Strain Imaging of Arterial Wall with Translational Motion Compensation and Error Correction

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Abstract—To assess elastic properties, correlation techniques are widely used to measure the displacement and strain of the arterial wall caused by the heartbeat. However, the displacements estimated by the phase-sensitive correlation methods are biased when the frequency used for the displacement estimation is different from the center frequency of RF echo. One of reasons for the frequency variation is the interference of echoes. In the case of the arterial wall, the displacement due to radial translation is larger than that contributing to strain by a factor of 10 and, thus, the error resulting from the translational motion due to the mismatch between the frequency used for displacement estimation and the actual center frequency is not negligible compared with the minute displacement contributing to strain. In this study, a method is proposed in which the radial translation is removed prior to the calculation of complex correlation between echoes in two different frames to estimate the phase change between the echoes. The radial translation is removed by tracking the echo from the luminal interface of the wall because it is dominant compared with echoes from scatterers in the wall and is less affected by the interference. Using this procedure, the significant error resulting from the large translation can be greatly suppressed. After the removal of translation, an error correcting function based on complex correlation is introduced to further reduce the error due to the frequency mismatch. Accuracy improvement by the proposed method was validated using phantoms. As shown in the figure, the error in strain estimated by the proposed method was 12.0% from the theoretical strain profile, significantly smaller than that (23.7%) by the conventional method. Furthermore, in the in vitro experiments using extracted femoral arteries, the arterial wall containing calcified tissue showed very small strain in comparison with that almost homogeneously composed of fibrous tissue (mixture of smooth muscle and collagen). The means and the standard deviations of distensibility (calculated from strain and internal pressure) of fibrous and calcified tissues obtained for 7 sections of 5 femoral arteries were 2.42±2.1 and 0.35±0.50 MPa⁻¹, respectively.

I. INTRODUCTION

We have developed a phase-sensitive correlation method, namely, phased-tracking method, for estimating the displacement distribution in the arterial wall for strain imaging [1], [2]. In such methods, ultrasonic pulses with finite frequency bandwidths are used. Therefore, the apparent change in center frequency occurs due to the interference of echoes from scatterers in the wall. Under such a condition, the displacement estimates are biased because they are obtained using the phase changes and center frequencies of received RF echoes. An autocorrelation-based method was proposed to compensate for this apparent change in center frequency [3]. However, the performance of this method is limited because the autocorrelation technique prefers the narrow band signals [4].

To reduce the influences of the change in center frequency, the large translational motion of the arterial wall should be compensated because the error due to the change in center frequency depends on the magnitude of the displacement ( = phase change) and the translational motion of the arterial wall is much larger than strain. To improve the accuracy of the strain estimation, in this study, a method is proposed to compensate for the translational motion before the estimation of the displacement component contributing to strain.

II. METHODS

A. Displacement estimation using conventional autocorrelation methods

The phases of echoes from a target, which is located at depth \(d\) in the initial frame and moving along the ultrasonic beam, are different in two consecutive \(n\)-th and \((n+1)\)-th frames. This phase shift, \(\Delta \theta_d(n)\), depends on the displacement of the target between two frames. Therefore, the instantaneous displacement \(\Delta x_d(n)\) between these two consecutive frames is estimated as follows:

\[
\Delta x_d(n) = \frac{c_0 \Delta \theta_d(n)}{4\pi f_{0c}},
\]

where \(c_0\) and \(f_{0c}\) are the speed of sound and the frequency used for displacement estimation, respectively. Phase shift \(\Delta \theta_d(n)\) is estimated by the complex cross-correlation function \(\gamma_{d,n}\) using the complex quadrature demodulated signals \(z(d;n)\) of received RF echoes as follows [1]:

\[
\Delta \theta_d(n) = \angle \gamma_{d,n},
\]

\[
\gamma_{d,n} = \sum_{d\in R} z^*(d+x_d(n);n) \cdot z(d+x_d(n);n+1),
\]

where \(\angle\) and \(*\) show the phase angle and the complex conjugate, respectively, and \(R\) was set to the width at -20 dB of the maximum of the ultrasonic pulse. The accumulated displacement \(x_d(n)\) is obtained by accumulating the instantaneous displacement \(\Delta x_d(n)\) with respect to frame, and the displacement distribution is obtained by estimating the accumulated displacement \(x_d(n)\) at each depth \(d\) along the arterial radial direction. Finally, radial strain \(\varepsilon_{r,d}\) is obtained by the spatial differentiation of displacements with respect to the arterial radial direction.
B. Translational motion compensation

As shown by eq. (1), in conventional phase-sensitive autocorrelation methods, the knowledge of center frequency is required to obtain the unbiased displacement estimates. The estimated instantaneous displacement $\Delta x_t$ can be expressed by the sum of two components $\Delta x_t$ and $\Delta x_s$ due to the translational motion and contributing to strain as follows:

$$\Delta x = \Delta x_t + \Delta x_s = \frac{c_0}{4\pi f_0e} \Delta \theta_t + \frac{c_0}{4\pi f_0e} \Delta \theta_s,$$

(4)

where $\Delta \theta_t$ and $\Delta \theta_s$ are phase shifts of echoes caused by the translational motion and strain, respectively. The estimated displacement is biased when frequency $f_0e$ used for estimation is different from the actual center frequency $f_0$ of RF echo. As an typical case, the translational motion of the arterial wall is larger than strain (change in thickness) by a factor of ten. Therefore, the error resulting from the translational motion is far greater than that from strain. This significant error can be reduced by removing the first term in the right-hand side of eq. (4), which corresponds to the compensation of the translational motion. In this study, the translational motion is removed before estimating the phase shift of echoes due to strain as described below.

![Fig. 1. Displacement estimation with translational motion compensation.](image)

The translational motion of the wall is compensated by tracking the echo from the luminal interface using the conventional method because the echo from the luminal interface is dominant and is less influenced by the interference from other weak echoes from scatterers in the wall. Translational motion compensation is done as follows: The region of interest (ROI) is manually assigned in the initial frame $n = 0$ and, then, its position in each frame $n$ is automatically determined by estimating the displacement of the luminal interface $\Delta x_l(n)$ (the shallowest edge of the ROI) using the conventional autocorrelation method [1]. In this study, key frames are defined so that the displacement of the luminal interface $\Delta x_l(n)$ between two successive key frames is the closest to a spacing, $\Delta X$, of sampled points. Figures 1(a) and 1(b) show RF echoes from the cylindrical phantom used in this study in two consecutive key frames, and the translational motion can be compensated by shifting the RF signal in one of two consecutive key frames by one point as shown in Figs. 1(c) and 1(d). After compensation, the phase shift of echoes between these two key frames is estimated at each depth $d$ along the radial direction of the phantom. In this case, the estimated phase shift contains only the phase shift due to strain. Therefore, the error resulting from the translational motion can be reduced.

C. Error correcting function

After translational motion compensation, the displacement contributing to strain is estimated using the phase shift of RF echoes. In such a situation, a large error resulting from the translational motion is removed, but error due to the mismatch between the frequency $f_0e$ used for estimation and the actual center frequency $f_0$ still remains in the estimation of the displacement $\Delta x_s$ contributing to strain even after the compensation of translational motion $\Delta x_t$. This error is much smaller than that in the case when the simultaneous estimation of displacements $\Delta x_t$ and $\Delta x_s$ due to the translational motion and strain. However, in this study, a function is introduced to further reduce this small error in addition to the error due to the translational motion. The error correcting function $\beta_{d,k}$ in $k$-th key frame ($n_k$-th frame) is defined as follows:

$$\beta_{d,k} = \frac{c_0}{4\pi f_0e} \left| \frac{\gamma_{d,k} \cdot \gamma_{d,k}^*}{\Delta X} \right|,$$

(5)

$$\gamma_{d,k} = \sum_{d \in R} z^*(d + x_l(n_k) - x_l(n_{k+1})), \quad \gamma_{d,k}^* = \sum_{d \in R} z^*(d + x_l(n_k)) \cdot z(d + x_l(n_{k+1})), \quad \gamma_{d,k} = \sum_{d \in R} z^*(d + x_l(n_k)) \cdot z(d + x_l(n_k) - x_l(n_{k+1})), \quad \gamma_{d,k}^* = \sum_{d \in R} z^*(d + x_l(n_k)) \cdot z^*(d + x_l(n_{k+1})).$$

(6)

(7)

Key frames are assigned so that $|x_l(n_{k+1}) - x_l(n_k)| = \Delta X$ and, thus, the signal $z(d + x_l(n_k) - x_l(n_{k+1}))$ in $\gamma_{d,k}$ is artificially displaced by just a spacing, $\Delta X$, of sampled points in comparison with $z(d + x_l(n_k + 1) - x_l(n_{k+1}))$ in $\gamma_{d,k}^*$. As shown by the numerator of eq. (5), this artificial displacement $\Delta x_o$ estimated by the difference between angles of $\gamma_{d,k}$ and $\gamma_{d,k}^*$, which can be obtained by $\left| \gamma_{d,k} / \gamma_{d,k}^* \right|$. Although it should be $4\pi f_0 \Delta X/c_0$, the frequency used for displacement estimation $f_0e$ is different from the actual center frequency $f_0$. Equation (5) shows the ratio of the artificial displacement $\Delta x_o$ estimated by phase shift to actual one $\Delta X$. The error-corrected displacement $\Delta x_{s,d}(n_k)$ between $k$-th and $(k+1)$-th key frames contributing to strain can be obtained as follows:

$$\Delta x_{s,d}(n_k) = \frac{1}{\beta_{d,k} \cdot c_0} \frac{c_0}{4\pi f_0e} \gamma_{d,k}.$$

(8)

D. Elasticity estimation

When only the normal stresses, $\sigma_r$, $\sigma_\theta$, and $\sigma_z$, in the radial, circumferential, and longitudinal directions are applied to a cylindrical shell, radial and longitudinal strains, $\varepsilon_r$ and $\varepsilon_z$, of the wall are expressed as follows [5]:

$$\varepsilon_r = \frac{\sigma_r}{E_r} - \frac{\nu \sigma_\theta}{E_\theta} - \frac{\nu \sigma_z}{E_z},$$

(9)

$$\varepsilon_z = \frac{\sigma_z}{E_z} - \frac{\nu \sigma_r}{E_r} - \frac{\nu \sigma_\theta}{E_\theta}.$$

(10)
where $E_r$, $E_\theta$, and $E_z$ are elastic moduli in the radial, circumferential and longitudinal directions, respectively, and $\nu$ is Poisson's ratio. In this study, the arterial wall is assumed to be incompressible ($\nu = 0.5$) and isotropic ($E_r = E_\theta = E_z = E$). Under in vivo condition, the artery is strongly restricted in its longitudinal direction. Thus, the longitudinal strain is assumed to be zero ($\varepsilon_z = 0$). By applying these assumptions, eq. (10) can be modified as follows:
\[
\sigma_z = \frac{1}{2}(\sigma_r + \sigma_\theta).
\] (11)

By substituting eq. (11) to eq. (9), the distensibility, $S = 1/E [\text{Pa}^{-1}]$, is defined as follows:
\[
S = \frac{1}{E} = \frac{4}{3} \frac{\varepsilon_r}{\sigma_r - \sigma_\theta}.
\] (12)

Stresses $\sigma_r$ and $\sigma_\theta$ of a cylindrical shell are expressed as follows [6]:
\[
\sigma_r = \frac{-1}{2}p_r,
\] (13)
\[
\sigma_\theta = \frac{-r_0}{h_0}p_r,
\] (14)

where $r_0$ and $h_0$ are the inner radius and entire wall thickness in the end of cardiac diastole, respectively, and $p$ is the internal pressure. By substituting eqs. (13) and (14) into eq. (12), distensibility $S$ is expressed as follows:
\[
S = \frac{-\varepsilon_r}{\frac{3}{8}\frac{2r_0}{h_0} + 1}p.
\] (15)

In cardiac systole, the artery diameter expands due to the pressure increment, and the wall thickness becomes small. Therefore, $\varepsilon_r < 0$. As shown in eq. (15), distensibility $S$ corresponds to the tissue strain normalized by the average stress. In this study, by defining the average stress by $3(2r_0/h_0 + 1)/8$, distensibility $S_d$ at depth $d$ is obtained by substituting $\varepsilon_{r,d}$ into $\varepsilon_r$ in eq. (15).

E. Experimental setup

In this study, a cylindrical phantom and excised arteries were measured by ultrasound. The outer and inner diameters of the phantom, made of silicone rubber (elastic modulus $E = 749$ kPa corresponding to distensibility $S = 1/E = 1.33$ MPa$^{-1}$), are 10 and 8 mm, respectively. The phantom contains 5% carbon powder (by weight) to obtain sufficient scattering from the inside of the wall.

Figure 2 shows a schematic diagram of the measurement system. A change in pressure inside the phantom or artery was induced by circulating a fluid using a flow pump. The change in internal pressure was measured by a pressure sensor (NEC, Tokyo, 9E02-P16).

In ultrasonic measurement, the phantom and arteries were measured with a linear-type ultrasonic probe (Aloka, Tokyo, SSD-6500). The nominal center frequency was 10 MHz. RF echoes were acquired at 40 MHz at a frame rate of 286 Hz.

In the in vitro experiment, femoral arteries excised from patients with arteriosclerosis obliterans were measured. The arteries were placed in a water tank filled with 0.9% saline solution at room temperature. A needle was attached to the external surface of the posterior wall using strings to identify the sections to be imaged during the ultrasonic measurement. After the ultrasonic measurement, pathological images of the measured sections were obtained by referring to the needle.

III. Basic Experimental Results

Figure 3(a) shows the strain distribution along the ultrasonic beam (radial direction of the phantom) estimated by the conventional autocorrelation method [1], [2]. Plots and vertical bars show means and standard deviations of 46 ultrasonic beams. The dashed line shows the theoretical strain profile, which is obtained by the measured internal pressure and elastic modulus ($E = 749$ kPa) of the wall given by [7]
\[
\varepsilon_r = \frac{-3}{2} \frac{r_i^2 - r_o^2}{(r_o^2 - r_i^2)r_i^2} p,
\] (16)

where $r_i$ and $r_o$ are original inner and outer radii, respectively.

![Fig. 3. Estimated strain distribution. (a) Conventional method [1], [2]. (b) Conventional method with center frequency estimation [3]. (c) Proposed method with translational motion compensation and error correction.](image)

The elastic modulus of the phantom was measured by the different pressure-diameter testing. Although mean values follow the theoretical strain profile, the standard deviations of strains estimated by the conventional method are large. In Fig. 3(b), the center frequency estimation proposed in [3] was applied to the conventional method. However, the strain estimates are not improved so much. Figure 3(c) shows the strain distribution obtained by the proposed method. Strain estimates were improved significantly by translational motion compensation and error correction. From the measured internal pressure and the estimated strain shown in Fig. 3(c), the distensibility $S$ was determined to be $1.49\pm0.37$ MPa$^{-1}$ (1.33 MPa$^{-1}$ measured by pressure-diameter testing).

IV. In Vitro Experimental Results

Figures 4(a-1) and 4(b-1) show the B-mode images of two different regions in the excised femoral artery. The region
shown in Fig. 4(a-2) shows very low strain. On the other hand, the region shown in Fig. 4(b-2) shows relatively larger strain and there is a clear strain decay with respect to the distance from the lumen [7]. Figures 4(a-3) and 4(b-3) show the images of distensibility obtained by eq. (15) using the measured internal pressure $p$ and estimated strain distributions $\{\varepsilon_{r,d}\}$. By comparing the images of distensibility and pathological images of the corresponding sections shown in Figs. 4(a-4), 4(b-4), 4(a-5), and 4(b-5), the very hard region surrounded by the green line in Fig. 4(a-3) corresponds to calcified tissue. On the other hand, the region composed of smooth muscle and collagen shows a relatively homogeneous and higher distensibility distribution. By applying the same procedure to 7 regions in 5 femoral arteries, the means and standard deviations of distensibility $S$ for fibrous and calcified tissues were determined to be $2.42\pm2.1$ MPa$^{-1}$ and $0.35\pm0.50$ MPa$^{-1}$, respectively. These results show that the proposed method successfully reveals the difference in the deformation properties resulting from different tissue compositions.

V. Discussion

The estimation of stress distribution in the arterial wall is one of the most difficult problems. In this study, the stress distribution in the arterial wall is assumed to be constant as shown in eq. (15). Even in the case of a homogeneous cylindrical shell, the stress (strain) distribution is dependent on the distance from luminal surface. When an artery is a homogeneous cylindrical shell, the stress (strain) distribution can be theoretically derived as shown in Fig. 3. However, it is difficult to estimate the stress (strain) distribution when an artery is inhomogeneous in the wall thickness and elastic properties. Therefore, in this study, the average stress was used to normalize the measured strain, as shown in eq. (15).

This method is effective to compensate for the dependence of strain on the pulse pressure. When only the strain itself is considered to evaluate the elasticity, the strain is dependent on the pulse pressure which does not relate to the elasticity. Use of the stress distribution of a cylindrical shell given by eq. (16) is another solution to compensate for the pulse pressure variation.

VI. Conclusion

In displacement estimation based on the phase change of echoes, the displacement estimates are biased when the center frequency of the RF echo changes. Such an apparent change in the center frequency is caused by the interference of echoes from scatterers. In the case of the arterial wall, the radial translation of the arterial wall is much larger than the displacement contributing to strain. Therefore, the error resulting from the translational motion is often larger than the displacement contributing to strain. To reduce this error, in this study, the translational motion was compensated prior to the estimation of the phase shift between echoes. As a result, the proposed method provides better strain estimates in comparison with conventional autocorrelation-based methods.

REFERENCES