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Improvement of accuracy and computational efficiency in intracardiac blood velocity estimation
心腔内血流速度推定における精度および計算効率の向上

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1. Introduction

In recent years, for assessment of the cardiac function, visualization of dynamic motions of echoes from blood cells in the human heart with high-frame-rate echocardiography [1] has been studied and estimation of two-dimensional (2D) velocity vector of the cardiac blood flow by using the echo motions was shown to be feasible [2].

The accuracy in estimation of inter-frame displacements of echoes is improved with an increase of the spatial frequency of an echo. In addition, speckle patterns resulting from interferences of echoes from blood cells change rapidly because blood cells in the cardiac blood flow move complexly. Hence, the spatial freqeuncy bandwidth of an echo would be desired to be high to reduce interfering echoes. The echo bandwidth in the transverse direction (perpendicular to ultrasound beam) is changed with the receive apodization.

The blood velocity vector are estimated by a method using the correlation function of echoes between consecutive frames, namely, speckle tracking (ST) [3]. Echoes from blood cells are measured at a high frame rate of several thousand Hertz to capture fast (over 1 m/s) and complex flows in the cardiac cavity. The correlation function needs to be upsampled to estimate sub-sample displacements within hundreds of microseconds, which makes the computational load high.

In this study, the effect of the receive apodization on estimation of 2D blood flow velocities with high-frame-rate echocardigraphy was examined. Furthermore, a velocity vector estimator with 2D Fourier transform (VEFT), which requires no upsampling of the correlation function, was proposed to reduce the computational load.

2. Principles

An in-house computer simulation of tubular flow and basic experiment with a flow phantom (model 523A, ATS, USA) were performed to evaluate the accuracy of the 2D velocity estimators. An array ultrasound probe with 96 transducer elements at a nominal center frequency of 3.75 MHz was used and equipped on an ultrasound system enabled to acquire echoes received by individual elements. The diverging ultrasound from a virtual point source positioned at 30 mm behind the array was transmitted and beamformed echo signals were obtained with parallel receive beamforming [1]. The frame rate was 6250 Hz.

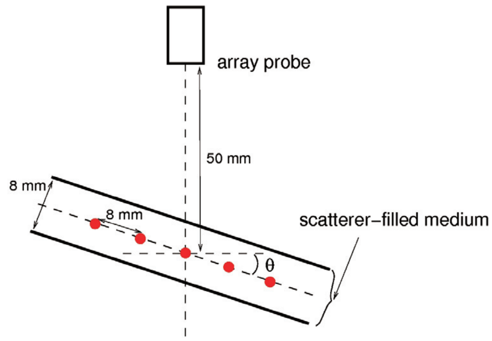

Fig. 1 Illustration of simulated phantom.

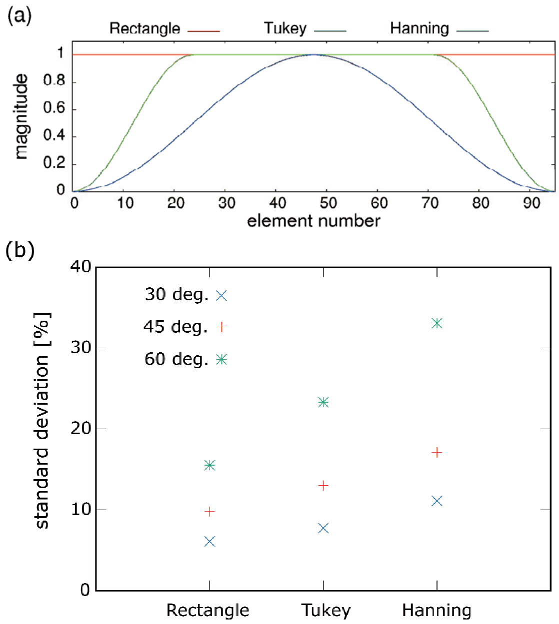

Fig. 2 (a) Apodization functions and (b) standard deviations of transverse velocity estimated by ST.

In the simulation, echoes from a large number of point scatterers filled in a tube were computed, as illustrated in **Fig. 1**. Movements of scatterers were governed by Hagen-Poiseuille equation at a velocity of 0.4 m/s. 2D velocity vectors were estimated in 5 red points shown in Fig. 1 and the

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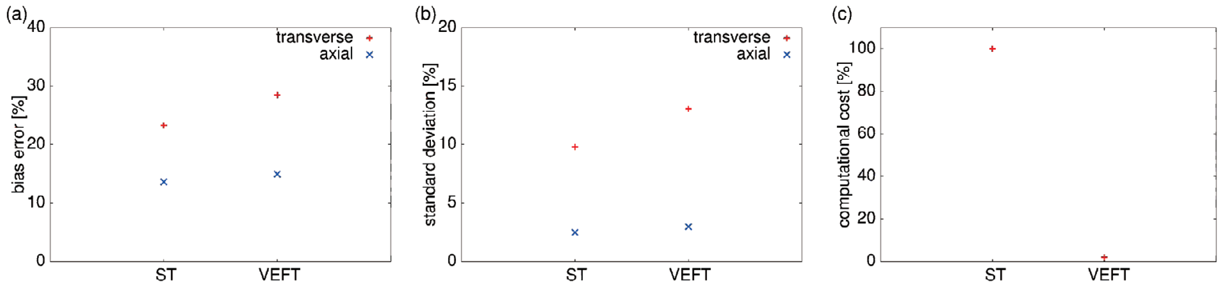

Fig. 3 (a) Bias errors and (b) standard deviations in 2D velocity estimation with ST and VEFT, and (c) computational cost normalized by that of ST.

standard deviations in 20 frames and bias errors were evaluated. **Figure 2(a)** shows apodization functions used for the receive beamforming in the simulation study. The correlation function for ST was upsampled at factors of 30 and 200 in axial and transverse directions, respectively, which realized resolutions in velocity estimation of about 10 mm/s in both directions at a depth of 50 mm.

The 2D frequency spectrum of the beamformed echo signal was used for VEFT. The axial and transverse velocities were computed by the least-square fitting between temporal changes in phase of measured echoes and a linear model corresponding to the group delay between echoes in two consecutive frames. The accuracy of velocities obtained by VEFT was compared with that of ST in the simulation and phantom experiments. The computational time was also measured on a personal computer (Core i7-4710MQ, Intel, USA). Furthermore, blood velocity vectors of a healthy human heart were estimated by VEFT.

3. Results and Discussion

Figure 2(b) shows means of standard deviations of transverse velocities estimated by ST under flow angles θ of 30, 45, and 60 degrees. The standard deviation of the estimated velocities decreased with the transverse bandwidth of the beamformed signal (corresponding to receive apodizations in Fig. 2(a)) at all flow angles. The difference in the standard deviations between rectangular and Hanning apodizations reached to almost 15% at a flow angle of 60 degree.

Figure 3 shows mean bias errors and standard deviations for 2D velocities obtained with VEFT at a flow angle of 45 degree. The rectangular receive apodization was used. The bias error and standard deviation in VEFT were slightly higher than that in ST, but the computational time was drastically reduced to 1.9% of ST. In the basic experiment, the rectangular apodization was also effective to decrease the standard deviation of the transverse velocity as shown in **Fig. 4**. In the *in vivo* experiment, the diastolic vortex-like flow was visualized by VEFT as shown in **Fig. 5**.

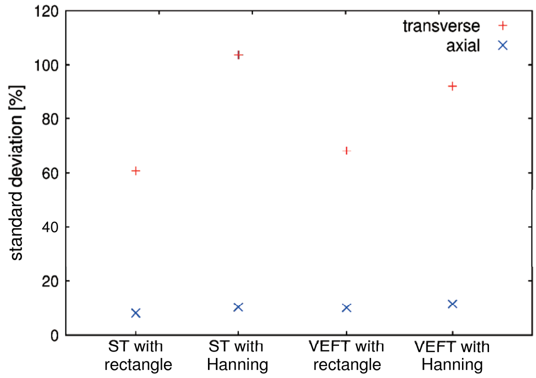

Fig. 4 Standard deviations of 2D velocities in basic experiments with flow phantom.

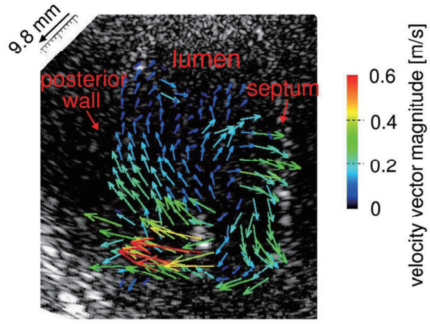

Fig. 5 Distribution of 2D blood velocity vectors within left ventricle in mid diastole estimated by VEFT.

4. Conclusion

The rectangular apodization was most effective for a stable estimation of 2D blood velocities. The computational load was significantly reduced by VEFT at the expense of slight increase of the bias error and standard deviation.

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