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Blood pressure estimation from pulse wave velocity measured on the chest

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Abstract— **Recently, monitoring of blood pressure fluctuation in the daily life is focused on in the hypertension care area to predict the risk of cardiovascular and cerebrovascular disease events. In this paper, in order to propose an alternative system to the existed ambulatory blood pressure monitoring (ABPM) sphygmomanometer, we have developed a prototype of small wearable device consisting of electrocardiogram (ECG) and photopelthysmograph (PPG) sensors. In addition, it was examined whether blood pressure can be estimated based on pulse wave transit time (PWTT) only by attaching that device on the surface of the chest. We indicated that our system could also sense tendency of time-dependent change of blood pressure by measuring pulse of vessel over the sternum while its propagation distance is short.**

I. INTRODUCTION

It is reported in World Health Statistics 2012 of World Health Organization (WHO) [1] that 29.2 percent of men and 24.8 percent of women have high blood pressure. Hypertension is universally-recognized as a predictor of cardiovascular, circulatory and cerebrovascular disease events. The adequate blood pressure control and early treatment of hypertension are required.

Usually, blood pressure tends to be measured at the specified times and circumstances. On the other hand, blood pressure actually fluctuates during a day. Ambulatory blood pressure monitoring (ABPM) equipment [2] can measure it at regular time intervals for 24 hours and sense some blood pressure behaviors that are correlated with disease risk. For example, non-dipper blood pressure pattern during sleeping which is difficult to identified by measurements in a day is reported as a predictor of cardiovascular and cerebrovascular risk. In addition, ABPM is useful to evaluate the effectiveness of antihypertensive medication after ingestion. However, ABPM equipment is the sphygmomanometer which measures blood pressure by applying external pressure to the arm in order to stop the flow of blood periodically in inflation of cuff. Because the repeated inflation has possibility to cause tissue damage, an interval of measurement is limited from 15 min to 30 min. This sampling frequency is too low to detect the fluctuation of blood pressure. Moreover, because of the pain, noises from cuff, and its appearance, users tend to hesitate to wear it in the daily life.

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Considering the problems of ABPM stated above, many researches has proposed the method to estimate the blood pressure from pulse wave velocity (PWV) or pulse wave transit time (PWTT) so far. PWV or PWTT are calculated from outputs of sensors such as electrocardiogram (ECG), phonocardiogram (PCG), or photopelthysmograph sensor (PPG) by wearing them at different points on the body at long interval of distance. For example, in the research of Lopez et al [3] and Ribeiro et al [4], they attached ECG electrodes to the chest and PPG sensor to the earlobe. McCombie et al. [5] and Parry et al [6] proposed the methods based on PWV which was calculated with the outputs of devices attached to the arm or finger. However in these methods, the relative positions of the heart and sensors change easily and outputs are sensitive to effect of hydrostatic pressure because joints (the wrist, arm, or neck) exist between two measurement points. Therefore, the model becomes complicated to improve the estimation accuracy.

In this paper, we present the result of study to estimate systolic blood pressure based on PWTT by utilizing a developed prototype of wearable sensing system. On the surface of that device, the PPG sensor is placed adjacent to ECG electrodes and is attached to the chest to measure pulse of blood vessel whose propagation distance is short.

II. SYSTOLIC BLOOD PRESSURE ESTIMATION FROM PWTT

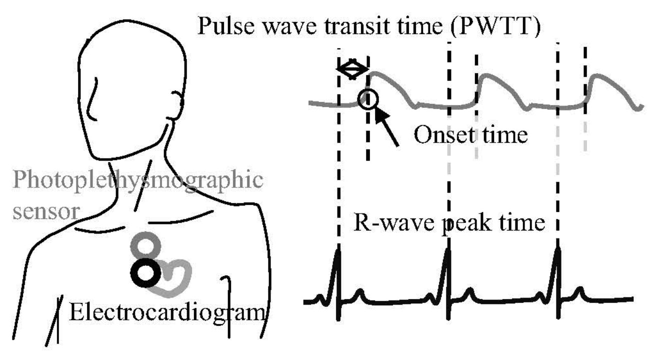


Figure 1 Pulse Wave and ECG

PWTT is computed using ECG and PPG sensor outputs. Typical output waveforms of both sensors at different positions are indicated in Figure 1. In this paper, PWTT is defined as difference between R-wave time on ECG data and lagging onset time of the pulse wave on PPG data.

Based on Moens-Korteweg equation [7] and equation which can model Young's modulus of vessel walls [7] with some simplifying assumptions, the relationship between

PWTT(T_{pwtt}) and systolic blood pressure (P) is shown as equation (1).

$$P = a\ln(\frac{b}{T_{pwtt}^{\ 2}}) \qquad (b = \frac{2*\rho*r*x^2}{E*h})$$

$$= a(\ln(\frac{1}{T_{pwtt}^{\ 2}}) + \ln(b)) = a\ln(\frac{1}{T_{pwtt}^{\ 2}}) + c \qquad (1)$$

Here, ρ is blood density, r is the radius of blood vessel, and x is the length of artery. E is Young modules of vessel walls with no pressure, h is thickness of artery and a is the parameter which correlates to variation of Young modules E depending on pressure. As far as the pulse wave is measured at the same position in a short period of time, b and c can be assumed to be constant terms. The variable term in equation (1) is expressed as Y as shown in equation (2).

$$\mathrm{Y} = \ln(\frac{1}{T_{pwtt}^{\ 2}}) \qquad (2)$$

III. EXPERIMENTS

We performed the experiments to investigate whether systolic blood pressure could be estimated from the data measured on the chest based on the relationship with PWTT (T_{pwtt}) in equation (1).

A. Measurement device

We have developed a prototype of wearable sensing device in Figure 2 [8]. Two metal electrodes are placed at 30[mm] intervals to detect the difference between electrical potential (voltage) on the surface of the body. The ECG voltage is amplified with gain (1360 or 2380) and the second-order low pass filter whose cutoff frequency is 17[Hz] is applied in the internal circuit. The reflected-type green light PPG sensor is placed between electrodes. It is reported to be less influenced by the tissue and vein region compared to infrared sensor [9]. In this experiment, sampling frequency of ECG sensor is 256[Hz] and frequency of PPG sensor is 64[Hz].The liner interpolation values are calculated from measured PPG data and the sampling frequency is increased to 256[Hz]. Outputs of both sensors are transferred to PC with Bluetooth.

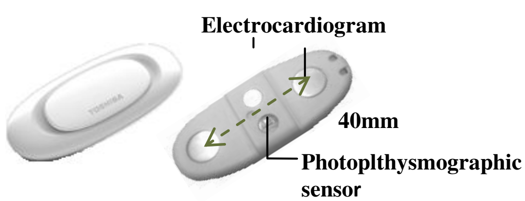


Figure 2 Prototype of wearable sensing device

B. Measurement position

The device is placed on the chest of participants as shown in Figure 3 by setting the one edge of the device 40[mm] beneath hollow of the throat. The electrodes are arranged along the sternum. This arrangement is similar to the NASA induction method of Holter ECG system. As an advantage of this arrangement, ECG signals are less affected by myoelectric signals of breast muscle compared to other positions in moving and breathing. Before reaching subcutaneous vascular at the measurement position, the pulse wave propagated through aorta, aortic arch, subclavian atery and internal thoracic artery. It is reported that the internal thoracic artery is similar to radial artery [10]. This device is attached to the body with conductive and adherent gel for ECG sensors.

From the view point of users, this system is simple enough to attach it regularly in the daily life. Moreover, when it is placed over the sternum, it is less likely to be peeled in spite of changes of a body shape

Figure 3 Wearable sensing device attached on the sternum

C. Experiment condition

We applied an exercise to 4 participants (3 males and 1 female from 26 to 42 years old) after getting informed consent. First, they were laid on their back in the bed and stayed still for 20 minutes. In the next step, in order to induce changes in blood pressure while preventing the influence of hydrostatic pressure, they had done an ergometer exercise by keeping the upper bodies supine as indicated in Figure 4 for 20 minutes. During exercise, the load of ergometer was set to 75[W] and the participants was asked to maintain a constant rate by using a metronome which is set to 66 bpm (one beat/pedal cycle). Each participant repeated these sets again with 20 minutes recovery time of tissues.

During experiment, the ECG signals and PPG signals were detected with the developed device on the chest and transferred to PC. In addition, the systolic blood pressures were measured with ABPM sphygmomanometer (A&D Medical TM-2431) at 2 minutes intervals on left arm. By using the data in the first set, the parameters of equation (1) for the systolic blood pressure estimation were calibrated. The ECG and PPG data collected in the second set was input to calibrated equation. The accuracy of estimation was evaluated by comparing against the actually measured blood pressures.

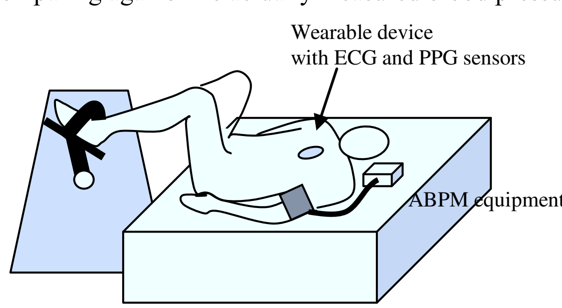


Figure 4 Experimental condition

IV. Result

Black and gray lines in Figure 5 and 6 show the output waveforms of ECG and PPG sensors in calmly supine and in the ergometer exercise. Because of body motion and diaphoresis in breathing and exercise, it seems that some noises and irregularly undulating baseline are superimposed. After implementing the digital filters, the R-wave peak time of ECG signals and onset time of PPG signals were calculated as follows.

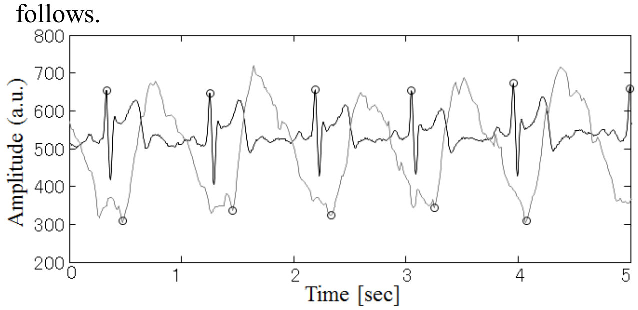


Figure 5 Output waveforms in calmly supine

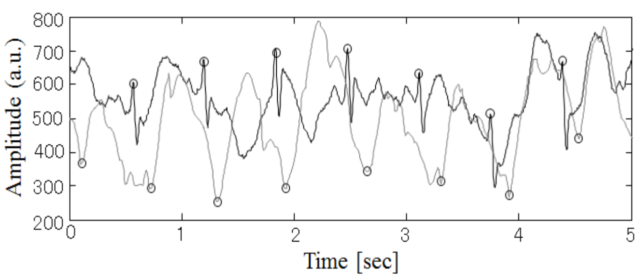


Figure 6 Output waveforms in exercise

A. Detecting the R-wave peak time of ECG signals

In order to stabilize the baseline of ECG signals, a high-pass IIR filter was implemented. The cutoff frequency was 0.3[Hz]. The same filter was applied in the forward and backward direction again to nullify the phase delays.

The local maximum values of the filtered signals *E(t)* in every 250[msec] periods were calculated. If the product of the derivative values: *E(t+R)-E(t)* and *E(t)-E(t-R)* (*R* is the predetermined step. In this case, R is 5 step) was smaller than threshold, *t* was defined as the R-wave peak time. Each computed R-wave peak time is plotted as points in Figure 5 and 6.

The output waveforms for 200 heart-beat cycles both in calm state and in the exercising state were extracted from the data of every participant for evaluation. By visual inspection, it is confirmed that 800/800(100%) peak times in the calm state and 791/800(98.9%) peak times in the exercise state could be detected with this method.

. B Detecting the onset time of PPG signals

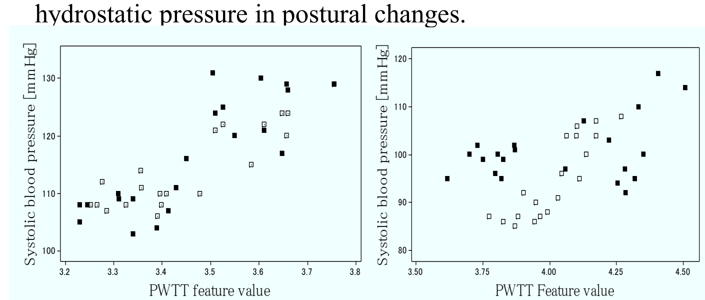
In order to reject noise signals which were caused by body movement, the band-pass filter was implemented in forward and backward direction.

We calculated the time *Td* as max differential filtered PPG signal during the periods between two consecutive R-wave peak times. Starting analysis from *Td* to the former R-wave peak time in the backward direction, when slope became under the predetermined threshold, it was calculated as an onset time.

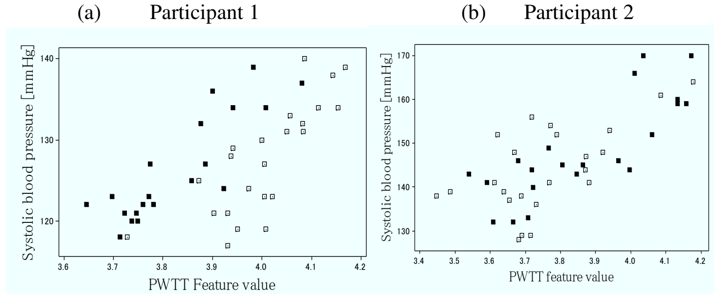
. C Calculating PWTT and systolic arterial pressure

PWTT (*Tpwtt*) at each heart beat was calculated as the difference between detected R-wave peak time and onset time. If *Tpwtt* was over 0.3[sec], that data was eliminated as an outlier. In addition, Because *Tpwtt* also varied periodically depending on breathing cycle, moving average deviations of *Tpwtt* were calculated.

In the experiment, it took about 60[sec] to measure blood pressure with sphygmomanometer. Therefore, median value of *Tpwtt* was calculated from the data in each 60[sec] interval as the representative values of PWTT. The representative PWTT computed from the data only in the first sets were input to the equation Y. Furthermore, by inputting the measured systolic blood pressure in the first set, parameters in equation (1) were computed with respect to each participant based on the method of least squares. Figure 7 shows the actual systolic blood pressures as reference values and calculated Y values. Data in the first sets for calibration are plotted as black points, and data in the second sets for evaluation are plotted as white points. Figure 8 shows the Bland- Altman plots [11] of reference blood pressure and estimated systolic blood pressure. It reveals a proportional bias in case of Participant 1. It indicates that estimation errors increase according as blood pressure rises. In case of participant 3, there is a fixed bias and it is supposed that parameters of equation (1) changed between the first set and the second set because of fine displacements of sensing artery and tissue or effect of hydrostatic pressure in postural changes.



(a) Participant 1 (b) Participant 2



(c) Participant 3 (d) Participant 4

Figure 7 Systolic blood pressures and PWTT feature value Y

Table 1 showed the results of the blood pressure estimation accuracy that were calculated with data in the second set. The blood pressure could be estimated with the error in the range of 5.87[mmHg] to 9.41[mmHg]. It was equivalent to the errors of the method in the previous researches, and our system attached on the chest could also sense tendency of time-dependent change of blood pressure.

Moreover, we calculated mean error and standard deviation in consideration of the ANSI.AAMI SP10:2002 standards for noninvasive blood pressure accuracy (±5 [mmHg] mean error, 8[mmHg] standard deviations). In respect of the mean error, the standard could not be still achieved. It is necessary to improve the prior digital processing of waveforms and the artery models for estimation.

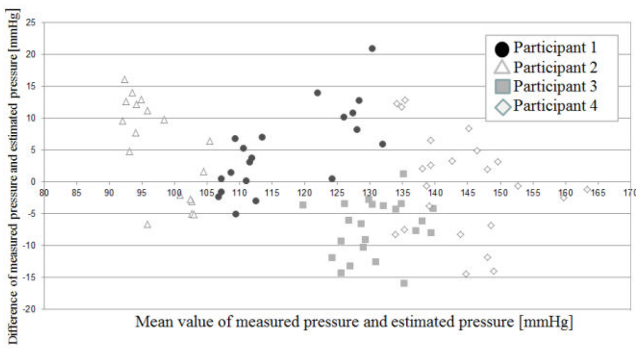


Figure 8 Bland-Altman plots between measured and estimated pressure

Table 1 Error of blood pressure estimation

	Mean error	Standard deviation	RMSE
Participant 1	7.57 [mmHg]	4.57 [mmHg]	8.78 [mmHg]
Participant 2	7.57 [mmHg]	5.70 [mmHg]	9.41 [mmHg]
Participant 3	5.17 [mmHg]	2.83 [mmHg]	5.87 [mmHg]
Participant 4	7.34 [mmHg]	3.82 [mmHg]	8.03 [mmHg]
Average	6.91 [mmHg]	4.23 [mmHg]	8.02 [mmHg]

V. Conclusion

In this paper, we have developed a prototype of small wearable device consisting of electrocardiogram (ECG) and photopelthysmograph (PPG) sensors, and examined whether blood pressure could be estimated based on pulse wave transit time (PWTT) by attaching it to the surface of the chest. Although the pulse propagation distance is short, the result showed that our system could also sense tendency of time-dependent change of blood pressure. On the other hand, there are still some problems which should be tackled with to achieve the standard of blood pressure monitor.

In the next step, it is considered that we can increase the estimation accuracy by improving the way to detect the onset time in PPG signals. Especially, when the reflected pulse wave had bigger amplitude, it became difficult to calculate the accurate onset time. It is required that filter processing should be improved also for artifact cancellation.

It is also necessary to examine the influence of definition PWTT. For example, the pre-ejection period (PEP) is also reported to be correlated with systolic blood pressure [12]. We continue the validation of definition by using data of participants, whose tissue conditions and ages are different. In addition, now we tackle with improvement of device itself. Sensor functions such as sampling frequency is going to be enhanced to detect the more accurate period.

In our approach, the parameters in estimation model are calculated from some distributed data. For higher-accuracy estimation, it is desirable that they are calibrated depending on user and placement of sensor. On the other hand, for daily use, it is important to simplify the calibration method or develop the correction algorithm with keeping consistent estimation performance, in the future.

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