3P-49

Proceedings of Symposium on Ultrasonic Electronics, Vol. 31 (2010) pp. 553-554
6-8 December, 2010

Flow-Mediated Change in Viscoelasticity of Radial Arterial Wall Measured by Automated Detection of Wall Boundaries
動脈壁境界の自動検出法を用いた内皮反応時の橈骨動脈壁粘弾性特性変化の計測

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1. Introduction

Since the endothelial dysfunction is considered to be the earliest stage of atherosclerosis[1)], it is essential to develop an *in vivo* measurement method to assess the regional endothelial function and mechanical property (viscoelasticity) of the arterial wall. To evaluate the endothelial function, there is a conventional technique for measuring the transient change in the diameter of the brachial artery caused by flow mediated dilation (FMD) after the release of avascularization[2, 3)]. However, this method does not directly evaluate the viscoelasticity of the intima-media region of the arterial wall. In the previous paper, therefore, we proposed a method for simultaneous measurement of waveforms of the radial strain and blood pressure of the intima-midia region of the radial artery[4)]. From *in vivo* experiments, the viscoelasticity parameters of the arterial wall were estimated from the measured stress-strain relationship and their transient changes after the release of avascularization were revealed[5)].

In the proposed method, it is necessary to measure the strain waveform using *phased tracking method*[6,7)]. For doing this, the lumen-intima boundary (LIB) and media-adventitia boundary (MAB) need to be detected in ultrasonic data. Therefore, in this study, we used the automatic and objective boundary detection, as explained next, and measured the transient change in the viscoelasticity of the radial arterial wall due to FMD.

2. Principles

2.1 Determination of optimum initial positions of arterial wall boundaries

In our previous study, LIB and MAB were manually determined by referring to the RF echo from the posterior wall[4,5)]. However, it is difficult to determine accurate positions by this method (e.g., the results would be different among different observers). Therefore, the technique is necessary to detect LIB and MAB objectively and adaptively.

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Simple methods, such as thresholding, have been used for boundary detection[8)]. Usually, these methods label the boundary by using the gradient information of the image intensity in a local area. A major disadvantage of these methods is its amplitude sensitivity, which complicates threshold setting. To overcome the disadvantage, in the present study, the LIB and MAB were detected using the complex template matching between the measured complex demodulated signal and the adaptive complex model signal.

Figure 1 shows a cross-sectional B-mode image of an radial artery and measured RF signals of the beams of interest.

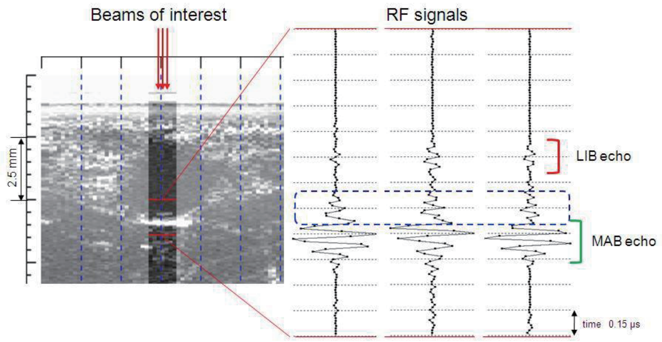

Fig. 1. Cross-sectional B-mode image of a radial artery and measured RF signals of the beams in interest.

These waveforms in the area of interest consist of many echoes including those from LIB and MAB. In this study, a waveform is added to the model between echoes from LIB and MAB, as the scattered wave from intima-media region to improve the fitness between model and measured signals. Therefore, let us model the echoes from the intima-media region including echoes from LIB and MAB as follows:

$$\hat{z}(nT_s) = \hat{z}_1(nT_s) + \hat{z}_2(nT_s) + \hat{z}_3(nT_s) \quad (1)$$

$$\hat{z}_i(nT_s) = C_i \exp[j2\pi f_0(nT_s - \tau_i)] \cdot w_i(nT_s - \tau_i), \quad (i = 1, 2, 3) \quad (2)$$

$$w_i(nT_s) = \begin{cases} 0.5 - 0.5\cos\left(\dfrac{2\pi f_0 nT_s}{N}\right), & \tau_i \le nT_s < \tau_i + NT_s \\ 0 \quad \tau_0 \le nT_s < \tau_i, & \tau_i + NT_s \le nT_s \end{cases} \quad (3)$$

where f_0 is the center frequencies of the echoes estimated from the region of interest, $\{C_i\}$ are the complex amplitude of echoes, $\{\tau_i\}$ are the time delays of respective echoes, N is the number of points constructing a pulse, and T_s is the sampling interval. The model echo $\hat{z}_i(nT_s)$ is formulated by the product of the sinusoidal wave and envelope $w_i(nT_s)$. Finally, the model echo $\hat{z}(nT_s)$ obtained by the sum of the echoes from LIB (i = 1), MAB (i = 3), and scattered waveform from intima-media region (i = 2). The optimal parameters $\{C_i\}$, and $\{\tau_i\}$ (i = 1, 2, 3) are determined so that the squared difference between the complex model signal $\hat{z}(nT_s)$ and the measured complex demodulated echo signal $z(nT_s)$ becomes the minimal.

2.2 Procedure for *in vivo* measurement

In this study, the right radial artery of a healthy subject was measured. In the measurement of the radial artery, ultrasonic RF echoes (transmit: 22 MHz) were acquired at a sampling frequency of 66.5 MHz for 2 s. This acquisition was repeated at every 20 s for 2 minutes at rest before avascularization and every 12 s for 3 minutes after recirculation. At the same time, the waveform of blood pressure on the left radial artery was continuously measured with a sphygmometer. The transient change in the stress-strain relationship during a cardiac cycle due to FMD was obtained from the measured strain $\gamma(t)$ and blood pressure $\tau(t)$. From the stress-strain relationship, the viscoelasticity of the intima-media region was estimated using the least-square method by assuming the Voigt model as a viscoelastic model of the intima-media region[5)].

3. Results

Figures 2(a), **2(b),** and **2(c)** show the detected boundaries (LIB, MAB) and the result of the complex template matching. In Fig. 2(a), the boundary positions, which are shown as LIB and MAB on the B-mode image, are detected appropriately and objectively by complex template matching. This result does not depend on an observer.

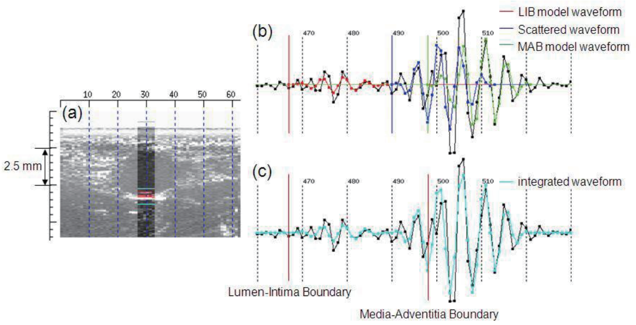

Fig. 2(a). Detected boundaries LIB and MAB.
(b). Result of the complex template matching. (measured and each model waveforms)
(c). Result of the complex template matching. (measured and sum of the waveforms)

Figure 3 shows the transient change in the means of static elasticity E_s and viscosity η averaged by 3 ~ 5 ultrasonic beams. The viscoelasticity of intima-media region of radial arterial wall changed gradually. The minimum static elasticity E_s was measured at 35 s after the release of the cuff. The maximum % change in static elasticity E_s was about 56% (400 kPa). The maximum viscosity η at 79 s after recirculation was about 670% (6.7 kPa·s) larger than the mean at rest. These temporal changes came around about 3 minutes after release of the cuff. These results show that transient change in the viscoelasticity due to FMD was measured.

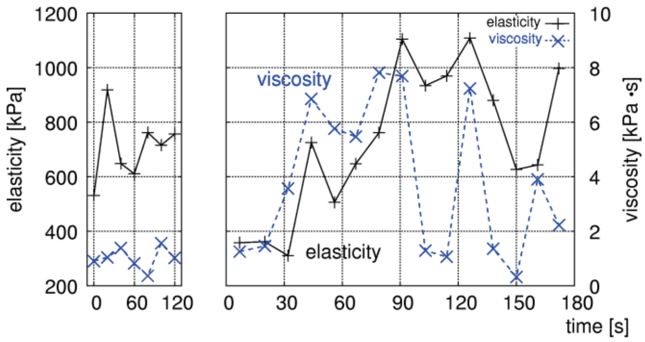

Fig. 3. Transient change in viscoelasticity of the intima-media region of radial arterial wall due to FMD.

4. Conclusion

In this study, we investigated a method to detect LIB and MAB whose initial positions are required to track their displacements and to obtain changes in thickness for measurement of the stress-strain relationship. This method detects the LIB and MAB appropriately and objectively, and has potential for reducing the operator-dependent variations, and it would be useful for the automatic and noninvasive measurement of viscoelasticity of the arterial wall.

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