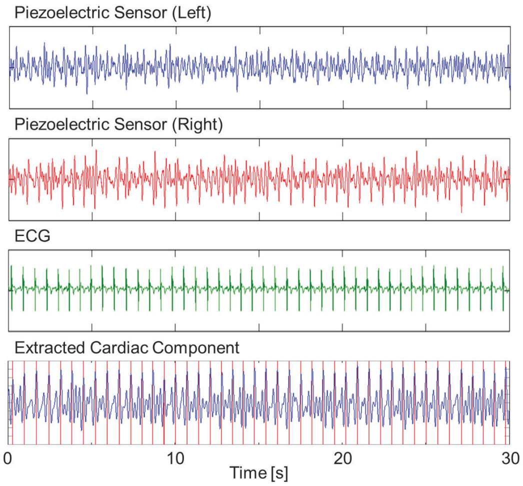

Fig. 4. Representative example of primary-filtered (0.7 Hz HPF) signals of Fig. 3 (1st and 2nd columns), simultaneously recorded ECG (3rd column), and extracted cardiac component (curves in the 4th column) with the corresponding occurrence of R waves in the ECG (lines in 4th column).

baseline drift-like components were repeated approximately every 5 s.

A. *Cardiac component*

The top three columns of Fig. 4 show the cardiac component of Fig. 3 (primary-filtered by a 0.7 Hz HPF) and its reference signal (ECG). We found that the spike-like waveforms in the primary-filtered signal corresponded to R waves in the ECG. The bottom column of Fig. 4 shows the two-dimensional cross-correlation, which was obtained from the secondary-filtered signals, and the corresponding occurrence of R waves in the ECG. We found that each local maximum of the correlation coefficient agreed with each occurrence of an R wave.

TABLE I lists the error ratio of the time difference between the heartbeat interval from piezoelectric sensors and the R-R interval from the ECG. The minimum and maximum error ratios for the overall data were 0.21% and 22.61%, respectively (mean ± SD $= 2.89 \pm 4.41\%$).

Fig. 5 shows the power spectrum of the heartbeat interval/R-R interval of subject #1 with controlled respiration every 3 (top), 4 (middle), and 5 s (bottom). A similar power spectrum was found in the other measurement. In particular, the same significant peak in the HF band was observed in both measurements.

TABLE I

ERROR RATIO OF TIME DIFFERENCE BETWEEN HEARTBEAT INTERVAL FROM PIEZOELECTRIC SENSORS AND R-R INTERVAL FROM ECGS FOR EACH SUBJECT AND ACQUISITION [%]

Subject	Acquisition				
	1st	2nd	3rd	4th	5th
#1	1.55	1.53	0.31	1.43	0.29
#2	0.91	1.74	0.26	0.26	1.66
#3	0.89	0.97	0.51	1.42	4.81
#4	3.45	0.21	0.72	0.31	1.70
#5	0.51	22.61	5.48	0.85	9.91
#6	6.98	1.44	1.96	15.90	2.81
#7	1.25	1.50	1.47	3.54	0.56
#8	1.21	0.85	6.42	4.47	1.08

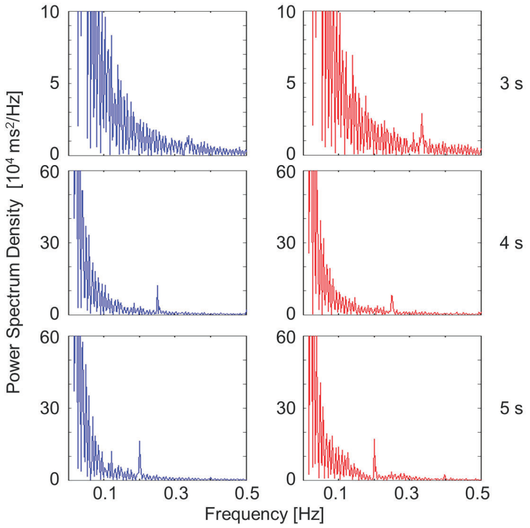

Fig. 5. Representative example of power spectrum of heartbeat interval (right) and that of R-R interval (left) of subject #1 with controlled respiration every 3 (top), 4 (middle), and 5 s (bottom).

TABLE II lists the error ratios of the value difference of the HRV parameters. The grand mean error ratios were 4.23±3.31% (SDNN), 4.24±3.33% (CVNN), 14.20±11.75% (RMSSD), 5.99±5.75% (LF), 3.07±2.75% (HF), and 3.84±5.00% (LF/HF).

B. *Respiratory component*

The top three columns of Fig. 6 show the respiratory component of Fig. 3 (primary-filtered by a 0.4 LPF) and its reference signal (abdominal respiration). We found that similar waveforms were appeared approximately every 5 s in each signal. The bottom column of Fig. 6 shows the two-dimensional cross-correlation, which was obtained from the secondary-filtered signals, and the corresponding peaks in the reference signal. We found that each local maximum of the correlation coefficient agreed with each peak in the reference signal.

TABLE III lists the error ratios of the time difference between the breath interval from piezoelectric sensors and that from the abdominal respiration sensor. The minimum and maximum error ratios for overall data were 0.72% and 16.74%, respectively (mean ± SD $= 5.57 \pm 4.42\%$).

TABLE II

MEAN ERROR RATIO OF VALUE DIFFERENCE OF HRV PARAMETERS BETWEEN HEARTBEAT INTERVAL FROM PIEZOELECTRIC SENSORS AND R-R INTERVAL FROM ECG DATA FOR EACH SUBJECT [%]

Subject	HRV parameter					
	SDNN	CVNN	RMSSD	LF	HF	LF/HF
#1	5.76	5.80	18.84	8.85	2.60	6.32
#2	3.03	3.04	11.56	3.90	3.27	1.60
#3	1.90	1.89	13.56	8.85	3.37	4.24
#4	2.89	2.92	9.61	1.63	1.12	1.47
#5	3.14	3.41	4.97	3.53	3.63	1.47
#6	4.48	4.42	20.06	10.76	4.57	6.52
#7	5.85	5.83	22.06	9.51	2.43	6.92
#8	5.79	5.85	13.94	4.96	3.58	2.18

TABLE III

ERROR RATIO OF TIME DIFFERENCE BETWEEN BREATH INTERVAL FROM PIEZOELECTRIC SENSORS AND THAT FROM ABDOMINAL RESPIRATION SENSOR OF EACH SUBJECT AND ACQUISITION [%]

Subject	Acquisition				
	1st	2nd	3rd	4th	5th
#1	5.00	6.48	2.64	8.03	4.82
#2	1.46	1.13	3.44	7.17	6.74
#3	12.49	2.07	1.55	2.88	15.53
#4	15.87	2.58	5.34	5.96	0.72
#5	6.06	2.03	2.32	4.56	10.72
#6	3.83	6.20	1.50	1.37	2.28
#7	13.01	7.23	2.16	4.34	7.80
#8	16.74	5.21	9.01	2.21	2.26

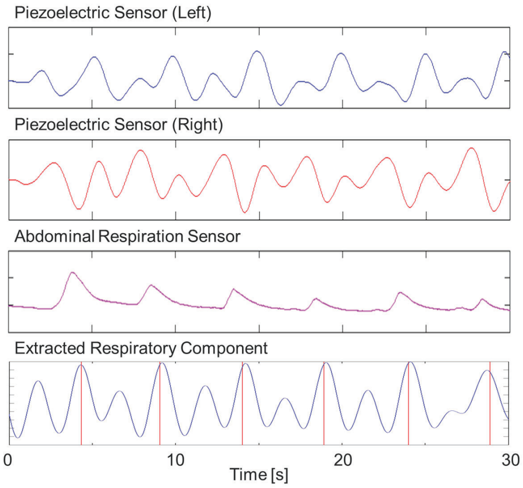

Fig. 6. Representative example of primary-filtered (0.4 Hz LPF) signals of Fig. 3 (1st and 2nd columns), simultaneously recorded abdominal respiration (3rd column), and extracted respiratory component with the corresponding peaks of the abdominal respiration signal.

IV. DISCUSSION

In this study, we employed piezoelectric sensors for the measurement of cardiorespiratory signals in a sitting position without direct skin contact of the electrodes or sensors. To extract cardiac/respiratory components from piezoelectric signals, we proposed the procedure of template matching using two-dimensional cross-correlation. For validation, we compared the time difference of the heartbeat/breath intervals obtained from piezoelectric sensors and reference sensors.

For the cardiac component, we obtained an overall error ratio of 2.89±4.41% in the comparison of the heartbeat intervals acquired from piezoelectric sensors and ECG. Tanaka *et al.*[5] reported that they measured heartbeat signals for 60 s in the supine position using a phonocardiograph and obtained an error ratio within 1.15%. Our result was less accurate than theirs; however, artifacts, which might be elicited by body movement may have distorted the accuracy, especially for subject #5 (2nd and 5th acquisitions) and subject #6 (1st and 4th acquisitions). If these four data items are ignored, the error ratio is reduced to 1.68±1.54%. Differences in the recording/analysis duration or in the position during recording may also have affected our results.

For the HRV properties, we obtained less than 7% of the error ratio for SD, CV, HF, and LF/HF indices. Because more than 20% or 10% of the error ratio represents the RMSSD or LF index, respectively, this error ratio may not have sufficient precision. However, in the case of Fig. 5, the significant peak frequency in the HF band [0.200 Hz (top), 0.251 Hz (middle), or 0.337 Hz (bottom)] acquired from piezoelectric sensors agreed with that obtained from ECG data. It also corresponded to the mean respiration frequency [0.197 Hz (top), 0.247 Hz (middle), or 0.330 Hz (bottom)].

For the respiratory component, the error ratio was 5.57±4.42% when comparing the breath intervals obtained from piezoelectric sensors and the abdominal respiration sensor. Again, Tanaka *et al.*[5] reported that they measured respiration signals for 60 s in the supine position using a phonocardiograph and obtained an error ratio within 5.20%. Our result was nearly as accurate as theirs, althogh more than 10% of the error ratio occurred in the 1st and the 5th acquisitions. Thus, variation of the template component, which was derived from the irregularity of the respiratory period, might worsen the accuracy in uncontrolled respiration. However, the respiratory rate during acquisition from piezoelectric sensors perfectly matched that of the abdominal respiration sensor.

These results suggest that heartbeat and respiration signals extracted from piezoelectric sensors in a sitting position can be used as an alternative method for extracting biological signals.

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