1Pa5-2

Investigation on effect of transmit condition on displacement estimation by phase-sensitive 2D motion estimators

2 次元位相追跡法を用いた変位計測における送信条件の影響の検討

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1. Introduction

A high frame rate is preferable for estimation of arterial wall motion due to cardiac pulsation. Therefore, the condition of ultrasonic transmission is very important.

In the present study, in silico phantoms mimicking a carotid arterial wall simulated such a motion due to cardiac pulsation. The accuracies in motion estimation were evaluated under the transmit conditions with plane wave imaging and synthetic aperture imaging using diverging beams.

2. Materials and Methods

2.1 Plane wave imaging

In the plane wave imaging, all available elements in a probe are excited to emit a non-steered plane wave. Then, the same elements received ultrasonic echoes. **Figure 1** shows illustration of a plane wave. Elements filled in red correspond to the active aperture. In the present study, the steering angle of the transmission wave was zero. Thus, forward propagation distance r_t is expressed as

$$r_t = z_t. \tag{1}$$

2.2 Synthetic aperture imaging

In the synthetic aperture imaging, spherical waves were emitted from a transmit aperture composed of 72 elements, and ultrasonic echoes were received with all elements in the probe. In this case, the active transmit aperture was shifted by 6 elements after every transmission, and the transmit-receive event was repeated 21 times. **Figure 2** shows the illustration of a spherically diverging beam. In the present study, a spherical wave from a virtual source (position(x_f, z_f)) was used. Thus, forward propagation distance r_t is expressed as

$$r_t = \sqrt{\left(x_t - x_f\right)^2 + \left(z_t - z_f\right)^2} - |z_f|, \tag{2}$$

where the delay time of the excitation signal applied to the element at the center of the aperture is zero and axial position z_f of virtual source is -40 mm.

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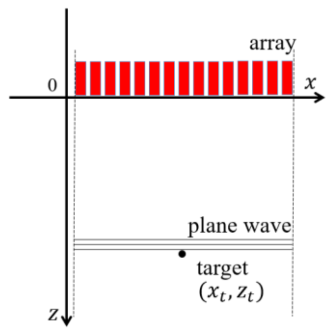

Fig. 1 Illustration of plane wave.

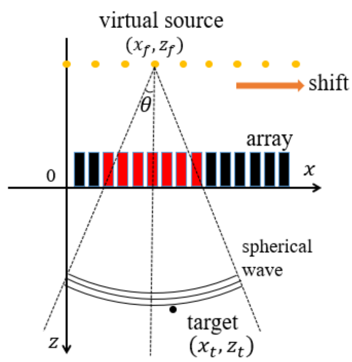

Fig. 2 Illustration of spherically diverging beam.

2.3 Phase-sensitive 2D motion estimator

The principle of phase-sensitive 2D motion estimator[1)] is described below. The 2D (lateral position x, axial position z) ultrasound radio-frequency (RF) signal received in the n-th frame and shifted 2D RF signal in the $(n + \Delta N)$-th frame were used to calculate the 2D cross spectrum. Mean difference α_n between phase $\varphi_n(f_x, f_z)$ of the cross spectrum of 2D ultrasound RF signals and its model $\hat{\varphi}(f_x, f_z) = -\omega_x u_x - \omega_z u_z$ is expressed as

$$\alpha_n = \sum_{f_x, f_z} w(f_x, f_z) \cdot |\varphi_n(f_x, f_z) + (\omega_x u_x + \omega_z u_z)|^2, \tag{3}$$

where f_x and f_z are lateral and axial spatial frequencies, respectively ($\omega_x = 2\pi f_x$ and $\omega_z = 2\pi f_z$), and $w(f_x, f_z)$ is a weight function. The least-square solution of Eq. (3) is obtained by setting the partial derivatives of α_n with respect to u_x and u_z at zero to determine lateral and axial inter-frame displacements $\hat{u}_x$ and $\hat{u}_z$, respectively. Moreover, ω_x and ω_z are estimated from the cross spectrum.

Estimated displacements $\hat{u}_x$ and $\hat{u}_z$ divided by frame interval correspond to average speeds $\hat{v}_x$ and $\hat{v}_z$ between frames.

2.4 Simulation phantom

Figure 3 shows a schematic diagram of the simulated carotid artery phantom. Point scatterers were distributed randomly in the phantom and moved at speeds as described below:

$$v_x = V_l, \tag{4}$$

$$v_y = v_r \sin\theta, \tag{5}$$

$$v_z = v_r \cos\theta, \tag{6}$$

$$v_r = \frac{V_r}{\sqrt{y^2 + (z - z_0)^2}}, \tag{7}$$

$$\theta = \tan^{-1}\frac{y}{z_0 - z}, \tag{8}$$

where v_x, v_y, and v_z are moving speeds in lateral, slice, and axial directions, respectively, and V_l and V_r are moving speeds in longitudinal and radial directions of the phantom. In the present study, V_l and V_r were set at 0 mm/s and 10 mm/s, respectively. In the present study, Field II ultrasound simulation program was used[2,3].

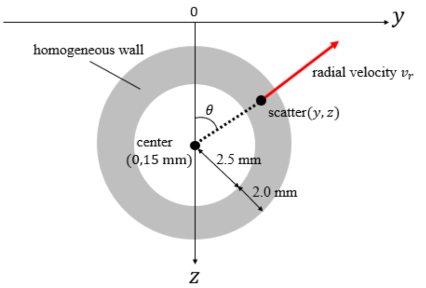

Fig. 3 Schematic diagram of numerical carotid phantom.

2.5 Simulation conditions

In each transmit condition, beamformed RF signals were generated for 8 frames. One frame was obtained with a single emission in plane wave imaging and 21 emissions in synthetic aperture imaging. A pulse repetition frequency is 10417 Hz. The cross spectrum was averaged for consecutive 4 frames before calculating velocities. The lateral and axial window sizes to calculate the cross spectrum were ± 6 mm and ± 0.3 mm, respectively.

The scatterer number density was 10 scatterers in a wavelength cube λ^3, and interferences among echoes occur. Thus, estimated velocities fluctuate due to such interferences. The difference between the accuracies in plane wave and synthetic aperture imaging was small and possibly influenced by such velocity fluctuations. Therefore, simulation phantoms were generated 5 times under each transmit condition. For evaluation of the accuracy, the bias error and root mean square error

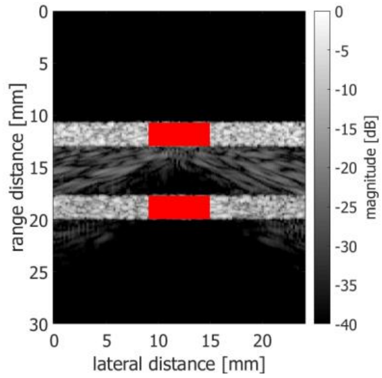

Fig. 4 B-mode image (plane wave imaging) and points of interest (red region).

(RMSE) were calculated. The points of interest were assigned in the anterior and posterior walls of the phantom. **Figure 4** shows B-mode image (plane wave imaging) and points of interest. The points of interest (29×100 points in lateral and depth directions, respectively) were located in anterior and posterior walls.

3. Results and Conclusion

The bias error and RMSE which calculated using the estimated axial velocity and true one are shown in **Table 1.** The synthetic aperture imaging realized smaller RMSEs than the plane wave imaging. The similar bias errors were obtained on the posterior wall under both transmit conditions. On the other hand, the bias error on the anterior wall was smaller in synthetic aperture imaging than plane wave imaging. This is considered that undesirable echoes in the lumen of the phantom were reduced by synthetic aperture imaging. A frame rate of at least around 200 Hz is preferable for measurement of arterial wall motion using a phase-sensitive motion estimator to prevent aliasing. In our future work, other transmit conditions will also be investigated to realize more accurate estimation of arterial wall motion with such a required frame rate.

Table 1 Bias error and RMSE

Method	Anterior wall	Posterior wall
Plane Wave	-1.54±3.25 %	-1.02±2.65 %
Synthetic Aperture	-0.67±1.85 %	-0.99±1.80 %

References

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