Noninvasive Blood Pressure Measurement Using PVDF Fibers Fabricated by NFES and A Photoplethysmography Sensor

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***Abstract*—Because the global population is aging, a long-term physiological monitoring system for senior citizens must be developed. Among the current physiological monitoring systems, that used to monitor blood pressure is one of the most critical. Conventional blood pressure monitors, such as mercurial and electric sphygmomanometers, all apply compression to arteries using an inflatable cuff. For users monitoring their blood pressure in the long term, artery compression may cause discomfort during measurement; moreover, it cannot be used to continuously monitor blood pressure. Estimating blood pressure through a cuffless method has become increasingly valuable and is being widely researched. This study proposes a method for estimating blood pressure by using the pulse transit time (PTT) and epidermis strain. First, thick-walled theory and pulse wave velocity were used to establish an equation for estimating blood pressure. In the proposed method, a piezoelectric strain sensor and photoplethysmography sensor are worn continuously on the wrist above the radial artery and on the fingertip, respectively. Because the two devices are at a fixed distance, a time delay results between the obtained blood pressure pulse signals and photoplethysmography signals. The PTT and blood pressure pulse signals are used to calculate the blood pressure by using the estimation equations.**

***Index Terms*—pulse transit time, pulse wave velocity, blood pressure pulse signals, Polyvinylidene fluoride**

I. Introduction

Conventional devices for measuring blood pressure can be distinguished as either invasive or noninvasive. Invasive blood pressure measurement directly measures continuous and absolute intravascular pressure and is only used in specific parts of operations. Noninvasive blood pressure measurement can be divided into the auscultatory method, such as in mercury-type sphygmomanometers, and the oscillometric method such as in electronic blood pressure monitors. However, these two types of sphygmomanometers require a cuff to measure blood pressure, which may cause discomfort during measurement in long-term use. At present, continuous blood pressure measurement is primarily optical and mechanical based, and both forms are still in the research stage, domestically and internationally. Optical measurement is performed by lighting skin tissue and blood vessels to observe how the pulse changes the flow of blood in the vessels; the reflected light varies in accordance with the amount of blood.

One method for continuously measuring blood pressure is with a tension transducer, which measures the force and deflection perpendicular to the skin surface. The tension transducer is the original model of tonometry [1]. To verify the accuracy of tonometry, JENTOW was used to measure the blood pressure of 20 normotensive subjects and 10 hypertensive patients. Finally, the blood pressure pulse waveform measured by tonometry was compared with invasive blood pressure measurements, revealing that tonometric measurement is medically acceptable and reliable [2]. The other method is placing two photoplethysmography (PPG) sensors at different places to measure the pulse wave velocity (PWV) caused by blood pressure pulse signals. One sensor is placed on the wrist above the radial artery; the other sensor is placed on the little finger. Because blood pressure pulse signals propagate along each artery, blood pressure can be measured through PWV by using the Moens–Korteweg equation. Moreover, the blood pressure obtained by measuring the PWV can be corrected according to the change in hydrostatic pressure [3]. Similarly, electrocardiography (ECG) measured by a body-sensor network system and PPG signals can be used to calculate the pulse transit time (PTT) and heart rate, from which the blood pressure can be derived [4]. Another proposed method employs a novel skin curvature (SC) sensor, which uses SC signals and ECG to obtain the PTT and thereby clarify the relationship between the change in blood pressure and change in SC signals. The SC sensor consists of a magnetoelastic bilayer partly enclosed by a coil. Bending the bilayer causes changes in magnetic permeability, which are measured by the coil

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[5]. Similarly, pulse strain signals can be measured with the strain sensor to obtain the PTT. Finally, the relationship between the PTT and blood pressure can be determined using the law of energy conservation [6]. In this study, the strain on the skin's surface that is caused by the pulse was used to obtain a mechanical model of a strain sensor passing through the radial artery to the surface of the skin at the wrist. Piezoelectric film composed of highly molecular polymer can be used to obtain more innovative and sensitive strain signals than a strain gauge. Materials associated with the piezoelectric effect have been extensively investigated and used in sensors because of their higher sensitivity and lower cost than those only associated with the piezoresistive effect. Near-field electrospinning (NFES) was developed by Sun [7] to fabricate nanofibers in a continuous, direct, and controllable manner. Polyvinylidene fluoride (PVDF) piezoelectric fibers fabricated through NFES have been extensively studied and applied [8-9]; moreover, the biocompatibility of PVDF was verified [10]. In consideration of these advantages of PVDF fibers, piezoelectric strain sensors were fabricated through NFES in this study. A PVDF strain gauge was used to measure the arterial pulse wave.

II. Methods

This study employed a strain sensor and a PPG sensor to measure signals at two positions. An accurate measurement of blood pressure was obtained using the time difference of the two signals and the simplified skin vascular tissue model. The strain sensor was placed on the wrist above the radial artery to measure blood pressure pulse signals, which are continuous pulse signals, as displayed in Fig. 1. For a complete pulse period, the signal's peak corresponds to the systolic blood pressure, and the valley corresponds to the diastolic blood pressure. The PPG sensor was placed on the finger pulp and measured PPG signals; skin tissue and blood vessels were illuminated, the pulse altered the flow of blood in the blood vessels, and the reflected light changed in accordance with the amount of blood in the blood vessel. The greater the amount of blood in the blood vessel, the greater the diameter of the blood vessel and the less hemoglobin is absorbed; that is, for stronger absorption, the signals are stronger, corresponding to systolic blood pressure; otherwise, the signals correspond to diastolic blood pressure.

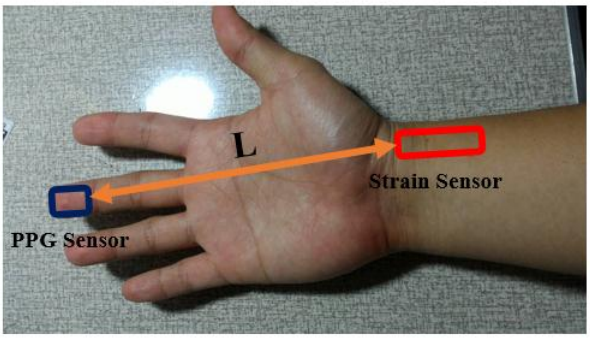


Figure 1. Schematic of sensor placement

As displayed in Fig. 1, the distance between the two sensors is represented by L. The time difference between two pulse signals was measured as the pulse propagated through the blood vessel. The peaks in both blood pressure pulse signals and PPG signals correspond to the systolic period. Therefore, the PTT is the time difference between the peak in the blood pressure pulse signal and the peak position in the PPG signal, as indicated in Fig. 2.

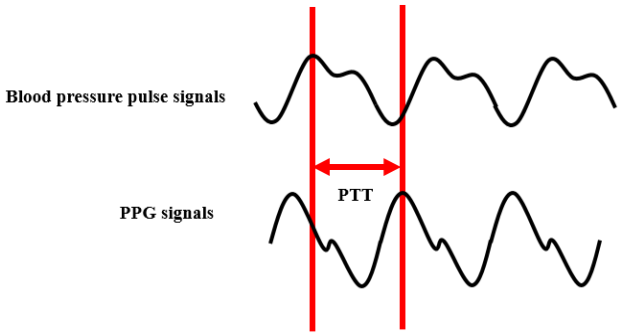
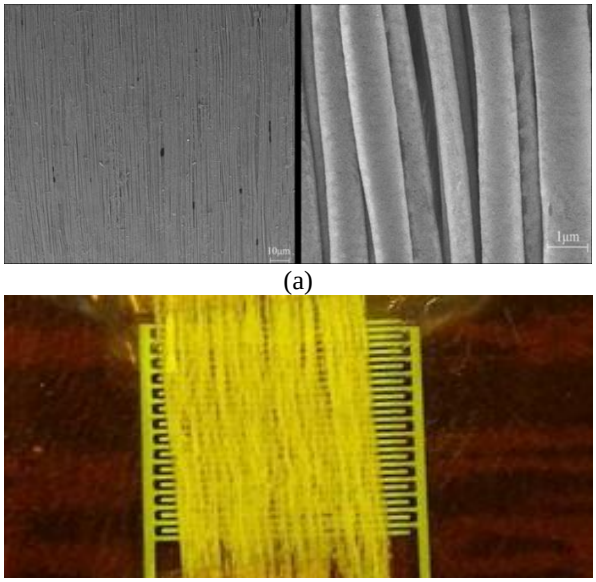


Figure 2. Waveform of two signals

A. *Piezoelectric Strain Gauge*

The strain gauge used for pulse wave sensing was fabricated from electrospun PVDF fibers. Firstly, the preparation of the solution, the concentration of the solution, the materials added, the preparation process and the placement of time will affect the electrospun fiber appearance and internal piezoelectric properties. PVDF power (Mw = 534,000; Aldrich) was dissolved in acetone to yield 18 wt% solutions mixed with dimethyl sulfoxide and surfactant under constant stirring on a heater for approximately 45 min. Because the solution is too thick, we need to scrape the solution in the bottle out by small spoon and scrape it into syringe. To extract the current produced by the PVDF fibers, a pair of comb-shaped electrodes was screen-printed onto a polyimide (PI) film in advance. The PVDF solution was electrospun into microfibrous polymer mats on the PI film with electrodes. Three electrospinning parameters, namely applied voltage, feeding volume flow rate, and syringe needle tip diameter, were strongly influential factors determining the piezoelectric properties of the PVDF mats. The applied voltage, needle tip diameter, and distance between the tip and mat in the experiment were 2 kV, 250 μm, and 1 mm, respectively. The feeding volume flow rate that demonstrated a strong effect on the piezoelectric response was 1.0 mL/h. The fiber diameters were in the range 0.6–1.5 μm, as displayed in Fig. 3(a). Planar polarization at 1 kV was conducted following electrospinning to increase the output voltage. A greater width and volume of the electrodes increased the output current and voltage, respectively. However, a larger electrode volume increased the internal resistance and decreased the tension strain. These effects reduced the output voltage. The prototype PVDF strain gauge with a length of 10 mm, width of 8 mm, and electrode space of 0.2 mm is displayed in Fig. 3(b).



(a)

(b)

Figure 3. Electrospun PVDF fibers on PI films with comb-shaped electrodes: (a) scanning electron microscopy image; (b) PVDF-based piezoelectric strain gauge.

B. Photoplethysmography Device

The PPG signal was obtained using the module placed on the skin surface area. When the light source (usually nearinfrared and red light) illuminates the skin, the light encounters multiple layers of human cells and tissues, such as skin tissues, blood vessels and bone, causing light refraction and changes in the light's speed. When the pulse causes the intravascular unit area of blood flow to demonstrate cyclical changes in blood volume, the light-sensing element generates different voltages in accordance with the change in blood volume. Because light is absorbed by hemoglobin in the blood and the most light is absorbed during contractions of the heart, the greatest amount of blood is present in the blood vessel also during heart contractions. Thus, the amplitude of the PPG signal is proportional to the volume of blood extracted from the tissue. This study used the Texas Instruments integrated development board AFE4404, as displayed in Fig. 4. Through USB connection to a computer, the actual output of the signal could be displayed using software AFE4404 EVM GUI.

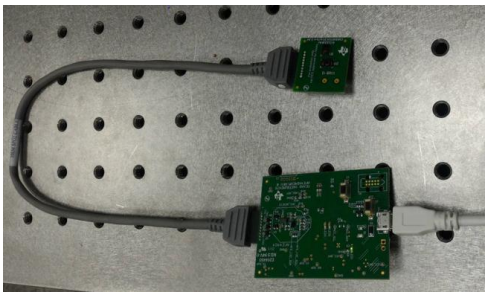


Figure 4. Signal sensor development board AFE4404

C. Blood Pressure Estimation Equation

When the left ventricle contracts, blood flows from the ventricle into an artery and causes a pressure pulse, which propagates along the blood vessel. The velocity of the pressure pulse, also referred to as the PWV, is approximately 3 to 15 m/s in the human body [11]. Because blood vessels expand as a result of pulse propagation, kinetic energy is converted into potential energy. Blood vessels easily expand because of their elastic composition, and the more easily that kinetic energy is converted into potential energy, the slower is the PWV. Consequently, the PWV is related to the compliance and elastic modulus of the blood vessel. If the blood is assumed to be nonviscous and incompressible and the thickness of the vessel wall is much smaller than the vessel radius, the PWV can be expressed using the Moens–Korteweg equation (E: elastic modulus; H: thickness of the blood vessel; d: inner radius of the blood vessel; and ρ: blood density):

$$PWV = \sqrt{\frac{EH}{d\rho}}. \tag{1}$$

When considering a fixed distance L, the relationship between the PWV and PTT can be expressed as follows:

$$PWV = \frac{L}{PTT} \tag{2}$$

The skin vascular tissue model regards blood vessels as thick-walled tubes and offers an equation for estimating blood pressure using thick-walled theory and PWV theory. Blood pressure was accurately estimated by measuring the time difference between two signals and the strain on the skin's surface. Thick-walled theory calculates the stress and strain on a tube that is caused by internal and external pressure for a thick-walled tube (typically a diameter-to-thickness ratio of less than 20) with inside radius r_i , outside radius r_o, internal pressure is p_i, and external pressure p_o (Fig. 5).

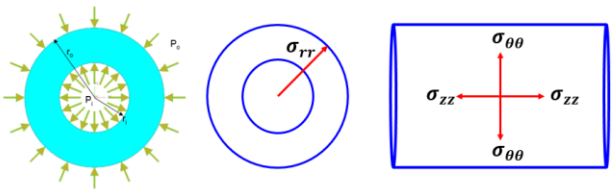


Figure 5. Skin vascular tissue model comprised of a thick-walled tube (radial stress: σ_{rr}, circumferential stress: $\sigma_{\theta\theta}$, and axial stress: σ_{zz})

By regarding the combination of skin and blood vessels as a thick-walled tube [12], the diastolic pressure can be expressed as follows:

$$p_{iD} = C_1 \cdot \left(\frac{1}{PTT}\right)^2 + C_2 \tag{3}$$

and the systolic blood pressure is expressed as follows:

$$p_{iS} = C_3 \cdot \frac{(\varepsilon_{\theta\theta S} - \varepsilon_{\theta\theta D})}{PTT^2} + p_{iD} + C_4 \tag{4}$$

C_1–C_4 are calibrated using an electronic blood pressure monitor and the two sensors mentioned. Because strain was induced along the longitudinal direction of PVDF

fibers, the output voltage could be regarded as a one-dimensional piezoelectric signal. According to *d*-type expression and Gauss's integral law, the rate of output voltage *V* is expressed as follows:

$$\dot{V} = \frac{d \cdot E \cdot l \cdot w}{c_p} \dot{\varepsilon} \tag{5}$$

where *d* is the piezoelectric coefficient of the PVDF fibers in the longitudinal direction, *E* is Young's modulus, and c_p is the capacitance of the PVDF mat. The effective length and diameter of the PVDF fibers are symbolized by l and w, respectively. The fraction in equation (5) can be regarded as a transfer coefficient between the rate of voltage change and strain. The transfer coefficient was acquired from the experiment when the strain rate was known.

Before estimating blood pressure, the coefficients C_1–C_4 had to be calibrated for every user by using equations (3) and (4); this accounted for individual bodily differences. The calibration involved substituting the blood pressure strain signals measured by the strain sensor and those measured using the electrical blood pressure monitor into equations (3) and (4). A flowchart of the calibration procedure is displayed in the purple line of Fig. 6. After calibration, users were only required to place the two sensors on suitable positions for measuring blood pressure strain and PPG signals, which were used to estimate the PTT. The result of substituting the blood pressure strain signals and PTT into equations (3) and (4) to obtain blood pressure is indicated by the red line in Fig. 6.

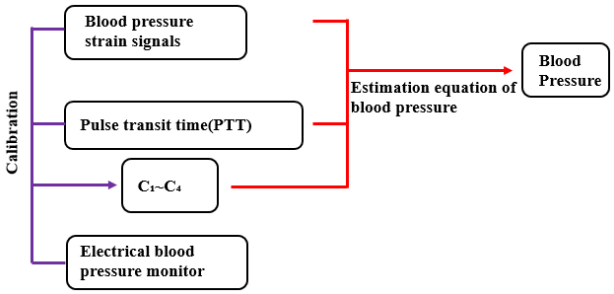


Figure 6. Flowchart of blood pressure estimation

III. Results and Conclusion

This research proposed a method to estimate blood pressure using pulse transit time and strain on skin. A piezo strain sensor and a photoplethysmography sensor were set at wrist above the radial artery and fingertip. Because two devices are set at a fixed distance, so that can cause a time delay between blood pressure pulse signals and photoplethysmography signals. The method used in the experiment to measure the PTT, involving use of a strain sensor and PPG sensor to separately obtain blood pressure strain signals and PPG signals as well as to identify the time difference between the two signals, is illustrated in Fig. 7. Voltage signals were captured using a DAQ and displayed using Labview software. An example blood pressure stain signal is displayed in Fig. 8. Table 1 presents the estimation errors and averages of the estimated blood pressure for the eight subjects. In the experimental results, the average errors were determined to be 1.7% for systolic pressure and 5.5% for diastolic pressure. That approved the invasive and continuous devices could measure blood pressures.

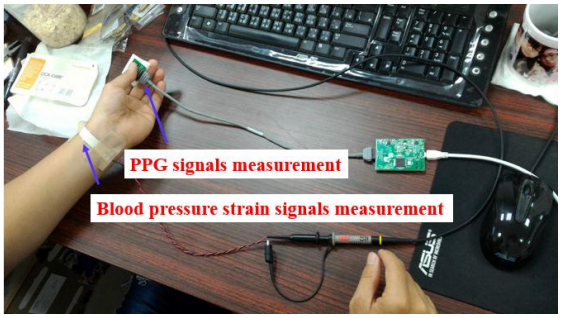


Figure 7. Experiment for PTT measurement

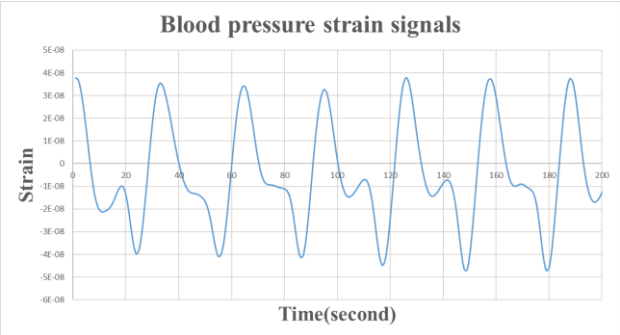


Figure 8. Blood pressure strain signals

TABLE I. Experimental Results Compared with Those Obtained Using the Clinically Validated Sphygmomanometer'

Clinically validated sphygmomanometer (mmHg)		This study (mmHg)		Error	
SBP	DBP	SBP	DBP	SBP	DBP
119	73	121	77	1.7%	5.5%

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