1P5-12

Estimation of flow velocity vectors using two types of multi-angle Doppler methods on stenosed blood vessel models

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1. Introduction

Carotid artery lesions are a crucial factor in diagnosing lifestyle-related and arterial occlusive diseases. Ultrasound technology enables the estimation of the stenosis rate in arterial lesions by measuring blood flow, which is then used to assess medical risk. However, the accuracy of traditional blood flow velocity estimation methods is influenced by the angle between the ultrasound probe and the direction of blood flow. To address this issue, the multi-angle Doppler method[1)] was proposed as an advanced ultrasound transmission and reception technique. Additionally, high frame rate ultrasound imaging, with its temporal resolution of several thousand frames per second, can significantly enhance the accuracy of blood flow velocity estimation.

In this study, we utilized plane wave imaging[2,3)] and multi-line transmission imaging[4,5)] techniques developed by our research group as methods for estimating blood flow velocity vectors using multi-angle Doppler and ultrafast ultrasound imaging. These techniques were applied to a vessel model with adjustable stenosis levels, created through fluid mechanics simulation software, to develop a system for evaluating the accuracy of blood flow velocity vector estimation.

2. Methods

2.1 Transmit-receive sequence

2.1.1 Plane wave imaging (PWI) [2,3)]

In this study, the vector Doppler method was implemented in plane wave imaging using the coherent plane wave compound (CPWC) and the repeated transmit sequence. **Fig. 1** shows the method of velocity vector estimation by three-angle transmission and repeated transmission sequences in plane wave imaging. The plane wave is transmitted twice successively at steering angles of $-\theta$, 0, and θ. Increasing the number of transmit directions increases the time interval between each signal pair, but by transmitting each signal pair successively, the time interval caneb minimized to the pulse repetition interval (PRI) and the aliasing limit can be maintained. During reception, delay-and-sum (DAS) beamforming was performed on the signal received at each transducer element, and the beamformed radio frequency (RF) signal was obtained.

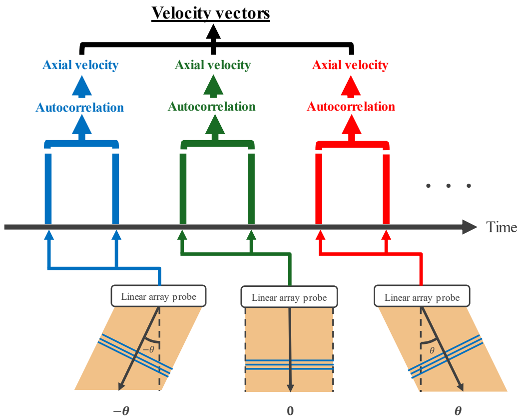

Fig. 1 Illustration of velocity vector estimation using plane wave imaging.

2.1.2 Multi-line transmission imaging (MLTI) [4-6)]

In this study, the vector Doppler method was also applied to the multi-line focused transmit beam sequence. **Fig. 2** illustrates the transmit-receive sequence and velocity vector estimation. During transmission, two parallel ultrasound beams were focused at a depth of 20 mm. The number of elements was 120 and the element pitch was 0.2 mm. Each ultrasonic beam was transmitted twice consecutively at the same position, then shifted by

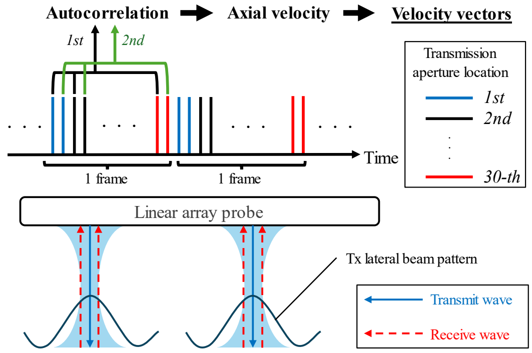

Fig. 2 Illustration of velocity vector estimation using multi-line transmission imaging.

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0.4 mm, and the series of operations is repeated 30 times. The reason for transmitting the ultrasound beam twice successively at the same position is to minimize the time interval between signal pair to the PRI so that the aliasing limit is maintained even with focused beams from running apertures. Two parallel receive lines were set up so that the sound pressure was equal for one focused ultrasound beam, and DAS beamforming was performed to obtain beamformed RF signals at steering angles of $-\theta$, 0, and θ.

2.2 Stenosed blood vessel models

In this study, a stenotic vessel model with variable degrees of stenosis was designed to simulate blood flow around a stenotic lesion in a carotid artery. A cylinder simulating a blood vessel extending along the x-axis direction is expressed by the following equations.

$$y = a + \frac{R}{2}\sin\varphi, \tag{1}$$

$$z = b + \frac{R}{2}\cos\varphi, \tag{2}$$

where R is the inner diameter of the vessel, and (a, b) are the coordinates of the center of the vessel at the location of the stenosis. At this point, a stenotic region can be created in the vessel by varying the vessel inner diameter R within a certain range of x values. This is expressed by the following equation.

$$R = R_{std} - \Delta T(x), \tag{3}$$

where R_{std} is the inner diameter of the vessel without stenosis, and $\Delta T(x)$ is the thickness function expressing stenosis. A function which connects stenotic and non-stenotic areas smoothly is desirable. In this study, we implemented a stenosis model in which $\Delta T(x)$ is sinusoidal.

$$\Delta T(x) = (R_{std} - R_{min})\left(1 + \cos\left(\frac{x\pi}{L}\right)\right), \tag{4}$$

where R_{min} is the vessel inner diameter at the maximum stenotic position, and L is the extent of the stenosis $(-L <= x <= L)$ when $x = 0$ is the maximum stenotic position. In this study, the vessel center of the stenosis was moved only in the z-axis direction, i.e., (a, b) were set at $(0, m\Delta T(x)/2)$. The thickening bias coefficient m takes the range $(0 <= |m| <= 1)$. When $m = 0$, the center axis of the stenotic segment passes through the center of the segment without stenosis, and when $m = -1$, the lumen of the maximum stenotic position touches the anterior wall with no thickening.

2.3 Simulation

In this study, COMSOL Multiphysics® 5.5 fluid dynamics simulation software was used to simulate blood flow in a stenotic vessel model, and the obtained flow velocity was used to place scatterers using MATLAB and Field II [7,8] to simulate the echo signals. The stenotic vessel model was set with a thickening bias coefficient of -1.0, a stenotic segment length L of 15 mm, a vessel inner diameter without stenosis of 5 mm, and an inner diameter of 3.54 mm at the maximum stenotic position so that the short-axis area stenosis rate, called a stenosis index, was 50%. The center frequency was set to 4.80 MHz, and the maximum flow velocity just before the stenotic region was set to 500 mm/s. The steering angle of the ultrasound beam was 15 degrees for PWI and 25 degrees for MLTI.

3 Results and Discussions

Figs. 3(a) and (b) show the estimated lateral velocities in PWI and MLTI, respectively. In this study, we applied the PWI and MLTI proposed by our research group[2,3,5,6] to blood flow simulated by a vascular vessel model with variable levels of stenosis. Then, we succeeded in constructing a numerical simulation environment to evaluate the accuracy of the flow velocity vector estimation.

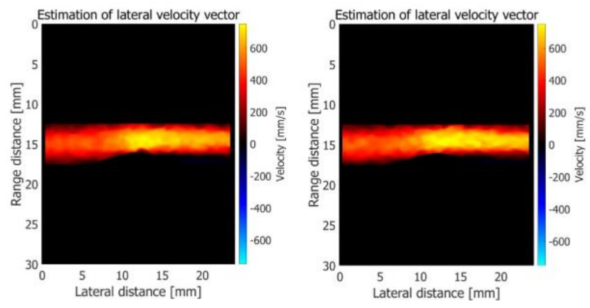

Fig. 3 Estimated lateral velocity in PWI (a) and MLTI (b).

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