Sheet-type Device for Unconstrained Heart Sound Measurement and White Noise Reduction by Wiener Filter

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Abstract – **Cardiac information is crucial in the diagnosis of cardiac diseases. Phonocardiography is widely used to obtain cardiac information at a medical institution. In this paper, we propose a sheet-type sensing device for unconstrained measurement of heart sound and filtering for heart sound extraction. We conducted a validation experiment with two males in their twenties and a female in her fifties. The experimental results show that the proposed system measured the first and second heart sound regardless of respiratory condition, thickness of bedding, and sleeping posture. Further validations are needed to establish that the proposed system is able to measure abnormal heart sounds which indicate the presence of cardiac diseases.**

I. Introduction

Cardiac information is crucial for the diagnosis of cardiac diseases. Three main methods are used to obtain cardiac information: electrocardiography (ECG), phonocardiography (PCG) and ballistocardiography (BCG). From the perspective of health care, unrestrictive monitoring of cardiac information at home is desirable. Methods for monitoring via BCG have been proposed: with RGB videos [1], bed-embedded methods [2], and radar-based methods [3-5]. Although studies on BCG are sparse, unconstrained monitoring via ECG [6,7] and unconstrained monitoring via PCG [8] have also been considered. In this paper, we focus on the use of PCG.

Concerning PCG, physicians at a medical institution comprehensively determine the need for a detailed examination (such as echocardiography) based on findings obtained by PCG. Since auscultation requires expertise, and requires that a stethoscope be pressed against the chest, it is not suitable for the monitoring of an individual's heart sound at home.

Methods have been proposed to automate the physician's determination of the need for a thorough examination from the use of PCG [9-17]. Ref. [9] applied an artificial neural network to classify normal and abnormal heart sound in the PASCAL database [16] and reported an accuracy rate of 0.97. Ref. [10] utilized a Gaussian-mixture model in approaching PCG data, as it appears in the PhysioNet database [17], and achieved an accuracy rate of 0.98.

If we can measure heart sounds at home, the algorithms introduced above can contribute to the early detection of cardiac disease, and can aid in the execution of telemedicine practices.

Ref. [8] proposes unconstrained PCG monitoring in the sitting position via the radar placed 20 cm from the participants. To cover the various sleeping postures and reduce the constraints related to the sensor position, this paper proposes a sheet-type sensing device for the unconstrained measurement of heart sounds during sleep, and filtering for heart sound extraction.

II. Proposed Method

An overview of the proposed method is shown in Fig. 1.

Heart sounds propagate in multiple directions on the bedding on which a person lies, and some of them propagate vertically downward into the bedding. Regardless of the sleeping posture, the heart sound can be measured under the bedding in order to achieve a certain degree of adhesion to the device. However, the forces under the bedding include the weight of the bedclothes itself as well as the weight of the human body, so it is necessary to measure only the variable force due to heart sound. The structure of a sheet-type device for heart sound measurements is described for this purpose.

The device consists of a chest piece of a stethoscope, a silicone tube, a pressure sensor, an extended polystyrene (EPS) spacer, and two polyvinyl chloride (PVC) boards. The chest piece is connected to one end of the silicone tube and

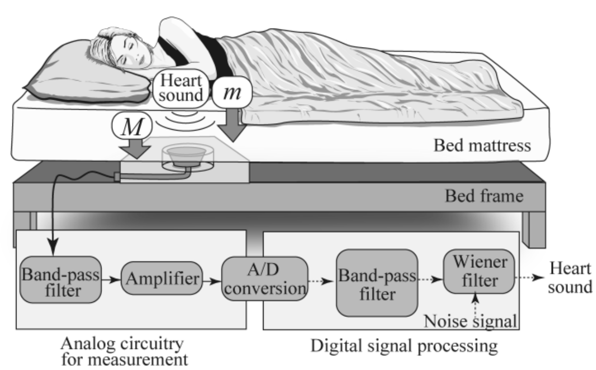


Fig. 1: Proposed System.

the pressure sensor is connected to the other end. To install them into the EPS spacer, grooves are dug on the surface of the EPS spacer. These are attached to the groove dug into the top of the EPS spacer. The chest piece is placed so that it protrudes slightly from the surface of the EPS spacer, and the silicone tube is completely embedded in the groove. These are sandwiched between two PVC boards from the top and bottom. The PVC plate attached to the top surface has a circular hole at the position of the chest piece.

This device is placed under the bedding on which an individual lies. Let M be the mass of the bedding and m be the mass of the human. Then a constant force $(M + m)\cdot g$ is applied to the device, where g is a gravitational acceleration. In addition, a variable force – due to the beating of the heart – is also applied to the device. The constant force $(M + m)\cdot g$ is mainly applied to the PVC plate and supported by the EPS spacer, while the varying force created by the beating of the heart is applied to the diaphragm of the chest piece. The varying force generates pressure changes in the rubber tube to which the stethoscope is attached. The pressure sensor measures the pressure changes, with output communicated via electrical signals.

To remove DC components and avoid aliasing during A/D conversion, the output signal from the pressure sensor is processed by an analog bandpass filter. The lower cut-off frequency f_{low} is set sufficiently low so as not to attenuate the heart sound. The upper cut-off frequency is set to a Nyquist frequency $f_s/2$ with respect to the sampling frequency f_s at A/D conversion. The pressure changes caused by the heart sound is minute, so the signal is amplified by a non-inverting amplifier with gain G. The amplified signal is then converted into a discrete signal, with a sampling frequency of f_s.

The signal not only contains the heart sound component, but the pulse component as well. In order to extract the heart sound component, a digital bandpass filter with a pass frequency band of 20Hz to $fs/2$ Hz is applied. In addition, the signal contains white noise, as well as the heart sound component; as a result, the Wiener filter is applied. The noise signal of the Winner filter is measured when the bedding is unoccupied.

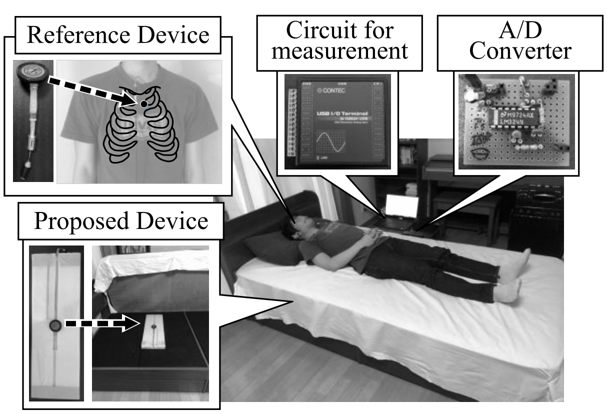


Fig. 2: Experiment environment.

III. Validation Experiment with Participant

In this chapter, we describe the validation experiment undertaken to evaluate the robustness of the proposed system in terms of sleeping posture, the distance between the sensing device and the human, and the presence of breathing.

A. Experimental Environment

Fig. 2 shows the experimental environment. This experiment is conducted in an ordinary room to assume a real living environment. The participant is asked to fix a Littmann® Master Cardiology™ Stethoscope (3M Company) to the right margin of the second intercostal sternum, using a corset as a reference. The proposed device is placed 60 cm from the head side under the mattress so that it is placed underneath the participant's heart. Both the output signals of the proposed device and the reference are filtered and amplified by the circuitry for measurement. The filtered signals are converted to discrete signals by A/D converter AI-1608AY-USB (CONTEC CO., LTD.) and uploaded to the computer. The sampling frequency f_s is set to 2kHz and the measurement time is 20 seconds. The lower cut-off frequency f_{low} is set to 0.008Hz. The white noise is removed via a Wiener filter in which the noise signal was measured in advance (when the bed was unoccupied).

B. Experimental Procedure

In terms of experimental conditions, five thicknesses, three sleeping postures and two respiratory conditions were set up. The thickness is the number of stacked mattresses, which is 8 cm in depth for each (t0: no mattress, t1: one, t2: two, t3: three, t4: four). Fig. 3, Fig.4 and Fig. 5 show the experiment settings of (t0), (t2), and (t4), respectively, in the supine position. The three sleeping postures are the supine, the right-lateral and the left-lateral positions. The validation of prone position is not feasible since the reference is attached to the chest. The two respiratory conditions included breathing normally and holding one's breath during measurement. Therefore, a total of 30 experimental conditions were established (as a combination of the three conditions).

The participants were two males in their twenties and a female in her fifties. For each of 30 experimental conditions, five data points were obtained from a participant, and one data point were obtained from the other two participants Thereby, a total of 210 data points were acquired.

C. Evaluation Criteria

For 30 experimental conditions, frequency spectra of the proposed system, along with references, are calculated for five trials. The frequency spectra are obtained by discrete Fourier transform. In order to take into account the minor spectral changes of wide frequency ranges, the logarithmic spectra is taken and the correlation coefficient between the proposed system and the reference is calculated. Then, the average and standard deviation of the correlation coefficients are calculated.

Fig. 3: Experiment settings in case of t0.

Fig. 4: Experiment settings in case of t2.

Fig. 5: Experiment settings in case of t4.

IV. Results

Fig. 6 shows an example of the time series from the proposed system, along with the reference, in case the participant stopped breathing. For all combinations of the thickness and sleeping posture, one example is randomly listed from the five trials. The black line is the output signal from the bandpass filter of the proposed system, while the gray line is that of the reference. The amplitudes differed depending on the combination of the two conditions, but there was no clear trend with respect to both the thickness and the sleeping posture. A comparison of the amplitudes of the left-lateral position shows that the amplitudes were similar across the five thicknesses.

Fig. 7 shows an example of the time series of the participant breathing normally. As confirmed in the case where the participant stopped breathing as well, it can be seen that the time series of the proposed system show variations – variations which appear to be the first and second heart sound.

Fig. 8 shows the signals before being processed by the Wiener filter (gray line) and after being processed by the Wiener filter (black line). Fig. 8 (a) and Fig. 8 (b) are examples in which white noise is removed comprehensively and where white noise remains after being processed by the Wiener filter, respectively. In Fig. 8 (a), it can be seen that the noise component – which is considered to be the white noise – became considerably smaller after the application of the Wiener filter. In addition, and as indicated in Fig. 8 (b) as well, it was found that the white noise was eliminated by Wiener filter. However, the filtered signal in Fig. 8 (b) appears to have more white noise left in it, compared to the filtered signal in Fig. 8 (a).

With respect to the cases where participants stopped breathing, Table I (a) shows the correlation coefficients without the application of Wiener filter and Table I (b) shows those with the application of the Wiener filter. Comparing Table I (a) and Table I (b), it can be seen that applying the Wiener filter increases the correlation coefficient for (t2), (t3), and (t4).

With respect to the cases where participants breathed normally, Table II (a) shows the correlation coefficients without the Wiener filter, and Table II (b) shows those with the Wiener filter. In cases where participants breathed normally as well, the application of the Wiener filter increases the correlation coefficient for (t1), (t3), and (t4). Moreover, there is no significant difference between the two respiratory conditions after applying the Wiener filter, as shown in Table I (b) and Table II (b).

V. Discussion

From Fig. 6 and Fig. 7, we confirmed that the proposed device is able to measure the first and second heart sound through all thicknesses. This result suggests that the EPS spacers in the proposed device support a constant force by the bedding and the human, and that the varying force caused by heart sounds are measured by the pressure sensor through the diaphragm.

It was also found that the white noise was removed by the Wiener filter, but that the effectiveness varied depending on the condition. This may be because the signal measured beforehand was used as the noise signal of the Wiener filter in the verification experiment. Incorporating an additional pressure sensor to the proposed device for simultaneous noise measurement could resolve the problem.

TABLE I

(a) Participant stopped breathing and the Wiener filter is not applied.

	t0	t1	t2	t3	t4
supine	0.73±0.16	0.78±0.06	0.84±0.03	0.79±0.07	0.84±0.03
right	0.77±0.03	0.83±0.05	0.81±0.05	0.83±0.07	0.76±0.03
left	0.80±0.03	0.82±0.05	0.74±0.04	0.68±0.09	0.78±0.04

(b) Participant stopped breathing and the Wiener filter is applied.

	t0	t1	t2	t3	t4
supine	0.72±0.15	0.78±0.06	0.83±0.02	0.80±0.06	0.85±0.03
right	0.77±0.04	0.83±0.05	0.82±0.05	0.84±0.07	0.77±0.04
left	0.79±0.03	0.82±0.05	0.75±0.04	0.69±0.08	0.79±0.04

TABLE II

(a) Participant breathed and the Wiener filter is not applied.

	t0	t1	t2	t3	t4
supine	0.70±0.15	0.77±0.02	0.79±0.02	0.78±0.06	0.76±0.03
right	0.77±0.03	0.78±0.02	0.80±0.02	0.78±0.04	0.76±0.03
left	0.76±0.04	0.77±0.03	0.75±0.03	0.75±0.02	0.76±0.01

(b) Participant breathed and the Wiener filter is applied.

	t0	t1	t2	t3	t4
supine	0.69±0.15	0.78±0.02	0.79±0.02	0.79±0.06	0.77±0.03
right	0.76±0.03	0.79±0.02	0.80±0.03	0.79±0.04	0.77±0.03
left	0.75±0.04	0.77±0.02	0.75±0.03	0.76±0.02	0.77±0.01

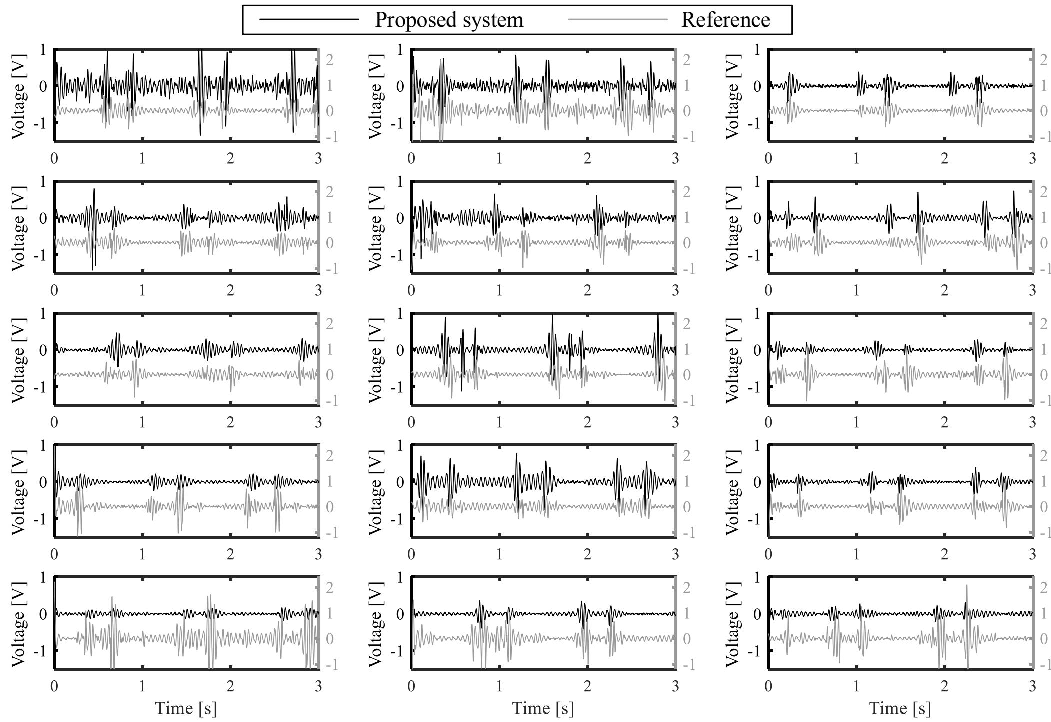


Fig. 6: Time series of proposed system and reference device in case the participant stopped breathing. The black and gray lines represent the proposed method and reference, respectively.

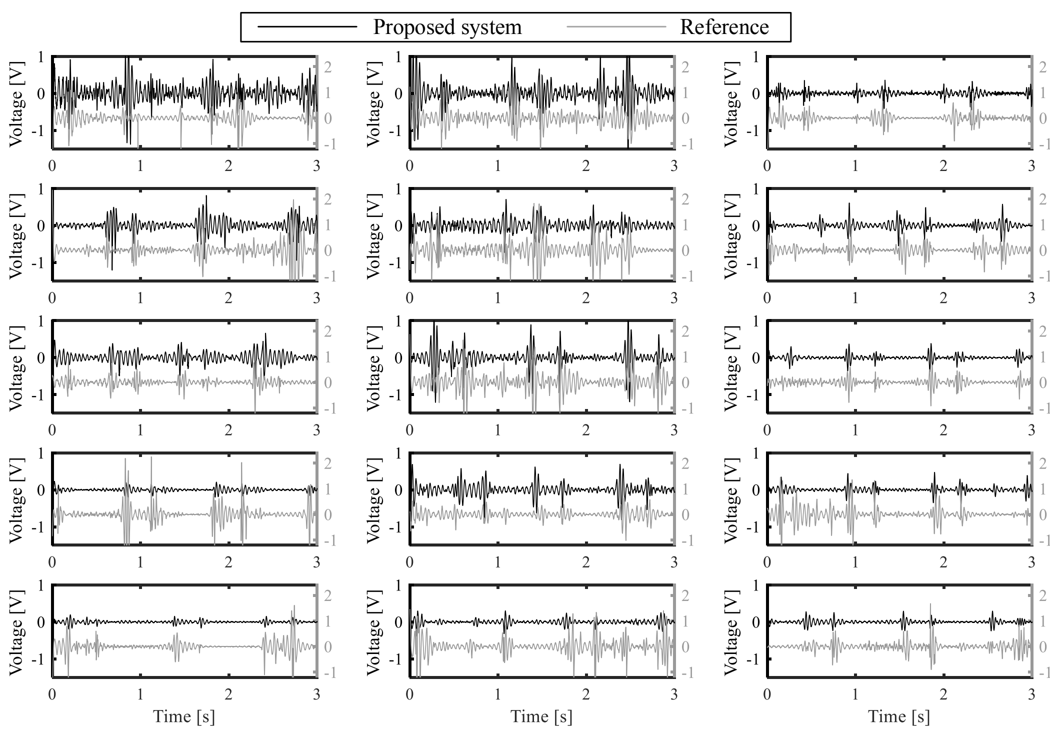
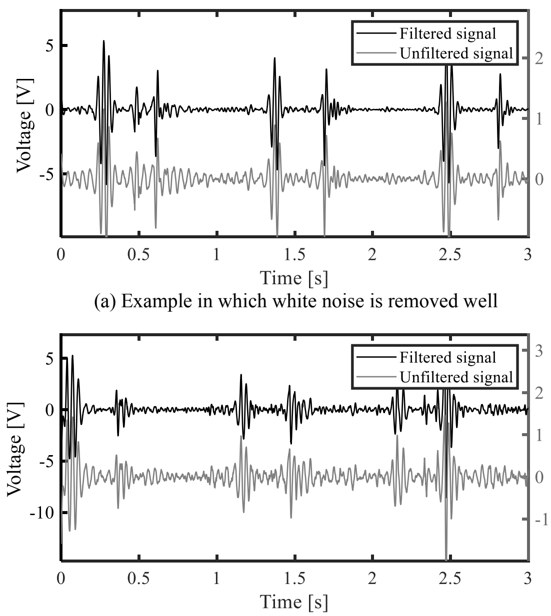


Fig. 7: Time series of proposed systems and reference devices in case the participant breathes normally. The black and gray lines represent the proposed method and reference, respectively.



(a) Example in which white noise is removed well

(b) Example in which white noise remains after Wiener filter

Fig. 8: Examples of signals before and after processing via the Wiener filter. The filtered signal in (b) appears to contain more white noise than the filtered signal in (a).

VI. Conclusions

In this paper, we proposed a sheet-type sensing device for unconstrained measurement of heart sounds and filtering for heart sound extraction. The experimental results show that the proposed system measured the first and second heart sound which was synchronized with the reference, regardless of mattress thickness, sleeping posture and respiratory condition.

In the validation experiment, however, a participant was a healthy male in his twenties. Further validations are needed to validate the proposed system is able to measure abnormal heart sound which indicates the possibility of cardiac diseases.

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References

[1] M. Poh, D.J. MacDuff, and R. W. Picard, “Non-contact, automated cardiac pulse measurements using video imaging and blind source separation,” *Opt. Express*, vol.18, no.10, pp. 10762-10774, 2010.

[2] A. Albukhari, F. Lima, and U. Mescheder, “Bed-embedded heart and respiration rates detection by longitudinal ballistocardiography and pattern recognition.” *Sensors* (Basel, Switzerland) vol.19, no.6, 1451. Mar. 2019.

[3] T. Sakamoto et al., “Accurate heartbeat monitoring using ultra-wideband radar,” *IEICE Electrinics Express*, vol.12, no.3, p. 20141197, 2015.

[4] V. L. Petrović, M. M. Janković, A. V. Lupšić, V. R. Mihajlović, and J. S. Popović-Božović, "High-accuracy real-time monitoring of heart rate variability using 24 GHz continuous-wave doppler radar," *IEEE Access*, vol.7, pp. 74721-74733, 2019.

[5] S. Wu et al., "Person-specific heart rate estimation with ultra-wideband radar using convolutional neural networks," *IEEE Access*, vol.7, pp.168484-168494, 2019.

[6] T. Schlebusch, “Unobtrusive health screening on an intelligent toilet seat,” *Acta Polytechnica*, vol.51, no.5, 2011.

[7] H. J. Lee, S. H. Hwang, S. M. Lee, Y. G. Lim, and K. S. Park, “Estimation of body postures on bed using unconstrained ECG measurements”, *IEEE J. Biomed. Health Inform.*, vol.17, no.6, Nov. 2013.

[8] C. Will, *et al*., “Radar-based heart sound detection,” *Sci Rep* 8**,** 11551, (2018).

[9] M. Nabih, E. A. El-Dahshan, and A. S. Yahia, “Heart diseases diagnosis using intelligent algorithm based on PCG signal analysis,” *International Journal of Biology and Biomedicine*, vol.2. pp.81-85, 2017.

[10] M. V. Shervegar, and G. V. Bhat, “Heart sound classification using Gaussian mixture model,” *Porto Biomed J.*, vol.3, no.1, Aug. 2018.

[11] M. Singh, and A. Cheema, “Heart sounds classification using feature extraction of phonocardiography cignal,” *International Journal of Computer Applications*, vol.77, no.4, pp.13-17, 2013.

[12] F. Li et al., “Feature extraction and classification of heart sound using 1D convolutional neural networks,” *EURASIP J. Adv. Signal Process.*, 59 (2019), Dec. 2019.

[13] G. Redlarski, D. Gradolewski, and A. Palkowski, “A system for heart sounds classification,” *PLoS One*, vol.9, no.11, Nov.2014.

[14] S. A. Singh, and S. Majumder, “Short unsegmented PCG classification based on ensemble classifier,” *Turk J Elec Eng & Comp Sci*, vol.28, pp.875-889, Jan.2020.

[15] B. Farzam, and J. Shirazi, “The diagnosis of heart diseases based on PCG Signals using MFCC coefficients and SVM classifier,” *IJISET*, vol.1, no.10, pp.654-659, Dec. 2014.

[16] P. Bentley, G. Nordehn, M. Coimbra, and S. Mannor. 2011. *The PASCAL Classifying Heart Sounds Challenge 2011*. https://www.peterjbentley.com/heartchallenge/

[17] A. L. Goldberger et al., “Physiobank, physiotoolkit, and physionet: components of a new research resource for complex physiologic signals,” *Circulation*, vol.101, no.23, pp.e215-e220, June 2000.