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ORIGINAL ARTICLE

Temporal averaging of two-dimensional correlation functions for velocity vector imaging of cardiac blood flow

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Abstract
Purpose Ultrasonic imaging of blood flow in the cardiac lumen is a very useful tool to evaluate the pumping function of the human heart. The speckle tracking technique makes it possible to estimate the blood velocity vector. However, a stable estimation of the velocity vector of blood flow is difficult because signal-to-noise ratios of echoes from tiny blood particles are low. In this study, the speckle tracking technique with averaging of multiple two-dimensional correlation functions was employed for stable estimation of the blood velocity vector.
Methods Multiple two-dimensional correlation functions can be averaged during a very short period by using the echo data acquired by high-frame-rate echocardiography with diverging beam transmission. A steady flow experiment using blood-mimicking fluid (mean fluid velocity 0.2 m/s, flow angle 56° from the transducer surface) was implemented to investigate the effect of the averaging of two-dimensional correlation functions at a frame rate of 6024 Hz.
Results First, to examine the averaging duration required for stable estimation of the flow velocity vector, the accuracies of vector estimates were evaluated at different durations for averaging of two-dimensional correlation functions in the steady flow measurement. It was found that the proposed averaging process with an averaging duration of over 8 frames could reduce the directional error in vector estimation to almost half that of the conventional speckle tracking technique. In subsequent experiments, the averaging duration was set at 12 frames corresponding to 2 ms. Measurements of steady flow at higher velocities were further implemented. The steady flow measurements with higher flow velocities of 0.4 and 0.6 m/s were simulated by changing the frame interval of the echo data at a flow velocity of 0.2 m/s. Although the averaging duration was a mere 2 ms, directional errors at mean flow velocities of 0.2, 0.4, and 0.6 m/s were reduced significantly. In an in vivo experiment of the healthy human heart, to produce a fine B-mode image, the diverging wave transmissions with different steered angles for compounding were interleaved in the transmission sequence. From the in vivo experimental result, the blood velocity vector of the left ventricular cavity showed the flow getting into/out of the cavity in ejection and early diastolic phases. Furthermore, estimated flow directions revealed rotating flow in the cavity in mid-diastole.
Conclusion Our proposed method has the feasibility to visualize the vortex flow by velocity vector mapping without a contrast agent.

Keywords Cardiac blood flow · Blood velocity vector · Speckle tracking · High-frame-rate acquisition

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Introduction

Ultrasonic imaging of blood flow in the cardiac lumen is a very useful tool to evaluate the pumping function of the human heart. This has been a major diagnostic modality to investigate problems of the human heart [1]. Color Doppler flow imaging (CDI) is a typical method to obtain the information of intracardiac blood flow and is on-board in most commercial systems. Recently, the frame rate of

Doppler flow imaging was dramatically improved using parallel beamforming with plane wave transmission [2, 3]. The high-frame-rate imaging with plane wave transmission has been employed for measurement of the propagation velocity of the pressure wave in a local region in the carotid artery [4, 5]. CDI, however, does not show the direction of blood flow because only velocity along an ultrasonic beam is measured.

In a different approach, the transverse component (perpendicular to the ultrasonic beam) of blood flow velocity has been mathematically calculated from the distribution of the axial velocity obtained by CDI on the basis of hydrokinetic assumptions [6]. In addition, another method with the continuity equation using velocities of surrounding heart walls as boundary conditions has been proposed [7]. Recently, for improvement of the accuracy of the method, optimal smoothing of the axial velocity distribution and weighting at the bilateral velocities of the heart walls have been examined [8]. However, such assumptive approaches cannot visualize real blood flow.

On the other hand, the blood velocity vector was estimated by combining Doppler measurements from two or more directions [9–11]. This application, however, requires a large angle formed by the intersection of beams; thus, it is limited to blood flow in a shallow region. A phased-array transducer with a small aperture, which is normally used in cardiac measurement, is not suitable for such a combined Doppler method. As a result, a linear-array transducer with a large aperture is being used in ongoing studies.

Echocardiographic particle image velocimetry (E-PIV) realizes the mapping of velocity vectors and stream lines of blood flow that are computed on the basis of motions of ultrasonic echoes from a contrast agent [12, 13]. However, the intravenous injection of a contrast agent is required. On the other hand, Løvstakken et al. [14] visualized motions of echoes from blood particles utilizing CDI with customized beam sweeping without the use of a contrast agent. To solve the fatal issue of an insufficient frame rate with this method, Hasegawa and Kanai [15] have developed a method for high-frame-rate imaging of echoes from blood particles inside the carotid lumen at a frame rate of several kilo-hertz with plane wave transmission and parallel receive beamforming using a linear-array transducer. In cardiac ultrasound, the ultrasonic beam is swept in a sector fashion. Hence, such plane wave imaging significantly limits the frame rate in cardiac measurement because the widths of insonified plane waves do not increase with the range distance from the transducer in the near field, whereas lateral intervals of scan lines increase with range distance.

For visualization of blood flow with a velocity of over 1 m/s and complex flow patterns such as vortex flow in the cardiac lumen, high-frame-rate measurement is desirable. In our previous study, we proposed a method for visualization of the direction of intracardiac blood flow using high-frame-rate measurement with a spherically diverging beam [16, 17]. In that study, blood velocity vectors in the cardiac lumen could be estimated by speckle tracking [18]. However, it was confirmed that the accuracy was not sufficient even in steady flow experiments. In addition, the quality of the B-mode image was much lower than that of the conventional image due to the number of compounding per frame of a mere two or three.

In the present study, we proposed an improved estimator of blood velocity vectors. In the proposed estimator, two-dimensional correlation functions (2D-CFs) between consecutive frames were temporally averaged for stable estimation of blood velocity vectors, and the averaged correlation function was up-sampled. High-frame-rate measurement with transmission of diverging beam contributes to suppress degradation of the temporal resolution in the vector estimation suffered from averaging of 2D-CFs because multiple 2D-CFs during a very short period can be used owing to the high-frame-rate measurement. Furthermore, to acquire a B-mode image with better image quality, a large number of steered diverging beams for compounded B-mode imaging were interleaved before every transmitting event for estimation of blood velocity vectors.

Materials and methods

Echo data acquisition scheme

The acquisition of echo signals at a high frame rate is implemented with parallel receive beamforming with transmission of spherically diverging beam (PBF-DB) [19, 20]. With PBF-DB, the ultrasonic beam is diverged from a virtual point (sound) source (VPS) placed behind the array surface. Then, multiple receiving beams focusing on different spatial positions are generated using echo signals received by transducer elements in parallel. Therefore, PBF-DB can facilitate high-frame-rate measurement of echo signals from the human heart by decreasing the number of transmissions per frame. The lateral spatial resolution of PBF-DB is, however, degraded due to low directivity of the transmitted ultrasound. To compensate for it, compounding of echo signals obtained by diverging beam transmissions with different steering angles is effective [19]. Although the lateral resolution after compounding is further improved by increasing the number of transmissions with different steering angles, the frame rate is decreased due to the increase in the number of transmissions per frame. For estimation of the velocity vector of blood flow with a velocity of over 1 m/s and complex flow like vortex flow in the cardiac lumen, high-frame-rate measurement is desirable to suppress the signal

decorrelation caused by the variation of relative distances between scatterers, which corresponds to blood particles.

In this study, a large number of steered diverging beams for B-mode imaging were interleaved before every transmitting event for acquisition of blood velocity vectors. Hence, the transmission sequence comprised transmissions of steered diverging beams for acquisition of compounded B-mode images and single non-steered diverging beams per frame for estimation of the velocity vector of blood flow, as illustrated in Fig. 1a. In the transmission event for B-mode imaging, 15 diverging beams with angular intervals of 6° were transmitted, and the corresponding beamformed signals were compounded to produce the B-mode image with better image quality, as shown in Fig. 1b. On the other hand, in the transmission event for blood vector acquisition, a single non-steered diverging beam was transmitted and the beamformed signals were generated without compounding, as shown in Fig. 1c. This diverging beam was transmitted in the same direction in 40 frames for a robust estimation of the velocity vector by 2D-CF averaging, as described in next section. Finally, the acquisition frame rate of the B-mode image overlaid with the distribution of the blood velocity vector is defined as:

$$\mathrm{FR} = \frac{\mathrm{PRF}}{N_{\mathrm{tx1}} + N_{\mathrm{tx2}}} \quad (1)$$

where PRF, N_{tx1}, and N_{tx2} are the pulse repetition frequency, the number of steered diverging beam transmissions for B-mode imaging, and the number of single diverging beam transmissions for blood vector estimation, respectively. In this study, PRF, N_{tx1}, and N_{tx2} were set at 6024 Hz, 15, and 40, respectively. The VPSs in transmission events of B-mode and blood velocity vector imaging were placed at 100 and 50 mm behind the array surface, respectively.

Echo signals were measured by a modified diagnostic ultrasound system (Hitachi-Aloka Medical Alpha 10). The phased-array transducer was operated at a center frequency of 3.75 MHz. Individual RF echo signals received by 96 elements were sampled at 15 MHz. Using these receive echoes, echo amplitudes on 241 scan lines with an angular interval of 0.375° were created by delay-and-sum focusing. Figure 1d shows an illustration of the created scan lines. These beamforming processes were performed in all transmissions for B-mode imaging and mapping blood velocity vectors. Note that the receive beamforming and the proposed method were implemented with off-line processing on a personal computer.

Estimation of blood velocity vector

At the transmission event for blood vector acquisition, the receive-beamformed signals were high-pass filtered in the direction of the frame (called moving target indication filter) to visualize echoes from blood particles by suppressing clutter echoes from stationary and slowly moving tissues except for blood such as the ribcage and myocardium. High-pass filtering was performed by a 12th order Butterworth filter whose cutoff frequency was set to 500 Hz corresponding to the axial velocity of almost 0.1 m/s at the ultrasound frequency of 3.75 MHz. The amplitude of filtered signals was weighted by the coherence function of signals filtered in the direction of the frame to enhance echoes from blood particles coherently moving [17]. The number of frames used for computation of the coherence was determined so that echoes from almost the same scatterers were used during the period. Here,

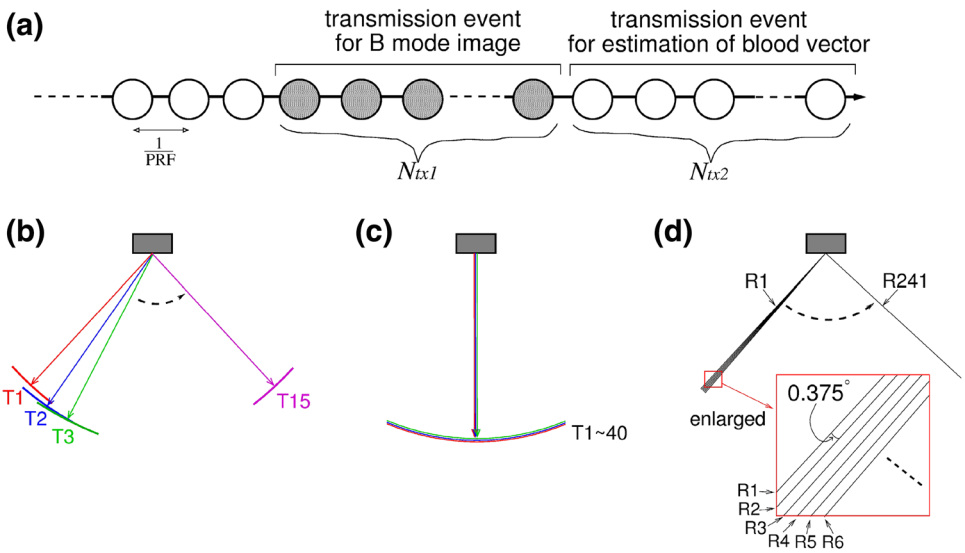

Fig. 1 **a** Emission sequence of spherically diverging ultrasound and illustration of wavefront of transmit ultrasound in events for, **b** B-mode imaging and **c** estimation of blood velocity vector. **d** Illustration of positions of scan lines created by receive beamforming in every transmission

we estimate the elapsed time in which blood cells pass through the ultrasonic beam transversely. The width at half-maximum of the lateral profile of an echo from a fine wire (i.e., spatial resolution) is almost 1.9 mm. On the assumption that the transverse velocity (perpendicular to the ultrasonic beam) of blood flow is 1 m/s, a blood cell is estimated to slip from the ultrasonic beam during 11 frames at a frame rate of 6024 Hz. In the present study, the number of frames for the coherence estimation was set at neighboring 6 frames to fill this necessary condition (<11 frames) for the use of echoes from the same scatterers.

The speckle tracking technique with computation of 2D-CF makes it possible to estimate the blood velocity vector without depending on the angle between an ultrasonic beam and the blood flow [18]. However, blood cells are small and acoustic outputs of defocused ultrasonic beams employed for high-frame-frame measurement are low. Under such conditions, signal-to-noise ratios (SNRs) of echoes from blood cells are low; therefore, it can be challenging to accurately estimate the blood velocity vector. In this study, to compensate for this issue, the velocity vector of blood flow was estimated by speckle tracking with 2D-CF averaging for a stable estimation of the velocity vector.

There are several types of averaging operations through multiple frames for estimation of the velocity vector: (1) averaging of velocity vectors computed in each frame; (2) averaging of unnormalized 2D-CFs; (3) averaging of 2D-CFs normalized by the variance. Low SNRs of echoes from tiny blood particles can derive the peak at the wrong position (displacement) on the 2D-CF because the correlation coefficient at the true peak decreases due to low SNRs. The rise in the correlation coefficient at the wrong position on the 2D-CF should be suppressed to avoid errors in estimation of the velocity vector. The correlation coefficient at the true position on the 2D-CF is considered to be high on average during a short duration. On the other hand, deviation of the correlation coefficient caused by the noise component in echoes can be assumed to be random with respect to each inter-frame, which causes the wrong peak on the 2D-CF. Hence, the averaging operation of 2D-CFs for the short period is effective to produce the peak at the true position on the 2D-CF by suppressing the rise in the correlation coefficient caused by low SNRs of echoes. In the present study, the 2D-CF normalized by the variance was used for the averaging operation to emphasize the temporal change in the phase of echoes for estimation of the flow velocity vector.

Normalized 2D-CFs between two consecutive frames at a specific spatial point are temporally averaged N_{cf} times (the number of inter-frame) as follows:

$$\bar{C}(i,j,n) = \frac{1}{N_{cf}} \sum_{k=-N_{cf}/2}^{N_{cf}/2-1} C(i,j,n+k) \quad (2)$$

where $C(i, j, n)$ denotes the normalized 2D-CF of the lateral and axial lags of i and j, respectively, between the nth and $(n + 1)$th frames. Figure 2 illustrates the proposed averaging process of normalized 2D-CFs. Then, the averaged 2D-CF is up-sampled with a reconstructive interpolation to follow a small displacement between consecutive frames below the sampling interval [21].

The spatial resolution cell of the speckle tracking technique is regarded as the size of the spatial window for computation of 2D-CFs. Let us define this spatial window as the kernel. The kernel sizes in the lateral and axial directions are set to 9° (corresponding to 24 lines) and 8.2 mm, respectively. The signal amplitude in the kernel is weighted by the two-dimensional Hann window function. The effect on accuracy of blood vector estimation from the difference in N_{cf} is examined in the following section.

Steady flow measurement

Experimental setup

The aim of this experiment was to examine the effect of 2D-CF averaging. Figure 3 illustrates the experimental

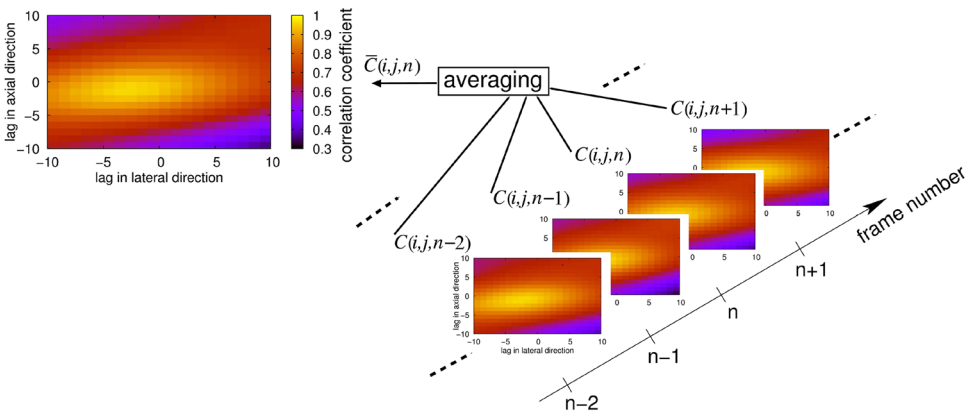

Fig. 2 Schematic illustration of temporal averaging of normalized 2D-CFs for stable vector estimation

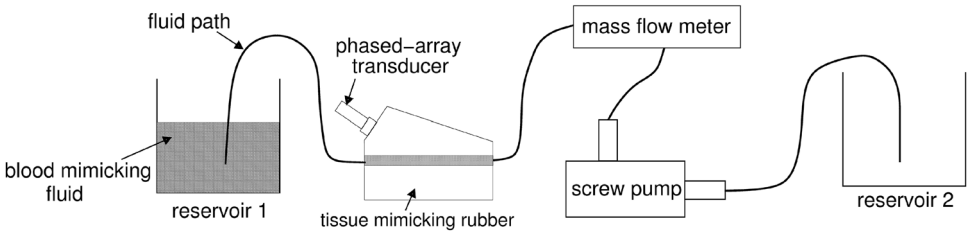

Fig. 3 Experimental system for measurement of steady flow. The blood-mimicking fluid was moved steadily through the flow path by a screw pump

setup. Echoes from a blood-mimicking fluid (ATS 707) flowing in a cylindrical flow path surrounded by a tissue-mimicking rubber (ATS 523A) were measured. The diameter of the flow path was 8 mm. The transducer surface was placed at a distance of 7 cm from the outer surface of the flow path. The angle between the transducer surface and the flow path was kept at 56°. The blood-mimicking fluid was kept flowing steadily using a screw pump (Heishin 2NL10PU). The flow rate was measured by a mass flow meter (Keyence FD-SF8) connected to the flow path.

The echo data were measured at a flow rate of 0.6 L/min. Higher flow rates of 1.2 and 1.8 L/min were simulated by changing the frame interval for estimation of the correlation function, i.e., twice and triple the original frame interval. Therefore, a Reynolds number of 960 at a flow rate of 0.6 L/min was maintained across these simulated data. These flow rates of 0.6, 1.2, and 1.8 L/min corresponded to the mean fluid velocities of 0.2, 0.4, and 0.6 m/s, respectively. The step size of the flow rate measured by the mass flow meter was 10 mL/min. The echo data were acquired at a frame rate of 6024 Hz with the single transmission of a non-steered diverging wave per frame. The VPS was located at 50 mm behind the array surface.

Experimental results

Velocity vectors of the fluid inside the flow path were obtained at an interval of 0.4 mm along the manually assigned line perpendicular to the flow axis, as shown in Fig. 4.

To examine the averaging duration required for stable estimation of flow velocity vectors, directional errors of estimated velocity vectors from the direction of the flow path were evaluated at different durations of averaging. The directional error e_{dir} was estimated as follows:

$$e_{\mathrm{dir}} = \frac{1}{\pi}\sqrt{\frac{1}{N_j}\sum_{j=0}^{N_j-1}\left\{\cos^{-1}\left(\frac{\mathbf{v}_j\cdot\mathbf{n}}{|\mathbf{v}_j|\,|\mathbf{n}|}\right)\right\}^2} \tag{3}$$

where $\mathbf{v}_j$ and $\mathbf{n}$ represent the estimated velocity vector at the jth spatial point and the unit vector parallel to the flow direction, respectively. Figure 5 shows the directional errors of estimated flow vectors at the different numbers of frames for averaging of 2D-CFs at a flow rate of 0.6 L/min (a mean fluid velocity of 0.2 m/s). The standard deviations

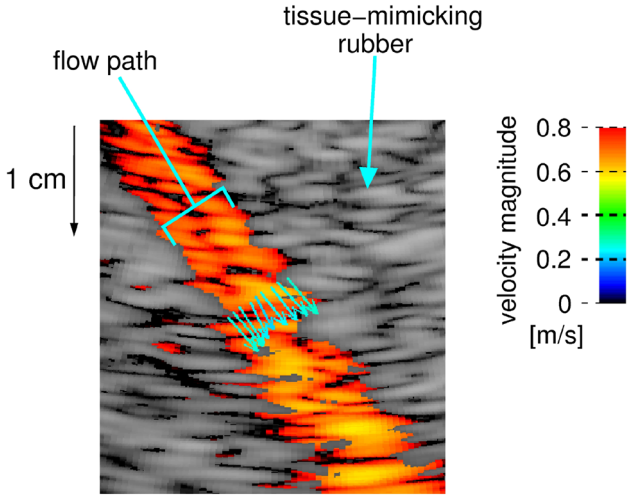

Fig. 4 Velocity vectors overlaid with B-mode image (*gray scale*) and echo speckles of blood-mimicking fluid (*hot scale*). A subset of estimated velocity vectors along the direction perpendicular to the flow axis was visualized

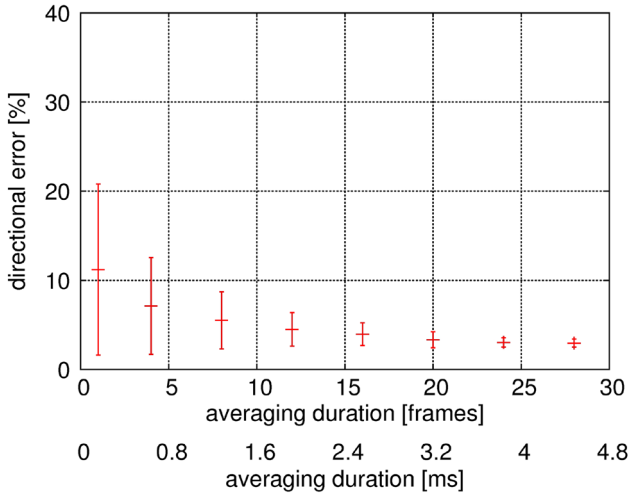

Fig. 5 Directional errors with respect to the various numbers of frames for averaging of normalized 2D-CFs at a flow rate of 0.6 L/min (a mean fluid velocity of 0.2 m/s). The standard deviation was computed from 50 frames

in Fig. 5 were computed from 50 frames. The deviation of directional errors in the vector estimation by the speckle tracking technique with averaging of normalized 2D-CFs was reduced to almost half that of the conventional speckle tracking technique (non-averaging of 2D-CFs) at an averaging duration of over 8 frames. In the present study, N_{cf} was set to 12, which corresponds to 2 ms, to prioritize the temporal resolution in the estimation of velocity vectors.

Velocity vectors estimated with and without averaging 2D-CFs during 12 frames were compared based on flow rates (mean flow velocities) and directional errors. Figure 6a shows flow rate Q estimated by the use of the velocity vectors as follows:

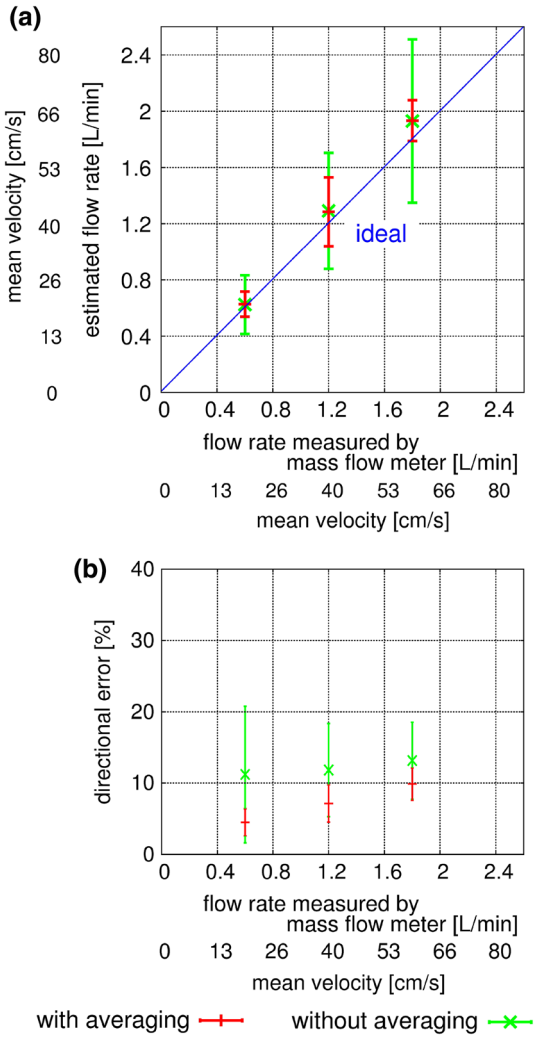

Fig. 6 **a** Flow rates (mean fluid velocities) of blood-mimicking fluid estimated by the proposed method and a mass flow meter and **b** directional errors between the flow axis and velocity vectors estimated by the proposed method. The *red* and *green crosses* indicate estimates yielded by speckle tracking with and without averaging of normalized 2D-CFs. The *horizontal axis* represents the flow rate, which was increased by changing the frame interval in the echo data acquired at a flow rate of 0.6 L/min (a mean fluid velocity of 0.2 m/s) measured by the mass flow meter

$$Q = \pi \sum_{j=0}^{N_j-1} \left(r_j \Delta r\right) \mathbf{v}_j \cdot \mathbf{n} \tag{4}$$

where Δr and r_j denote the interval between points at which the vectors were estimated and the distance from the center of flow path, respectively. Figure 6b shows directional errors of the estimated velocity vectors from the direction of the flow path. The standard deviations in Fig. 6a, b were computed from 50 frames. In Fig. 6a, b, the red and green crosses indicate estimates yielded by the speckle tracking technique with and without the averaging operation of 2D-CFs, respectively. Although the averaging duration of 2D-CFs was a mere 2 ms, both errors were significantly reduced by 2D-CF averaging.

In vivo experiment

RF echo signals from the left ventricle were acquired in a transthoracic three-chamber view of a 27-year-old healthy male. The healthy subject gave informed consent to the study. N_{tx1} and N_{tx2} were 15 and 40, respectively; therefore, the eventual frame rate was 109 Hz, as described in "Echo data acquisition scheme". One blood vector image was obtained using 40 successive transmissions of non-steered diverging beams. The VPSs used for B-mode and blood velocity vector imaging were located at 100 and 50 mm behind the array surface.

Figure 7 shows the information obtained for blood flow overlaid on compounded B-mode images in frames in early systole, early diastole, and mid-diastole. As for the blood flow information, the cyan-colored arrow and the intensity of the hot color indicated the direction and the velocity magnitude of the blood velocity vector, respectively. The vector magnitudes at all pixels were obtained using reconstructive interpolation of the vector velocity field obtained by the speckle tracking technique with 2D-CF averaging. The mitral valve and the aortic valve are located to the basal side of the left ventricle. As shown in Fig. 7a, b, the fluxes flowing into and out of the cardiac cavity were visualized by blood flow vectors in early systolic and early diastolic phases. In mid-diastole, the vortex-like flow appeared as shown in Fig. 7c. Recently, complex flow such as vortex formation has been verified and studied by magnetic resonance imaging [22] and E-PIV. The proposed method shows great potential for visualizing vortex flow using contrast-free ultrasound.

Discussion

In the case of conventional E-PIV, it has been reported that the frame rate (below 200 Hz, corresponding to the

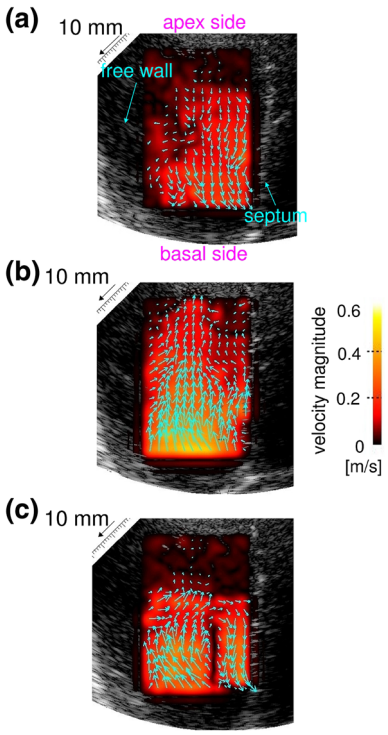

Fig. 7 Velocity vector images of the cardiac lumen in **a** systole, **b** early diastole, and **c** mid-diastole. The direction and the velocity magnitude of the blood velocity vector are indicated by the *cyan-colored arrow* and the intensity of hot color

temporal resolution of over 5 ms) is not sufficient to estimate the flow velocity vector inside the cardiac lumen [23]. Normalized 2D-CF computing between a very short frame interval (a couple of hundred microseconds) comes from the high-frame-rate measurement with diverging wave transmission. In this study, speckle tracking with averaging of normalized 2D-CFs was proposed for stable estimation of blood velocity vectors. The velocity vector obtained by averaging of 2D-CFs means a coherent component of blood flow vectors during the averaging duration. In the proposed method, normalized 2D-CFs of 12 frame intervals could be used for the averaging process during the very short duration of 2 ms owing to very high-frame-rate acquisition. Although the averaging duration was only 2 ms, the accuracy of the velocity vector estimator was strongly improved, as shown in the experimental result of steady flow. Therefore, fine blood vector estimation with the high temporal resolution of 2 ms can be achieved with our proposed method.

In the current sequence for mapping blood velocity vectors, there were unnecessary transmissions for the averaging duration of 12 frame intervals. From echoes in 40 frames, echoes in 27 frames were obtained by 12th order Butterworth filtering. Because the coherence was computed using echoes in 6 frames after Butterworth filtering, echoes in 21 frames weighted by the coherences were obtained, which was greater than the 12 frames required for averaging of 2D-CFs. The number of acquired frames for blood flow imaging can be reduced in optimization of the acquisition sequence.

With respect to cardiac blood flow imaging, the visualization of complex flow such as vortex flow has become important. As shown in Fig. 6b, the directional error of the estimator for the velocity vector was reduced from 11.2 to 4.5 % at a mean fluid velocity of 0.2 m/s by averaging of normalized 2D-CFs. A vector estimator with a better direction accuracy helps the visualization of vortex flow patterns. Furthermore, the diverging wave transmissions with different steered angles for compounding were interleaved in the transmission sequence to produce a fine B-mode image. Nevertheless, the frame rate in our study is comparable to that of several tens of hertz of conventional CDI; moreover, the blood velocity vector in the cardiac lumen can be estimated, which is the technical merit of our proposed method.

Conclusions

In this study, velocity vector estimation by the speckle tracking technique with averaging of normalized 2D-CFs was proposed for stable estimation of the blood flow vector in the cardiac lumen. The accuracies of estimates of flow rate (corresponding to flow velocity) and direction were evaluated by means of a steady flow experiment with a blood-mimicking fluid. The experimental results confirmed that the accuracy was improved by the proposed averaging procedure for normalized 2D-CFs. Furthermore, an in vivo experiment of healthy human heart was performed using an interleaved sequence of ultrasound transmission. This transmission sequence alternated transmissions of steered diverging beams for fine B-mode imaging and transmissions of a single diverging beam per frame for estimation of the blood velocity vector by means of speckle tracking with averaging of normalized 2D-CFs. The distribution of intracardiac blood vectors with the compounded B-mode image can be obtained at a frame rate of 109 Hz, which is comparable to conventional CDI. The blood flowing into/out of the cardiac cavity in ejection and early diastolic phases, and the vortex flow in diastole, could be visualized in the distribution of blood velocity vectors.

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Conflict of interest There are no financial or other relations that could lead to a conflict of interest.

Ethical standard The in vivo experiment in this study was approved by the institutional ethical committee. The subject gave informed consent to this study.

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