3P5-27

Proceedings of Symposium on Ultrasonic Electronics, Vol. 30 (2009) pp. 575-576
18-20 November, 2009

Transient Change in Viscoelasticity of Radial Artery due to Flow-Mediated Dilation Measured by Accurate Detection of Arterial Wall Boundaries

血管壁境界の高精度検出による内皮依存性弛緩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

The endothelial dysfunction is considered to be an initial step of atherosclerosis[1)]. Additionally, it is reported that the smooth muscle, which constructs the media of the artery, changes its characteristics owing to atherosclerosis[2)]. Consequently, it is important for early preventive treatment to noninvasively assess the endothelial function and the mechanical property of the media which is mainly composed of smooth muscle.

For the evaluation of the endothelial function, there is a conventional technique for measurement of the transient change in the inner diameter of the brachial artery caused by flow-mediated dilation (FMD) after the release of avascularization[3)]. For more sensitive and regional evaluation, we developed a method for direct measurement of the change in the elasticity of the intima-media region due to FMD[4,5)]. Additionaly, we proposed a new noninvasive method for measurement the transient change in the stress-strain relationship owing to the FMD at the radial artery for measurement of mechanical property of the intima-media region[6)].

From the stress-strain relationship, the viscoelasticity of the intima-media region was estimated using the least-square method. In the present study, we investigated the method to accurately detect these boundaries (lumen-intima boundary (LIB) and media-adventitia boundary (MAB)), which determine the initial region required to track their displacements and changes in thickness for measurement of the stress-strain relationship.

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[6,7)]. However, it is difficult to determine accurate positions by this method, and the results would be different among different observers.

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In the present study, therefore, the LIB and MAB were detected using the complex template matching between the measured complex demodulated signal and the adaptive complex model signal[5)] to accurately measure viscoelasticity of the arterial wall. **Figure 1** shows a B-mode image of an radial artery along the longitudinal axis and measured RF echoes of a beam of interest.

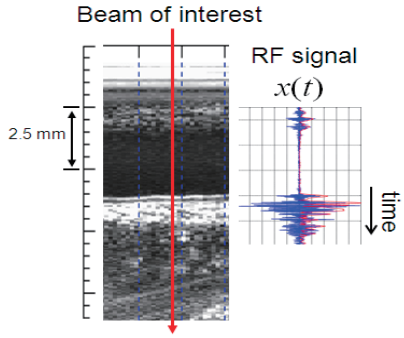

Fig. 1. B-mode image of a radial artery along the longitudinal axis and measured RF echoes of a beam of interest.

Let us model the echoes from the LIB and MAB as follows.

$$x_{\mathrm{LIB}}(t) = a_1 \sin[2\pi f_1 (t-\tau_1)] \cdot w_1(t-\tau_1), \quad (1)$$

$$x_{\mathrm{MAB}}(t) = a_2 \sin[2\pi f_2 (t-\tau_2)] \cdot w_2(t-\tau_2), \quad (2)$$

$$w_i(t) = 0.5 - 0.5\cos\left(\frac{2\pi f_i t}{N}\right) \quad (i = 1, 2), \quad (3)$$

$$(t = 0, 1, 2, \cdots nT_s)$$

where f_1 and f_2 are the center frequencies of the LIB and MAB echoes, a_1 and a_2 are amplitude coefficients of echoes from the LIB and MAB, τ_1 and τ_2 are time delays of respective echoes, N is the number of cycles at the center frequency in an echo, and T_s is the sampling interval. These parameters a_i, f_i, and τ_i (i = 1, 2) are determined so that the squared difference between the model signal $x_{\mathrm{LIB}}(t)+x_{\mathrm{MAB}}(t)$ and measured echo $x(t)$ becomes the minimal. The model echoes $x_{\mathrm{LIB}}(t)$ and $x_{\mathrm{MAB}}(t)$ are formulated by the sinusoidal waves and envelopes $w_1(t)$ and $w_2(t)$, and the normalized mean square errors during the pulse durations are calculated.

2.2 Viscoelasticity estimation of arterial wall using the least-square method

The smooth muscle constructs the media and it is the main source of the viscoelasticity of the vessel wall[8]. By assuming the Voigt model as a viscoelastic model of the intima-media region, the stress-strain relationship is given by

$$\hat{\tau}(t) = E_s\gamma(t) + \eta\dot{\gamma}(t) + \tau_0, \qquad (4)$$

where $\hat{\tau}(t)$ is the model stress and $\gamma(t)$, $\dot{\gamma}(t)$, E_s, and η are strain, strain rate, static elasticity, and viscosity coefficient, respectively. The minute strain $\gamma(t)$ of the right radial arterial wall during a cardiac cycle was measured using the *phased tracking method* [9]. The strain $\gamma(t)$ is the incremental strain due to the pulse pressure, whereas the measured stress $\tau(t)$ includes the bias stress (diastolic blood pressure). Therefore, τ_0 is added to the right-hand side of eq. (4) as the bias stress corresponding to diastolic pressure.

The parameters in eq. (4), E_s, η, and τ_0, are estimated using the least-square method by minimizing the mean squared error, α, between the measured stress $\tau(t)$ and model stress $\hat{\tau}(t)$.

2.3 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 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 and blood pressure.

3. Results

Figures 2(a) and **2(b)** shows 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.

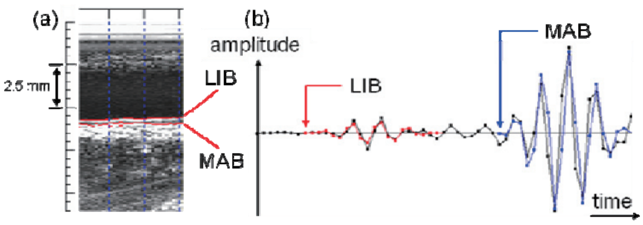

Fig. 2(a). Detected boundaries LIB and MAB.
(b). Result of the complex template matching.

Figure 3 shows the transient change in the means and standard deviations (SDs) of static elasticity E_s and viscosity η averaged by 5 ultrasonic beams.

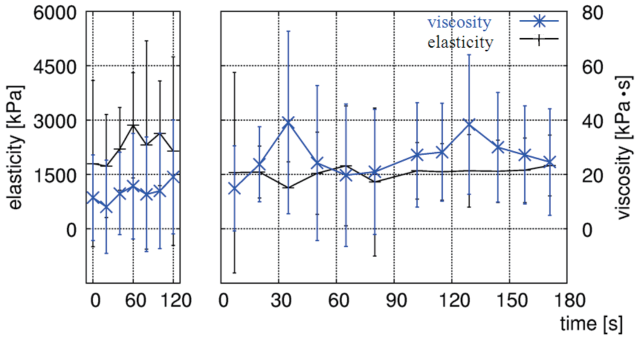

Fig. 3. Transient change in viscoelasticity due to FMD.

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 50% (1100 kPa). The maximum viscosity η at 35 s after recirculation was about 191% larger than the mean at rest. These result show that transient change in the viscoelasticity due to FMD was measured.

4. Conclusion

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

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