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Smart Sensing of Cardiovascular Physiological Information from Soles without Direct Skin Contact

Yuuki Kato[1], Masayuki Nambu[2], Masataka Imura[1], Yoshihiro Kuroda[1], and Osamu Oshiro[1]

Abstract— **This study proposes the development of a smart sensing measurement system using a mat-type sensor in order to obtain sole-electrocardiogram data from soles and toe-photoplethysmogram data from toes. In addition, the proposed measurement system can calculate pulse wave velocity from sole-ECG and toe-PPG. The subjective experiments revealed that the developed system can measure these parameters even when socks are being worn. Moreover, simultaneous measurements of systolic blood pressure and PWV determined by the developed system indicate a strong correlation was found between SBP and PWV. Therefore, this system can detect cardiovascular diseases or symptoms of common diseases and suggests the possibility to estimate the temporal changes in SBP without required compression using cuffs.**

I. INTRODUCTION

To realize the prevention or early detection of diseases, systems that measure daily physiological information over long periods must be developed. To this end, a measurement and evaluation technique called "smart sensing" has been attracting attention. This smart technique easily measures and evaluates physiological information by using various instruments. To introduce the smart sensing technology into the daily monitoring of physiological information, a ubiquitous and unrestrained measuring method is required [1].

In particular, many studies have been conducted to measure the cardiovascular physiological information. For example, blood pressure measurement in the toilet seats [2], electrocardiogram (ECG) and respiration measurements in the beds [3][4], and ECG measurements in the chairs [5] and bathtubs [6] have been studied. Furthermore, the estimation of blood pressure using the ECG and photoplethysmogram (PPG) without applying pressure to cuffs has been studied [7]-[12]. However, since previous studies did not resolve several issues such as giving a sense of restraint to the patients and scalability, practical applications have not yet been realized.

This study aims to develop a measurement system using a mat-type sensor to obtain physiological information from soles. In this study, ECG and PPG, which are indices used to measure cardiovascular systems, are adopted as physiological information. By simply placing a person's soles on the mat, ECG and PPG can be measured using the mat-type sensors. Therefore, measurements are possible without having to restrain a user by attaching a measuring device.

[1]Y. Kato, M. Imura, Y. Kuroda, O. Oshiro are with the Graduate School of Engineering Science, Osaka University, Toyonaka, Osaka 560-8531, Japan {ykato | imura | ykuroda | oshiro} at bpe.es.osaka-u.ac.jp

[2]M. Nambu is with Faculty of Biomedical Engineering, Osaka Electro-Communication University, Shijonawate, Osaka 575-0063, Japan nanbu at isc.osakac.ac.jp

II. MEASUREMENT SYSTEM

The structure of the proposed system for the measurement of sole-ECG and toe-PPG is shown in Fig. 1(a). The measurement system consists of a mat-type sensor, the analog circuits for the amplification and filtering of sole-ECG and Toe-PPG, an AD Converter, and a PC for digital filtering and calculation of PWV. The developed system can measure ECG and PPG using the mat-type sensor by simply placing soles on the mat.

A. Mat-type sensor

The mat-type sensor used in this study is shown in Fig. 1(b). It consists of fabric electrodes for measuring ECG, as well as an LED and a photodiode for measuring PPG. The fabric electrodes are active electrodes at the toes side and indifferent electrodes at the heel side. ECG is measured by either conductive coupling without using the paste or capacitive coupling, which can obtain ECG and PPG data even when socks are being worn. PPG is obtained by measuring the reflection intensity of the infrared light incident at the periphery. The LED's peak wavelength is 950nm. We place three LEDs to enable the measurement of PPG in case of socks being worn by the subjects. This three-LED method was proposed by Baek et al [13] to increase the incident light.

B. Analog circuit of ECG and PPG

The ECG circuit consists of a voltage follower for impedance transformation ($R_{in} = 1000\mathrm{T}\Omega$), an instrumentation amplifier, a high-pass filter ($f_c = 0.5$Hz), a noninverting amplifier, and a low-pass filter ($f_c = 40$Hz). The total gain of the ECG circuit is 80 dB. A driven-right-leg (DRL) circuit is introduced to reduce common mode noise by returning the biological signal to the body. The DRL circuit is effective even when socks are being worn [5].

The PPG circuit consists of an LED, a photodiode, a high-pass filter ($f_c = 0.3$Hz), a low-pass filter ($f_c = 40$Hz), and a noninverting amplifier. The total gain of the PPG circuit is 60dB.

C. Digital processing to calculate PWV

The analog signals obtained after filtering and amplification by the analog circuits are converted to digital signals using the AD converter (PicoScope 3224: Pico technology

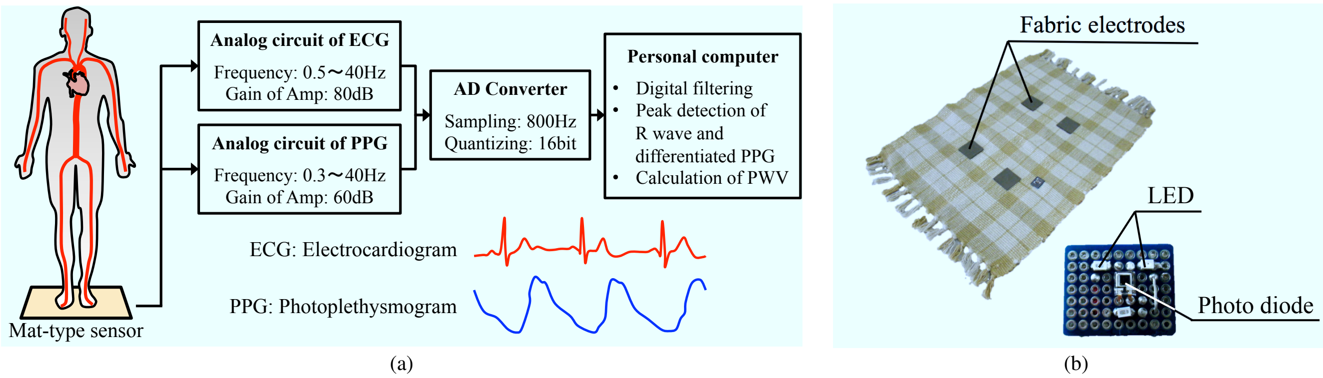


Fig. 1. Smart sensing of physiological information from soles: (a) Schematic diagram of measurement system (b) Mat-type sensor for measuring of cardiovascular physiological information.

corp.). The sampling frequency is 800 Hz with a 16bit resolution. Digital filtering and PWV calculation are performed on the PC. The PWV is calculated as follows:

1) The common mode noise (60Hz) is decreased by simple moving average (SMA).

$$X[n] = \frac{1}{M} \sum_{m=-L}^{m=L} x[n+m] \tag{1}$$

where $x[n]$ is the measured digital data, M is the number of moving averages, L is $(M-1)/2$, and $X[n]$ represents the data filtered by SMA. By subtracting the moving average signal $X[n]$ from the original signal $x[n]$, an SMA-type high-pass filter operation is performed.

2) The peak of the R wave of the ECG is detected by thresholding.

3) The differential waveform of PPG, $\dot{X}[n]$, is calculated using Equation (2).

$$\dot{X}[n] = \frac{X[n+1] - X[n]}{t_s} \tag{2}$$

where t_s is the sampling period. The peak of the differentiated PPG is detected using threshold processing.

4) The pulse transit time T_{PTT} is defined as the time from the peak of the R wave to that of the differentiated PPG. T_{PTT} is calculated for each beat, and the average value over 20 s is used for further calculations. When calculating the average value, the outliers from the standard deviation are removed and the average of T_{PTT} is again calculated to reduce the effects of noise.

5) The length of the blood vessel from the aortic root to the toes, L, is calculated by Equation (3).

$$L = 0.819H + 0.123 \ \ [\mathrm{m}] \tag{3}$$

where H is the height of the subject.

6) The pulse wave velocity (PWV) V_{PWV} is calculated by Equation (4).

$$V_{PWV} = \frac{L}{T_{PTT}} \tag{4}$$

In addition, in order to estimate to blood pressure using PWV, we consider the PWV-blood pressure (BP) model described in the next chapter.

III. PWV-BLOOD PRESSURE MODEL

Assuming that the pressure difference between the inside and outside of a blood vessel is zero and that the condition and thickness of a blood vessel are uniform, the pulse wave velocity V_{PWV} is represented by Equation (5), which is derived from the equation of continuity and conservation of momentum.

$$V_{PWV}^2 = \frac{dP/(dA/A)}{\rho} = \frac{A}{\rho}\frac{dP}{dA} \tag{5}$$

where A is the cross-sectional area of the artery, dA is the infinitesimal change in A, dP is the infinitesimal change in blood pressure, and ρ is the blood density. Furthermore, the relationship between arterial systolic blood pressure (SBP) P and artery diameter D is approximated by the following equation [14].

$$\ln\frac{P}{P_0} = \frac{\beta(D - D_0)}{D_0} \tag{6}$$

where subscript "0" represents the parameters of the blood vessel where a pulse wave has not arrived, β is a constant called the stiffness parameter, and β represents the hardness of blood vessel. The cross-sectional area of the artery is represented by $A = \pi D2/4$, From the above formulas, equation (7) is obtained.

$$\frac{dP}{dA} = \frac{\beta P}{2A} \tag{7}$$

From Equations (5) and (7), the relationship between P and V_{PWV}^2 can be given by Equation (8) as follows.

$$P = \frac{\beta V_{PWV}^2}{2\rho} \tag{8}$$

If the stiffness parameter β and blood density ρ are constant, Equation(8) can be expressed as shown in Equation(9).

$$P \propto V_{PWV}^2 \tag{9}$$

where P is proportional to the square of the V_{PWV}^2. In this study, we examine the possibility of estimating blood pressure using this relationship.

IV. EXPERIMENTS

We conducted the subjective experiments using the proposed mat-type sensor. Ethics approval for all of the experiments in this study was granted by the Ethics Committee of Engineering Science at Osaka University.

First, an example of the waveforms obtained by the proposed mat-type system is shown in Fig. 2. The measurement condition of Fig. 2(a) shows a subject sitting on a chair and wearing socks, while that of Fig. 2(b) shows a subject standing on the mat-type sensor and wearing socks. Fig.2 shows that the developed system can measure ECG and PPG from soles even when the subject is wearing socks. The developed system could detect the peak of the R wave and differentiated PPG, and could then calculate PWV. When the subject sat on a chair, the signal-to-noise ratio (SNR) values of the ECG and PPG were high. In this case, the developed system detected the T wave as well as the QRS complex. In contrast, when the subject stood on the mat-type sensor, the SNR of the ECG decreased. However, even in this case, the developed system could detect the peak of the R wave and differentiate PPG to calculate PWV. Therefore, the developed system can measure important cardiovascular information from the soles without direct skin contact irrespective of whether the subject is sitting on a chair or standing on the mat.

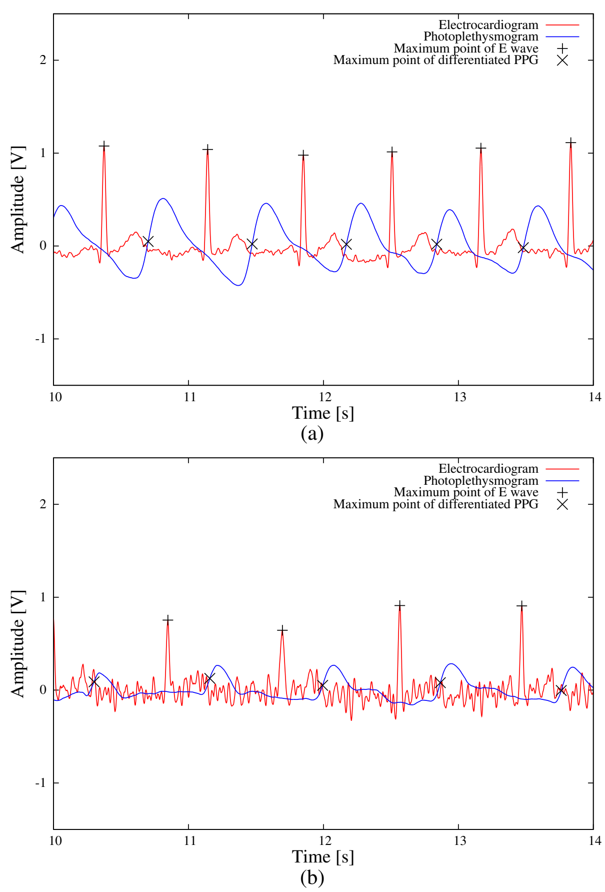


Fig. 2. Example waveforms obtained by the proposed system: (a) wearing socks and sitting on a chair. (b) Wearing socks and standing on the mat-type sensor.

Second, we conducted the simultaneous measurements of SBP using the sphygmomanometer (HEM-7020: Omron corp.) and PWV calculated by the developed system. The subject was a 24-year-old male with a height of 1.7m, and had no history of cardiovascular diseases. This experiment aimed to examine the possibility of detecting changes in blood pressure by PWV. SBP and PWV were measured every 2 min, and the total measurement period was 30 min. For the first 10 min, the subject was at rest, and then for next 10 min, to induce changes in blood pressure, the subject ran for 1 min before subsequent measurements were taken. For the last 10 min, the subject was again at rest. While measuring SBP and PWV, the subject sat on a chair. The experimental setup is shown in Fig. 3, and the result of applying an exercise stress test is shown in Fig. 4 and Table I. The horizontal and vertical axes of Fig. 4(a) and Fig. 4(b) represent the squared of PWV.

The results of Fig. 4 and Table I show that the temporal changes in SBP can be detected by PWV by using the proposed system. A strong correlation (r = 0.848, $p <$ 0.001) was found to exist between the SBP measured by the sphygmomanometer and the PWV measured by the proposed system. However, the measurement results contained several errors (standard error was 6.390 mmHg). This can be deduced from the following reasons.

- Because of motion artifacts due to small body movements, the peak position of the R wave or the differentiated PPG may shift from the original peak position.
- When measuring the PPG, the scattering and absorption of the incident and reflected light in the socks affect the peak position of the differentiated PPG.
- In the PWV-BP model, in fact, the blood vessels are not uniform and the hardness and internal diameter of the blood vessels change in vivo due to hormones.

However, although the estimation of SBP from PWV contained errors of the order of several millimeters of mercury, when signs or symptoms of shift to the severe high blood

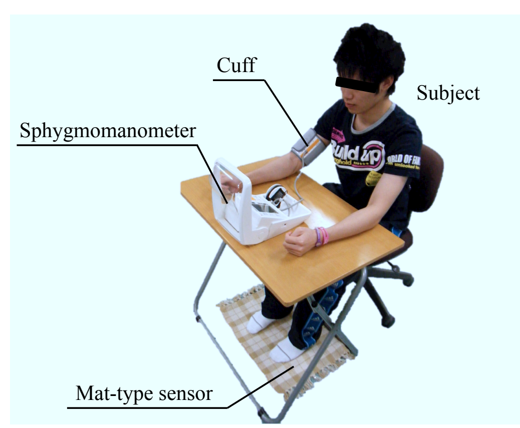
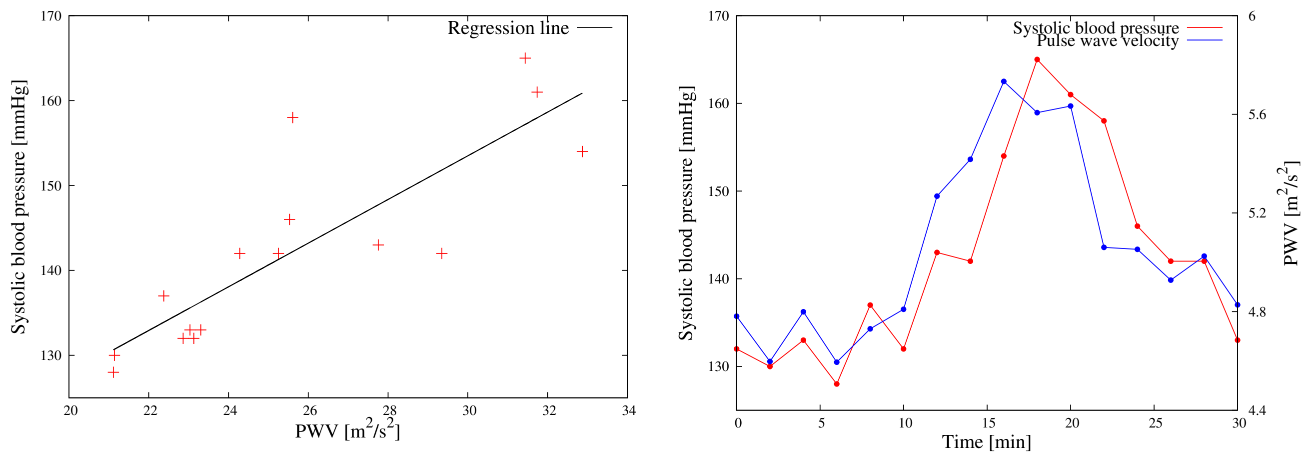


Fig. 3. Experimental setup.



(a) (b)

Fig. 4. Measurement data of SBP and PWV obtained using the exercise stress test: (a) The relationship between SBP and PWV. (b) Temporal variation in SBP and PWV.

TABLE I

CORRELATION BETWEEN SBP AND PWV BY EXERCISE STRESS TEST.

Subject	Height [m]	Correlation	Regression line (y:P, x:V_{PWV}^2)	SE [mmHg]
Male	1.700	$r = 0.848$ $p < 0.001$	$y = 2.571x + 76.374$	6.390

pressure (SBP $\geq$ 180 mmHg) occur, the temporal change in SBP could be detected using the proposed system. Therefore, these experimental results show that the proposed system has the potential to detect cardiovascular diseases or symptoms of commonly occurring diseases, and indicates the possibility of estimating the temporal changes in SBP without compression using cuffs.

V. CONCLUSIONS

This study proposed a smart sensing technique for the retrieval of cardiovascular information from the soles of the feet without direct skin contact. The subjective experiments revealed that the developed mat-type sensor and corresponding measurement system can measure ECG and PPG from soles when socks are worn. Moreover, by performing simultaneous measurements of systolic SBP using a sphygmomanometer and pulse wave velocity using a mat-type sensor, SBP and the squared of PWV were found be strongly correlated. Therefore, these experimental results show that this system has the potential to detect cardiovascular diseases or symptoms of commonly occurring diseases, and indicate that temporal changes in SBP can be estimated without compression using cuffs. If the proposed system is embedded into a weight scale, the smart sensing of physiological information may be realized in everyday life.

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