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Blood pressure measurement using piezoelectric effect by an ultrasonic probe

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ABSTRACT

An ultrasonic method to measure the changes in radial arterial diameter and blood pressure in a noninvasive manner was proposed to estimate viscoelastic characteristics of the arterial wall to diagnose vascular endothelial dysfunction at an extremely early stage. In the present study, a measurement method of blood pressure using the piezoelectric effect of the ultrasonic probe was investigated. At first, blood pressure waveform measured by the piezoelectric element was discussed using piezoelectric constitutive equations. We confirmed that the blood pressure waveform can be obtained by integrating the waveform measured by the piezoelectric element. Then, a conventional ultrasonic probe was modified to measure a blood pressure waveform and the measurement is demonstrated. Changes in the radial diameter was also measured using an ultrasonic diagnosis equipment with a conventional linear ultrasonic probe. The measured voltage by the piezoelectric element was of the same order as the result estimated from the theoretical consideration with typical material constants of the piezoelectric element. The diameter expanded with an increase in blood pressure and then gradually returned due to the decrease in blood pressure with viscosity. From the relationship between the arterial diameter and blood pressure, the hysteresis characteristic of the artery wall during one heartbeat was confirmed.

1. Introduction

Recently, the occurrence of cardiovascular diseases, such as heart disease and cerebrovascular disease has been increasing, accounting for 23.5% of the causes of death in Japan in 2016 [1]. Arteriosclerosis is the main factor of these cardiovascular diseases, and diagnostic techniques, such as X-ray imaging and intravascular ultrasound, are clinically used. However, iterative diagnoses by these methods are difficult because of their invasive nature. Therefore, it is important to develop a noninvasive diagnostic method for diagnoses.

As noninvasive diagnostic methods, Wetter and Kenner [2] and McDonald et al. [3] proposed an index of pulse wave velocity (PWV), and Weitz et al. [4] proposed an ankle brachial index (ABI). However, these are indicators for diagnosing irreversibly advanced lesions and are not suitable for diagnosis at an extremely early stage of arteriosclerosis. At such an extremely early stage before plaques appear, vascular endothelial functions decrease, but will recover because of reversible lesions [5,6]. Therefore, in order to diagnose and treat arteriosclerosis at an extremely early stage, development of a noninvasive diagnostic method to evaluate vascular endothelial functions is crucial.

Vascular endothelial functions are functions of endothelial cells to react to shear stress caused by blood flow and generate nitric oxide (NO), which stimulates and relaxes the blood vessel wall [7,8]. Vascular endothelial functions are conventionally evaluated by measuring the change in the inner diameter of the brachial artery by the flow-mediated dilation (FMD) method [9–11]. However, it is difficult to accurately evaluate vascular endothelial function by the FMD method [12,13], because the rate of change in the vessel diameter is only ~6% in healthy subjects.

Some methods based on the velocity dispersion of shear waves were developed as noninvasive evaluation methods of the elasticity of arteries and soft tissue [14–16]. Noninvasive evaluation methods based on force and displacement hysteresis characteristics were developed to evaluate viscoelasticity [17].

In our group, the viscoelastic parameters, *viz.*, stiffness parameter β and viscosity parameter η, were estimated by the stress-strain characteristics obtained by measuring a relationship between blood pressure and wall thickness [18–23]. We also estimated

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the temporal change of β and η by simultaneously measuring changes in arterial diameter and the blood pressure during vascular relaxation reaction [24]. Comparing the results before and after a vascularization suggested the possibility of evaluating the vascular endothelial functions and the viscoelasticity of blood vessels. However, the arterial diameter and blood pressure were measured at different positions, and the delay time between the pressure sensor and the ultrasonic probe were estimated and corrected to use as the measurements at the identical position. However, the hysteresis characteristics of the blood vessel wall could not be accurately estimated since the delay time changes during a heartbeat because of PWV changes caused by changes in blood pressure, causing an error in the delay time estimation.

In order to solve this problem, it is ideal to measure the arterial diameter and blood pressure simultaneously at an identical position by a single ultrasonic probe. For measurement, it is necessary to investigate whether a conventional ultrasonic probe can be used as a blood pressure sensor utilizing the piezoelectric effect in advance. This is because the frequency ranges of ultrasonic probes are much higher than that of the blood pressure waveform, which is mainly less than 12 Hz [25]. In this basic study, first, the frequency characteristic of the output voltage from a piezoelectric element was theoretically derived using piezoelectric constitutive equations and material parameters of an ultrasonic probe. Then, a conventional ultrasonic probe was modified to measure the blood pressure waveform and the measurement is demonstrated. Changes in the radial diameter was also measured using an ultrasonic diagnosis equipment with a conventional linear ultrasonic probe. As a result, the hysteresis characteristic of the blood vessel wall is estimated.

2. Principle and method

2.1. Measurement by piezoelectric effect

In order to measure the arterial diameter and the blood pressure at an identical position, it is necessary to transmit and receive ultrasonic waves and to detect the blood pressure using only a single ultrasonic probe. This will be realized if the blood pressure waveform is measured using the piezoelectric effect because piezoelectric elements are generally used for the ultrasonic probe. However, the frequency of a blood pressure waveform is lower than approximately 12 Hz [25] and far outside the bandwidth (usually several to 10 MHz) of ultrasonic probes. Therefore, it is necessary to investigate whether blood pressure can be detected with enough voltage by the piezoelectric elements of the ultrasonic probe.

The piezoelectric constitutive equations are as follows:

$$\mathbf{T} = \mathbf{c}^{\boldsymbol{E}} : \mathbf{S} - \mathbf{e} \cdot \mathbf{E}, \tag{1}$$

$$\mathbf{D} = \mathbf{e} : \mathbf{S} + \boldsymbol{\varepsilon}^{S} \cdot \mathbf{E}, \tag{2}$$

where $\mathbf{c}^{E}$, $\mathbf{e}$, and $\boldsymbol{\varepsilon}^{S}$ are the tensors of elastic constant under constant electric field, piezoelectric constant, and permittivity under constant stress, respectively; and $\mathbf{T}$, $\mathbf{S}$, $\mathbf{D}$, and $\mathbf{E}$ are the tensors of stress, strain, dielectric displacement, and electric field, respectively.

Lead zirconate titanate (PZT) ceramic has a perovskite structure, and the independent components of elastic constants, piezoelectric constants, and permittivity are the same as those of class *6mm* of the hexagonal system [26]. The coordinate system for the piezoelectric element with width w, length l, and thickens h, is shown in Fig. 1, where P means spontaneous polarization. Voltage V occurs when the force F is applied to the piezoelectric element.

It is assumed that there is no strain along the horizontal directions because the piezoelectric elements used in the ultrasonic probe are usually fixed putting into a casing. That is, strain components other than S_3 which corresponds to the strain along the

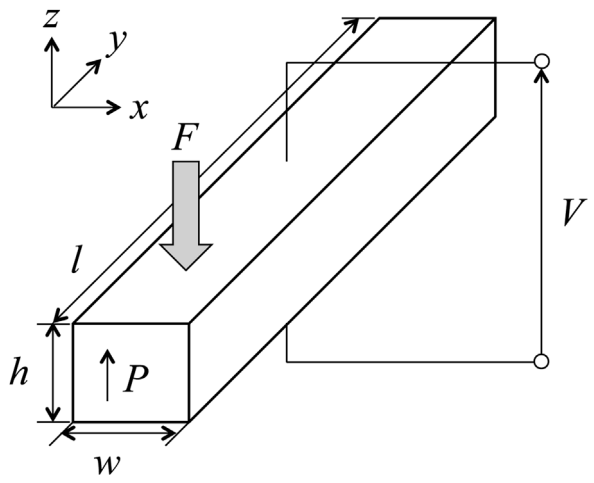

Fig. 1. Schematic view of voltage generation at a piezoelectric element by applying force.

z-axis were assumed to be 0. Thus, the following equations can be obtained from Eqs. (1) and (2):

$$T_1 = c_{13}^{E} S_3 - e_{31} E_3, \tag{3}$$

$$T_2 = c_{13}^{E} S_3 - e_{31} E_3, \tag{4}$$

$$T_3 = c_{33}^{E} S_3 - e_{33} E_3, \tag{5}$$

$$D_3 = e_{33} S_3 + \varepsilon_{33}^{S} E_3. \tag{6}$$

The following relationship among D_3, T_3, and E_3 can be derived by eliminating S_3 from Eqs. (5) and (6):

$$D_3 = \frac{e_{33}}{c_{33}^{E}} T_3 + \left(\frac{e_{33}^2}{c_{33}^{E}} + \varepsilon_{33}^{S}\right) E_3. \tag{7}$$

The electric charge Q generated at the piezoelectric element can be obtained by the surface integral of dielectric displacement D_3 by the following equation:

$$Q = wlD_3 \tag{8}$$

The relationship between the force F and the stress T_3 is as follows:

$$F = wlT_3 \tag{9}$$

The capacity of the piezoelectric element C_{d} is expressed by the following equation:

$$C_{\mathrm{d}} = \frac{\varepsilon_{33}^{S} wl}{h}. \tag{10}$$

The relationship between voltage V and electric field E_3 is as follows:

$$V = E_3 h. \tag{11}$$

The following equation can be derived by substituting D_3 in Eq. (7) into Eq. (8) using Eqs. (9) – (11):

$$Q = \frac{e_{33}}{c_{33}^{E}} F + \frac{c_{33}^{D}}{c_{33}^{E}} C_{\mathrm{d}} V, \tag{12}$$

where c_{33}^{D} is the elastic constant at constant electric displacement, and has the following relationship with c_{33}^{E}:

$$c_{33}^{D} = c_{33}^{E} + \frac{e_{33}^2}{\varepsilon_{33}^{S}}. \tag{13}$$

Next, let us consider the measurement of electric charge Q by the voltmeter with input resistance R. The current flowing from the

transducer I is the derivation of the electric charge Q. The following equation can be obtained from Eq. (12):

$$I = \frac{dQ}{dt} = \frac{e_{33}}{c_{33}^E} \cdot \frac{dF}{dt} + \frac{c_{33}^D}{c_{33}^E} C_d \frac{dV}{dt}. \tag{14}$$

The output voltage V is obtained by the following equation:

$$V = RI = R\left(\frac{e_{33}}{c_{33}^E} \cdot \frac{dF}{dt} + \frac{c_{33}^D}{c_{33}^E} C_d \frac{dV}{dt}\right). \tag{15}$$

Treat the force F and the voltage V at each frequency component f as follows:

$$F(f) = F_0(f) \cdot e^{j\omega t}, \tag{16}$$

$$V(f) = V_0(f) \cdot e^{j\omega t}, \tag{17}$$

where ω is angular frequency. From Eq. (15), $V_0(f)$ and $F_0(f)$ have the following relationship:

$$V_0(f) = \frac{j\omega R \frac{e_{33}}{c_{33}^E}}{1 - j\omega R \frac{c_{33}^D}{c_{33}^E} C_d} F_0(f). \tag{18}$$

The transfer function $G(f)$ {= $F_0(f)/V_0(f)$} from the applied force to the output voltage, and the amplitude $|G(f)|$ and phase $\angle G(f)$ are expressed as follows:

$$G(f) = \frac{V_0(f)}{F_0(f)} = \frac{j\omega R \frac{e_{33}}{c_{33}^E}\left(1 + j\omega R \frac{c_{33}^D}{c_{33}^E} C_d\right)}{1 + \left(\omega R \frac{c_{33}^D}{c_{33}^E} C_d\right)^2}, \tag{19}$$

$$\left|G(f)\right| = \frac{\omega R \frac{e_{33}}{c_{33}^E}}{\sqrt{1 + \left(\omega R \frac{c_{33}^D}{c_{33}^E} C_d\right)^2}}, \tag{20}$$

$$\angle G(f) = -\frac{1}{\omega R \frac{c_{33}^D}{c_{33}^E} C_d}. \tag{21}$$

The calculated result of the transfer function $G(f)$ is shown in Fig. 2 (a), obtained using the following typical parameters: $c_{33}^E = 1.28 \times 10^{11}$ N/m^2, $e_{33} = 25.9$ C/m^2, $\varepsilon_{33}^S/\varepsilon_0 = 2,021$, $\rho = 7,500$ kg/m^3, $R = 1$ MΩ, $w = 0.2$ mm, $l = 4.0$ mm, and $h = 3.2$ mm. These parameters are the material constants of piezoelectric elements used in the present study. The gain increased with 20 dB/decade in the frequency ranges lower than 1000 Hz.

Assuming the difference of the systolic and diastolic arterial pressures is 50 mmHg, stress T_3 is obtained as 6.67×10^3 N/m^2 from the pressure difference. Then, F is obtained as 5.33×10^{-3} N from Eq. (9). The transfer function of the output voltage of the received signal is shown in Fig. 2(b). If the main frequency component of the blood pressure waveform is several Hz, the gain is obtained as approximately -90 dBV (0.03 mV) from Fig. 2(b). As shown in Fig. 2, the zero-frequency component of the applied force cannot lead to the generation of voltage. It is also understood from the fact that voltage is obtained by varying the force on the piezoelectric materials.

In this method, it is necessary to convert the measured voltage V into force F (blood pressure). The force $F_0(f)$ is expressed using $V_0(f)$ of Eq. (18) as follows:

$$F_0(f) = \frac{1 - j\omega R \frac{c_{33}^D}{c_{33}^E} C_d}{j\omega R \frac{e_{33}}{c_{33}^E}} V_0(f) = \left(\frac{1}{j\omega R \frac{e_{33}}{c_{33}^E}} - C_d \frac{c_{33}^D}{e_{33}}\right) V_0(f). \tag{22}$$

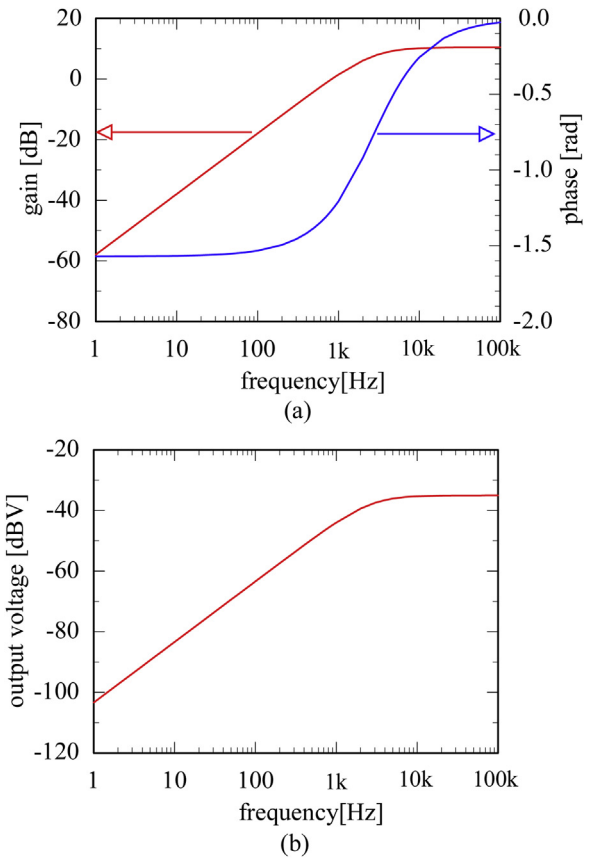

Fig. 2. (a) Amplitude and phase of transfer function of a piezoelectric element. (b) Transfer function of the output voltage.

The following equation is obtained by applying the inverse Laplace transform of Eq. (22):

$$f(t) = \frac{1}{R \frac{e_{33}}{c_{33}^E}} \int_0^t V(\tau) d\tau - C_d \frac{c_{33}^D}{e_{33}} V(t) + f_0, \tag{23}$$

where f_0 is force at $t = 0$.

Using the above parameters of the probe, the coefficients of the first and second terms of the right-hand side of Eq. (23) were calculated as $1/(Re_{33}/c_{33}^E) = 4.95 \times 10^3$ N/(Ω · m) and $C_d c_{33}^D/e_{33} = 3.00 \times 10^{-1}$ s · N/(Ω · m), respectively. Therefore, the first term is dominant in Eq. (23), and the blood pressure $f(t)$ can be obtained by integrating the measured voltage $V(t)$, as follows:

$$f(t) = \frac{1}{R \frac{e_{33}}{c_{33}^E}} \int_0^t V(\tau) d\tau + f_0. \tag{24}$$

To obtain the blood pressure waveform, it is necessary to calibrate $f(t)$ by the systolic blood pressure p_{sys} and diastolic blood pressure p_{dia}. The calibrated blood pressure waveform $\hat{p}(t)$ is expressed as follows:

$$\hat{p}(t) = \frac{p_{sys} - p_{dia}}{f_{max} - f_{min}} \left\{f(t) - f_{min}\right\} + p_{dia}, \tag{25}$$

where f_{max} and f_{min} are the maximum and minimum values of $f(t)$. This calibration is needed for each measurement because the zero-frequency component of $f(t)$ depends on p_{dia}.

2.2. Experimental equipment and procedure

In this study, two ultrasonic probes were prepared. One is a conventional linear array probe and another is a modified ultra-

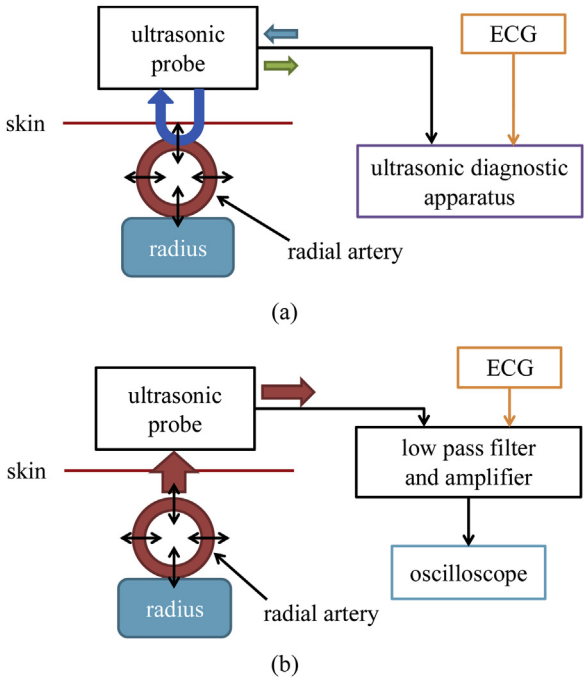

Fig. 3. (a) Schematic view of radial arterial diameter measurement by ultrasound, and (b) schematic view of radial arterial pressure measurement using the piezoelectric effect.

sonic probe to measure blood pressure waveforms. The changes in diameter and blood pressure were measured by ultrasound and the piezoelectric effect using the conventional linear array probe and modified ultrasonic probe, respectively, at different timings at an identical position in the left radial artery of a 23-year-old healthy male.

An ultrasonic diagnostic apparatus (Hitachi Aloka, ProSound F75) and a conventional linear array probe with 128 channels and a center frequency of 7.5 MHz were used for the measurement of the change in diameter. The change in diameter and electrocardiogram (ECG) were measured at a sampling frequency of 40 MHz and a frame rate of 252 Hz. The boundaries between the lumen and the anterior and posterior walls were manually determined from the received RF signal. The phased-tracking method [27–30] was applied to the determined boundary to estimate the change in diameter.

We modified a linear ultrasonic probe so that the blood pressure can be measured. The size of each rectangular piezoelectric element was $4 \times 0.2 \times 3.2\ \text{mm}^3$. To enhance the output voltage, 10 elements around the center of the ultrasonic probe were connected in parallel and these were used as a pressure sensor to measure the blood pressure waveform by the piezoelectric effect. The blood pressure waveform was measured at the same position as the ultrasonic measurement, and it was filtered with a low-pass filter having a cutoff frequency of 30 Hz and amplified with an amplifier having an amplification factor of 10. Therefore, the output voltage is expected to be approximately 3 mV. Moreover, the ECG was simultaneously measured by an electrocardiograph (Nihon Kohden Corp., ECG-1350). To calculate the blood pressure, systolic blood pressure p_{sys} and diastolic blood pressure p_{dia} were measured in the right radial artery by tonometry (Nihon Kohden Corp., JENTOW-7700).

Fig. 3 (a) and (b) shows schematic diagrams of measurements of changes in diameter and blood pressure, respectively. Virtual simultaneous measurement was realized by referring to the time of the R wave of the ECG in each measurement and correcting the timing of both waveforms of the change in diameter and the blood pressure.

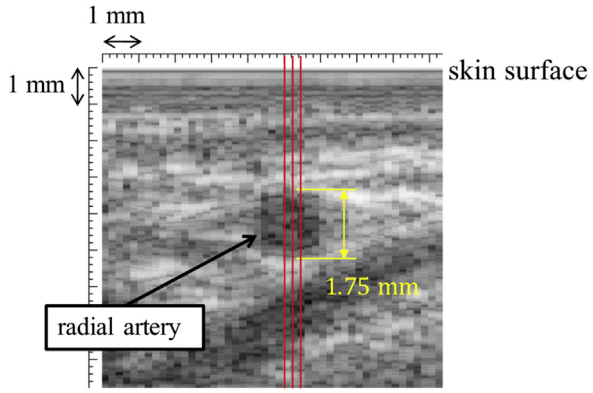

Fig. 4. B-mode image of the left radial artery.

3. Results and discussion

The B-mode image of the left radial artery is shown in Fig. 4. The diameter of the artery was determined by using three beams passing near the center of the artery as shown with red lines in Fig. 4.

The measured blood pressure waveform using this probe is shown with a dashed line in Fig. 5. The obtained amplitude of $V(t)$ in Eq. (24) was approximately 12 mV. The obtained waveform was filtered with moving average to suppress noise components. The systolic and diastolic blood pressures were measured as 114 and 61 mmHg by tonometry, respectively. The measured output was of the same order as the theoretical one. The time-integrated waveform $f(t)$ of Eq. (24) is shown with a solid red line in Fig. 5. It was confirmed that it resembles the typical blood pressure waveform measured by the tonometry in the radial artery, as shown with the blue line in Fig. 5. The effect of the frequency dependence of gain shown in Fig. 2 could be small because the frequency components of the blood pressure waveform in the frequency range of 0.9–6.0 Hz was constituted 50% of the amplitude spectrum [25]. The timing of measurement results of the change in diameter and the blood pressure waveform were synchronized with reference to the time

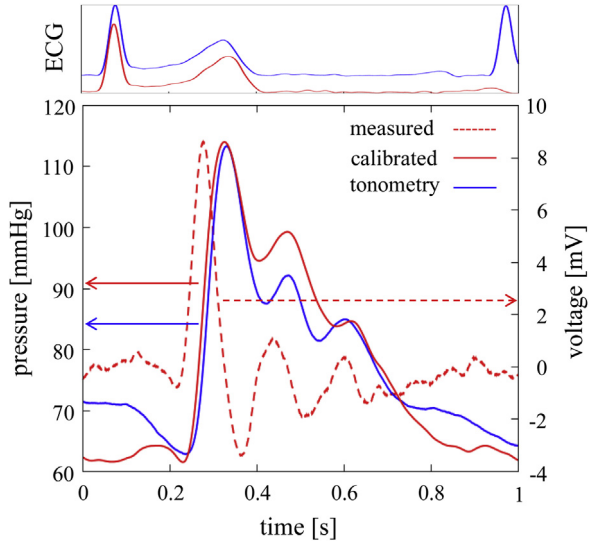

Fig. 5. Waveform measured by the piezoelectric effect (red dashed line), time-integrated waveform for the measured waveform (red solid line), and blood pressure waveform measured by the tonometry (blue line) (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article).

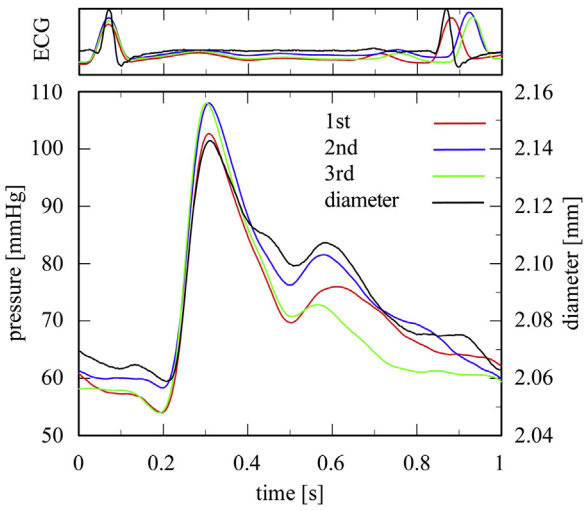

Fig. 6. Measurements for the change in diameter and blood pressure waveform by the piezoelectric effect with reference to the time of the R wave of the ECG.

of the R wave of the ECG. Little temporal error was observed in the rise of waveforms to a change in diameter and blood pressure.

Fig. 6 shows the results of blood pressure waveforms for three heartbeats and a diameter. Similar results were obtained among three heartbeats. The relationship between blood pressure and the change in arterial diameter is shown in Fig. 7. The hysteresis characteristics were confirmed as that the diameter expanded linearly with an increase in the blood pressure and then gradually returned with a decrease in blood pressure due to the viscosity. This shows that the elasticity of the arterial wall was dominant at the time of vasodilatation; otherwise the viscosity also influences the recovery. Therefore, relationships from the blood pressure to the change in diameter were successfully obtained by measuring the blood pressure waveform by the piezoelectric effect of the ultrasonic probe.

The viscoelastic parameters, *viz.*, the stiffness parameter β and viscosity parameter η, were estimated by the least squares method assuming the Voigt model [24,31,32]. The stiffness parameters β were 5.2×10^2, 4.5×10^2, and 6.6×10^2 kPa, and the viscosity parameters η were 1.4, 2.0, and 2.6 kPa·s, for the 1st, 2nd, and 3rd heartbeats, respectively. The variations of the viscoelastic parameters were relatively large because the blood pressure waveform and the change in diameter were not measured at the same time. The reproducibility will be improved by the development of a new ultrasonic probe that can measure blood pressure and the change in diameter simultaneously.

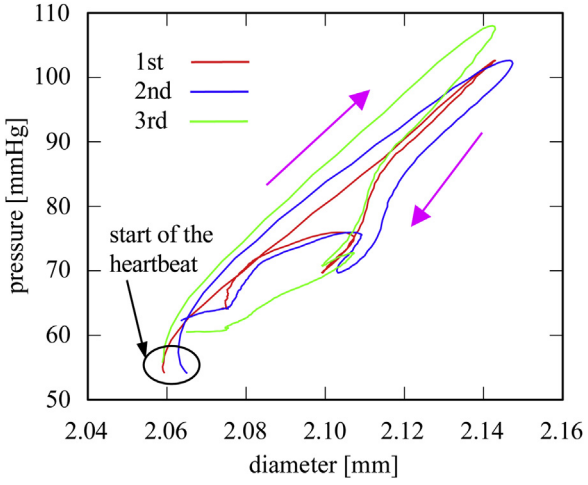

Fig. 7. Relationship between blood pressure and arterial diameter.

4. Conclusions

In this study, blood pressure waveforms measured using the piezoelectric effect by an ultrasonic probe was discussed. First, the waveform obtained by a piezoelectric element was discussed using piezoelectric constitutive equations, and we confirmed that the blood pressure waveform can be obtained by integrating the waveform measured by the piezoelectric element. An ultrasonic probe was modified to measure blood pressure waveforms. The measured voltage by the piezoelectric element was of the same order as the result estimated from the theoretical consideration and material constants of the piezoelectric element. Therefore, we confirmed that it is possible to measure the blood pressure waveform by ultrasonic probes using piezoelectric effect. Changes in the radial diameter was also measured using an ultrasonic diagnosis apparatus with a conventional linear ultrasonic probe. We observed that the timing of the rise of the blood pressure waveform and that of the change in diameter was almost same, referring to the R wave of the ECG. We successfully confirmed that the hysteresis characteristic in the radial artery during one beat was obtained from the measured blood pressure and the change in diameter. Therefore, there is a possibility of evaluating vascular endothelial functions of the arterial wall by measurements of the blood pressure waveform using an ultrasonic probe. We developed an ultrasonic probe for simultaneously measuring blood pressure and diameter. In a future study, we will perform simultaneous measurement by using this probe.

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