Compensation of the Error Rate for the Non-invasive Sphygmomanometer System Using a Tactile Sensor

In-Cheol Jeong†, Yoo-Nah Choi* and Hyung-Ro Yoon**

Abstract - The Purpose Of This Paper Is To Use A Tactile Sensor To Compensate The Error Rate. Most Automated Sphygmomanometers Use The Oscillometric Method And Characteristic Ratio To Estimate Systolic And Diastolic Blood Pressure. However, Based On The Fact That Maximum Amplitude Of The Oscillometric Waveform And Characteristic Ratio Are Affected By Compliance Of The Aorta And Large Arteries, A Method To Measure The Artery Stiffness By Using A Tactile Sensor Was Chosen In Order To Integrate It With The Sphygmomanometer In The Future Instead Of Using Photoplethysmography. Since Tactile Sensors Have Very Weak Movements, Efforts Were Made To Maintain The Subject's Arm In A Fixed Position, And A 40hz Low Pass Filter Was Used To Eliminate Noise From The Power Source As Well As High Frequency Noise. An Analyzing Program Was Made To Get Time Delay Between The First And Second Peak Of The Averaged Digital Volume Pulse (Δt_{dvp}), And The Subject's Height Was Divided By Δt_{dvp} To Calculate The Stiffness Index Of The Arteries (Si_{dvp}). Regression Equations Of Systolic And Diastolic Pressure Using Si_{dvp} And Mean Arterial Pressure (Map) Were Computed From The Test Group (60 Subjects) Among A Total Of 121 Subjects (Age: 44.9±16.5, Male : Female = 40:81) And Were Tested In 61 Subjects To Compensate The Error Rate. Error Rates Considering All Subjects Were Systolic 4.62±9.39mmhg, And Diastolic 14.40±9.62mmhg, And Those In The Test Set Were 3.48±9.32mmhg, And 14.34±9.67mmhg Each. Consequently, Error Rates Were Compensated Especially In Diastolic Pressure Using Si_{dvp}, Various Slopes From Digital Volume Pulse And Map To Systolic -1.91±7.57mmhg And Diastolic 0.05±7.49mmhg.

Keywords: Compensation Of Error Rate, Digital Volume Pulse, Regression Analysis, Sphygmomanometer, Stiffness Index And Tactile Sensor.

1. Introduction

Use of mercury-based sphygmomanometers, the non-invasive blood pressure measuring method using mercury, is currently most widely used throughout hospitals and homes and it is expected to be limited by the anti mercury-use campaign supported by 1,400 hospitals and health institutions in the U.S.A. by the year 2005. Therefore, the need for the use of electrical sphygmomanometers is arising rapidly. At the present time, the best alternative for mercury-based sphygmomanometers seems to be an electrical sphygmomanometer using the oscillometric method. However, electrical sphygmomanometers show a rather high error rate in their accuracy in blood pressure measurement. Such a fact can be well illustrated by Ursino and Cristalli's study stating that non-invasive measurement of systolic, diastolic and mean blood pressure using the oscillometric method leads to maximal error rates of 10 to 15% [1]. The oscillometric method assumes that the maximum amplitude of vibration occurring at the cuff is mean blood pressure and determines the characteristic ratio having a certain ratio to the amplitude of vibration based on prior experiences. Then, the pressure calculated from this information results in systolic and diastolic blood pressure. Therefore, it contains many sources of errors.

First of all, it is a very challenging task to accurately estimate the maximum amplitude of vibration occurring at the cuff. The location of the maximum of vibration does not appear as one sharp peak but as a flat plateau. So, it should be clearly defined which point in this pressure range should be chosen as mean blood pressure. In addition, the maximum amplitude of vibration may occur when the value of the cuff pressure is between mean blood pressure and diastolic blood pressure. Such a maximum value is very similar to other values located at the plateau and it is quite difficult to measure it in a laboratory environment that is surrounded by various sources of noise. Second, the compliance between the cuff and the

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arm and the cuff size interact with each other in such a way that has a great impact on the shape of the envelope of the oscillometric pulse. Therefore, it has a significant impact on blood pressure measurement using the oscillometric method.

Finally, the characteristic ratio may vary 10 to 20% depending on the person. Such a variance is affected by the cuff volume, the artery compliance and the systolic fraction (SF) representing the ratio of delays of heartbeat and pulse wave. Therefore, it should be accompanied by a proper method to measure the condition of the artery wall in order to avoid errors in measurement [1].

The subject matter of this study is to compensate the error rate of the above mentioned sphygmomanometer using a tactile sensor. Therefore, in this study, we considered the fact that we may yield the factor that represents the stiffness of blood pressure using photoplethysmography. Using a tactile sensor, we measured a plethysmogram at the cuff and calculated and comparatively analyzed the stiffness indices resulting from the plethysmogram and photoplethysmography. Eventually, through our efforts, we intend to compensate the error rates of electrical sphygmomanometers using tactile sensors.

2. Methods

2.1. How to measure the condition of the artery wall.

The increase of artery stiffness due to aging leads to the increase in the pulse wave velocity (PWV) [4-6], and it also affects peripheral blood pressure of the arm and the overall shape of the volume pulse waveform [7]. The change in the overall shape partially leads to the increase in the stiffness of the main artery, which in turn leads to the increase in the PWV. As a result, it reduces the time required for the pressure waveform reflected by the peripheral circulation area to return to the main artery and the arm, so that the reflected wave may arrive early during a cardiac cycle [8]. Millasseau et al. measured the stiffness of the main artery indirectly by measuring the PWV between the carotid artery and the femoral region. SI_{DVP} is the index representing the stiffness of the main artery. The height (h) of the participant is divided by the time delay (ΔT_{DVP}) of the waveform reflected by the digital volume pulse (DVP) in order to yield SI_{DVP}, as shown in Fig. 1. Since the time delay of the DVP is influenced by the PWV, stiffness measurement using SI_{DVP} is directly related to stiffness measurement using the PWV (r = 0.617, P <0.01). Hence, if the PPG is used near the peripheral area, it might be possible to measure the stiffness of the artery with a non-invasive measuring method [9].

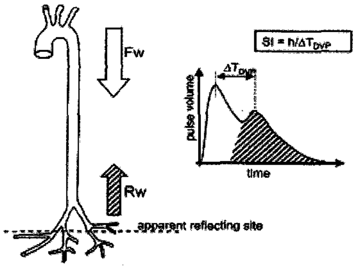


Fig. 1. Method for inducing SI_{DVP}.

2.2. Tactile Sensor

A tactile sensor is comprised of a feedback circuit and a sensor combining two units of dish-shaped piezoelectric ceramic transducers made of lead zirconate titanate. Its architecture allows one side of the sensor to be used as an ultrasonic transducer while the other side of the sensor should be used for picking up vibrations. The principle of a tactile sensor is based on the contact compliance method [10] and the phase shift method [11] as described in studies by S. Omata [12, 13]. As varying pressure is applied to the terminal of the driving piezoelectric element, the piezoelectric element starts vibrating freely and it delivers longitudinal waves. At this moment, if the vibrational frequency is identical to the unique resonance frequency, a fixed waveform will be generated within the sensor.

In a loaded condition, a tactile sensor detects movement occurring along the surface of the body due to blood flow by measuring the absorption coefficients between the surface of the PZT and the body. When blood flow pressurizes the surface of the PZT, the junction between the surfaces of the PZT and the body will be hardened and as a result, the absorption coefficient increases. The phase difference of varying voltages of input (ultrasonic transducer) and output (vibration pickup) is determined by the absorption coefficient of the body surface and such a phenomenon can be described by a simple forced vibration resonance model presented in the following equation.

$$m_0 \frac{d^2\delta}{dt^2} + r_0 \frac{d\delta}{dt} + k_0\delta = f_v \sin(\omega t) + F_s \sin(\omega t + \phi) \quad (1)$$

From the above equation, r_0 denotes the equivalent impedance of the sensor system, ω denotes the angular velocity, F_v is the magnitude of the driving force of the ultrasonic transducer, in other words, $F_v sin(wt)$ represents the input voltage. F_s is a function for the absorption coefficient of the body's surface and the distance between the sensor and the body's surface. ϕ is a function for the distance. The response of this resonance system, δ can be simplified into the following form.

$$\delta = A\sin(\omega t + \theta) \tag{2}$$

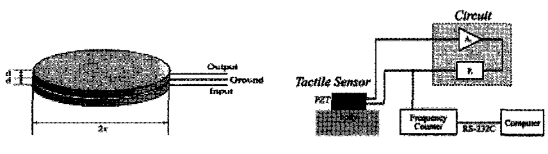
where we have the following relationships.

$$\theta = \tan^{-1}\frac{r_0\omega - (k - m\omega^2)\mu}{(k - m\omega^2) + r_0\omega\mu} \tag{3}$$

$$\mu = \frac{F_s \sin\phi}{F_v + F_s\cos\phi} \tag{4}$$

Therefore, F_s, the function for the absorption coefficient can be calculated by measuring the phase difference, θ for a given angular velocity, ω.

The phase difference appearing in Equation (3) can be measured as a high signal-to-noise ratio by the new phase shift circuit [14] shown in Figure 3. The vibration pickup detects vibrations and converts it to voltages. Afterward, the vibration is fed back to the driving PZT through an amplifier and a phase shift. The phase shifts of varying voltages through the sensor are expressed as θ_1 whereas the phase shifts occurring in the phase shift circuits are expressed as θ_2. In this system where $\theta_1 + \theta_2 = 0$ (Fig. 4), as amplification takes place, the phase shift circuit forces the sensor to follow its resonance frequency, f_0. When the sensor is loaded, the resonance curve shifts according to the acoustic impedance of material and the sensor starts resonating at the new frequency $f_c = f_0 - \Delta f_c$ that satisfies the relationship where $\theta_1 + \Delta\theta_1 = -(\theta_2 + \Delta\theta_2)$. Especially, this circuitry compensates for the phase shift of the sensor by adjusting the resonance frequency. This system has an excellent signal-to-noise ratio for its sensor part since it operates as a velocity resonance system by combining resonances of the mechanical vibration of the PZT transducer and the electrical circuit.



(a) (b)

Fig. 2. Basic architecture of a tactile sensor.
(a) Architecture of a tactile sensor
(b) Architecture of a sensor system

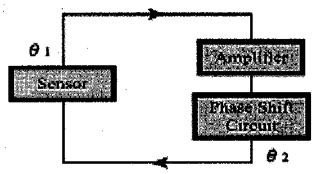


Fig. 3. Input/Output relationship between the basic components comprising a tactile sensor.
θ_1: Output Phase of a Sensor System
θ_2: Phase of the Feedback Circuit

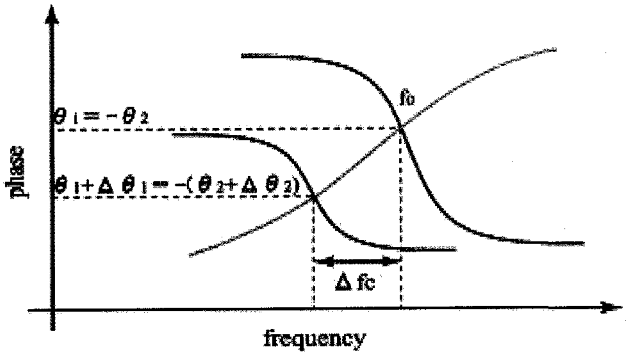


Fig. 4. Resonance frequency shifting according to the phase shift principle.

3. Results

3.1. Experimental Condition

We conducted an experiment to compensate for the error rate of an electrical sphygmomanometer. A total of 121 participants (age: 44.9±16.5, Male : Female = 40:81) aged from 15 to 85 were tested in this experiment. The height of participants ranged between 139 and 186cm (161.1 ± 8.9cm). The circumference of the upper arm ranged from 18 to 32.5cm (26.1 ± 3.0cm) and no participant had an upper arm whose circumference was larger than 35cm.

Eighty participants were measured to be 25 to 35cm (66%) while 41 participants were measured to be less than 25cm (34%). Among the participants, there were 5 cardiovascular disease patients, 1 kidney disease patient, 1 cancer patient, 19 and 9 patients who were diagnosed with hypertension or hypotension at least once in their life, respectively, 1 pregnant patient, 3 otorhinolaryngologial patients, and finally, 2 diabetic patients. To summarize, a total of 36 (30%) participants had special medical records worthy of notation. Participants were asked to sit with proper posture during testing. Prior to conducting the experiment, the heights of participants were measured whereas the circumferences of the upper arms were measured at the most bulging parts of their upper arms.

Table 1. Classification of Participants

Classification	Hospital "S" in Seoul		University "A" in Suwon		Church "M" in Seoul		Total
	Male	Female	Male	Female	Male	Female	
Normal	13	9	6	26	6	25	85
Hypertension	6	8	2	0	7	13	36
[illegible]	[illegible]	[illegible]	8	26	13	38	121

Table 1 summarizes the classification of participants.

First, we measured the blood pressure of each participant using a stethoscope in order to obtain the reference value for calculating the error rate of the electrical sphygmomanometer. Prior to use of the stethoscope, two measurers, with 4 years of experience at the medical school of Hanyang University, simultaneously measured each participant's blood pressure 30 times and calculated the error rate between the 2 sets of data provided by each measurer in order to confirm their correlation. A mercury-based sphygmomanometer (YAMASU-600, Japan) and a stethoscope that is audible to two measurers at the same time (KaWe, REF 43650, Germany) were used by two measurers to simultaneously measure a participant's systolic and diastolic blood pressure without sharing any information between the measurers [15].

Second, each participant's pulse wave was measured for 15 seconds using a tactile sensor on the location of their pulse on his or her wrist. Figure 5 represents the apparatus to measure a participant's pulse wave.

Third, a plethysmography measuring apparatus using the microprocessor MSP430F149 was mounted on a finger of each patient's arm (always the same arm) and the pulse wave was measured for 15 seconds.

Fourth, we removed any pressure remaining in the cuff of an electrical sphygmomanometer (Welch Allyn, NIBP Developer's Kit) and surrounded each patient's upper arm to measure his or her systolic and diastolic blood pressure. A type of bladder (Welch Allyn, 97S657) for the cuff was chosen to fit the range of circumference of each patient's upper arm, that is, 25.3-34.3cm [9].

Fifth, the measurement procedures shown above were repeated three times. In order to eliminate the effect of the auscultatory gap, additional pressure of 30mmHg was added to the maximum pressure before we took our second measurement with a mercury-based sphygmomanometer. Likewise, additional pressure of 30mHg was added in our third measurement. After using the sphygmomanometer, we let it rest for at least 1 minute before taking the next measurement, so that the blood vessel and flow that were once affected by the cuff pressure could recover their normal conditions. Each measurement period for each patient was set to be within 10 to 30 minutes.

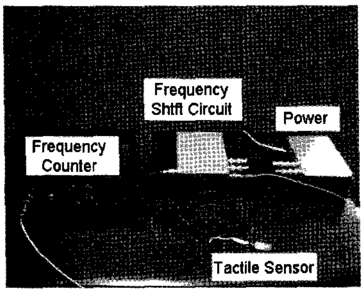


Fig. 5. Pulse wave measuring apparatus using a tactile sensor.

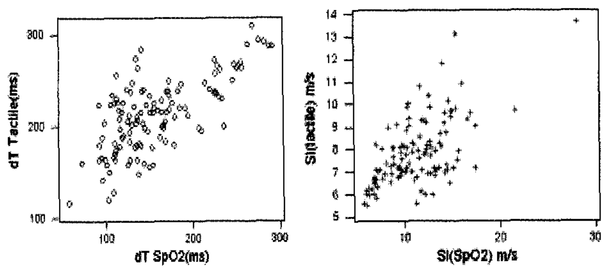
3.2. Error Rates of an Electrical Sphygmomanometer

Three pairs of stethoscopic measurement data obtained from each patient and three pairs of electrical sphygmomanometer measurement data were averaged for both systolic and diastolic blood pressure. The stethoscopic measurement data were subtracted from the electrical sphygmomanometer measurement data to yield the mean error rate of blood pressures measured on a total of 121 participants.

The calculated error rate is 4.62 ± 9.39mmHg for systolic blood pressure and 14.40 ± 9.62mmHg for diastolic blood pressure. These result in far less than 5 ± 8mmHg, which is governed by the specification set forth in AAMI SP10.

3.3. Artery Stiffness Measurement

In consideration of its effect on the artery stiffness, we used a tactile sensor to compensate for the error rates. In order to use a tactile sensor, which is a rather new method, we first measured the time interval between the starting point and the second peak for each digital volume pulse, which was obtained using a tactile sensor and photoplethysmography, the existing method used to measure the index of artery stiffness. This procedure was repeated three times for each of the 46 participants and based on this result, we measured the time delay (ΔT_{DVP}) for the direct and reflected waves. The height of a participant was divided by ΔT_{DVP} to yield the stiffness index (SI_{DVP}). From this result, we verified that there exists a correlation between the values obtained with photoplethysmography and a tactile sensor. (Fig. 6) We have $r = 0.695$ and $p < 0.01$ for ΔT_{DVP} and $r = 0.617$ and $p < 0.01$ for SI_{DVP} so we were able to verify that the correlation holds true and these values are statistically significant. Therefore, we may conclude that we may obtain the artery stiffness using a tactile sensor.



(a) (b)

Fig. 6. ΔT_{DVP} and SI_{DVP} measured with photoplethysmography and a tactical sensor
(a) ΔT_{DVP} and photoplethysmography
(b) SI_{DVP} and photoplethysmography

3.4. Compensating for the error rate

We classified a total of 121 participants into an experiment group of 60 persons and a control group of 61 persons in such a way that normal and high blood pressured persons were evenly divided into both groups. In the experiment group, we performed a regressive analysis on diastolic and systolic blood pressure using the statistical tool, Minitab. Table 2 represents the classification of the experiment and control group.

3.4.1. Compensating for the error rate of systolic blood pressure

The value of systolic blood pressure ($SBP_{Auscultatory}$) measured using a stethoscope was chosen as a dependent variable and then a regression analysis using the Minitab was performed on the value of diastolic blood pressure measured using an electrical sphygmomanometer, the stiffness index ($SI_{Tactile}$) measured with a tactile sensor and the mean artery pressure (MAP).

The regression equation for systolic blood pressure is as shown in Equation (5). The value of R-Sq is 78.1% and $p < 0.01$ and we conclude that it is statistically significant.

$$SBP_{Auscultatory} = 12.6 + 0.851 SBP_{Oscillometric} + 1.12 SI_{Tactile} + 0.020 MAP \quad (5)$$

We applied the above equation to the control group and observed that the error rate improved from 3.48 ± 9.32mmHg to -1.91±7.57mmHg.

Table 2. Classification of the Experiment Group and the Control Group

	Experiment Group		
	Normal	Hypertension	Total
Male	13	7	20
Female	29	11	40
Total	42	18	60
	Control Group		
	Normal	Hypertension	Total
Male	12	8	20
Female	31	10	41
Total	43	18	61

3.4.2. Compensating for the error rate of diastolic blood pressure

The value of diastolic blood pressure ($DBP_{Auscultatory}$) measured using a stethoscope was set as the final response value and a regressive analysis was performed on the value of blood diastolic pressure ($DBP_{Oscillometric}$) measured using an electrical sphygmomanometer and the stiffness index ($SI_{Tactile}$) measured with a tactile sensor and the mean artery pressure (MAP).

The regression equation for diastolic blood pressure is as shown in Equation (6). The value of R-Sq is 78.7% and $p < 0.01$.

$$DBP_{Auscultatory} = 16.9 - 0.444 DBP_{Oscillometric} + 1.13 SI_{Tactile} + 1.01 MAP \quad (6)$$

We applied the above equation to the control group and observed that the error rate improved from 14.34 ± 9.67mmHg to the compensated value of 0.05±7.49mmHg.

4. Conclusions

The purpose of this study was to measure the artery stiffness using a tactile sensor and to compensate for the error rate of an electrical sphygmomanometer based on the results. The results are summarized as follows.

(1) After measuring blood pressure with the electrical sphygmomanometer manufactured by Welch Allyn Co., we observed a very high error rate. From this, we convinced ourselves of the current standing of electrical sphygmomanometers.

(2) Our results demonstrate a correlation with the existing method proposed by MIllasseau et al. [10] to measure the artery stiffness, the main source of errors in the oscillometric method used by most of sphygmomanometers. We successfully verified that the stiffness index measured using a tactile sensor as according to the method proposed in this paper is statistically significant.

(3) The stiffness index yielded from the measurements using a tactile sensor was used to improve very high error rates unavoidable when using electrical sphygmomanometers. Consequently, the error rate improved from 3.48mmHg to -1.91mmHg on average while its distribution also improved from 9.32mmHg to 7.57mmHg. Diastolic blood pressure, which shows especially a very high error rate in an electrical sphygmomanometer, also improved from 14.34±9.67mmHg to 0.05±7.49mmHg on average, and we believe that we have successfully established a solid foundation for developing a more accurate all-in-one type of electrical sphygmomanometer.

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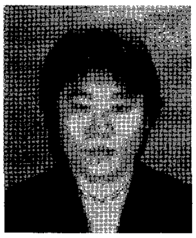

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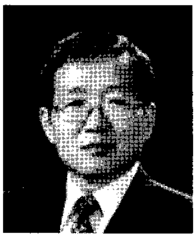

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