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Abstract— **An algorithm which measures the lateral component of blood flow velocity was developed in our previous studies based on the increase in speckle size due to relative motion between moving scatterers and spatial rate of scanner A-line acquisition (scan velocity). In this paper, the apparent dominant angle of the speckle pattern in a straight vessel was investigated and a new method of two-dimensional blood flow velocity estimation is introduced. Different scan velocities were used for data acquisition from blood flow traveling at an angle relative to the ultrasound beam. The apparent angle of the speckle pattern changes with different scan velocities due to mis-registration between the ultrasound beam and scatterers. The apparent angle of the speckle pattern was resolved by line-to-line cross-correlation in the fast time (axial) direction on a region-of-interest (ROI) in each blood flow image and used to spatially align the ROI. The resulting lateral speckle size within the aligned ROI was calculated. The lateral component of the blood flow is shown to be closest to the scan velocity which gives the maximum speckle size and the apparent angle of speckle pattern collected by this scan velocity is the best estimate for the actual angle of blood flow. These two components produce two-dimensional blood flow velocity estimations. Blood flow data were collected from a blood flow phantom with a 50^o beam-to-flow angle. Nine scan velocities were used to collect data for three different actual velocities. Estimation results for the 2-D velocity magnitude (mean ± std) were 40.2 ± 10.1, 61.8 ± 9.3, and 96.8 ± 12.3 cm/s for actual velocities of 41, 65, and 98 cm/s respectively. Estimation results for the angle (actual 50^o for all tests) were 52.7 ± 7.8, 51.6 ± 6.2, and 53.8 ± 5.2^o. These results indicate a promising new way to estimate 2-D blood flow velocity.**

Keywords—blood flow measurement; speckle pattern; 2-D; scan velocity;

I. Introduction

Blood flow measurement is one of the major functions of medical ultrasound imaging systems. To measure the velocity of blood flow noninvasively, an ultrasound beam is emitted by an ultrasound transducer into the human body. Ultrasound waves will then be reflected back to the transducer by the scatterers in the blood (typically red blood cells). The signal constituted by samples from successive received A lines (the "slow time" signal) contains frequency components proportional to the blood velocity along the ultrasound beam [1]. Using the Doppler equation, the velocity distribution at the sampling depth can be converted from the frequency spectrum, which is known as spectral Doppler. Instead of emitting the ultrasound beam in one direction, color Doppler performs data acquisition similar to spectral Doppler, but with much fewer samples, in different directions and then estimates the velocity at different depths. The velocity can be found through an autocorrelation approach developed by Kasai et al. [2], [3]. Although these two Doppler based methods have been widely investigated and commercially implemented, a limitation with these velocity estimation techniques is that only the 1-D projection of the blood velocity vector along the axial dimension of the ultrasound beam can be found. Since complex fields are present throughout the arterial system, multidimensional flow imaging is desirable. This would allow better assessment of the flow field and its associated hemodynamic parameters, and hence potentially improve cardiovascular risk assessment [4].

A number of alternative, 2-D estimation algorithms have been developed for resolving complex flow patterns by estimating both the axial and lateral velocity component. Including vector Doppler [5]-[8], speckle tracking [9], [10], feature tracking [11], [12] and transverse spatial modulation [13], [14]. Other techniques like multidimensional spectrum analysis [15] and maximum likelihood blood velocity estimation [16] have also been proposed. A more complete review can be found in references [17] and [18].

Our previous studies introduced a new way to measure the lateral component of blood flow using the change of apparent speckle size due to the direction and spatial rate of scanner A-line acquisition [19]. In this paper, instead of resolving the flow purely lateral to the ultrasound beam, a new method of two-dimensional blood flow velocity measurement based on the apparent angle of the speckle pattern is proposed.

II. Background

In our previous studies [19], the flow purely lateral to the ultrasound beam was collected by scanning in the same direction as the blood flow. A similar scanning geometry was used in this paper to collect data from the blood flow with a specific angle relative to the ultrasound beam axis. Different scan velocities were used for data acquisition. An ROI was selected from each B-mode image, an example of which can be seen in Fig. 1. Line-by-line cross-correlation in the fast time (axial) direction was used to resolve the apparent angle of speckle pattern, which can be described as follows:

$$R_{xy}(s) = \begin{cases} \sum_{n=0}^{N-s-1} x_{n+s} y_n^* & s \geq 0 \\ R_{yx}^* & s < 0 \end{cases} \tag{1}$$

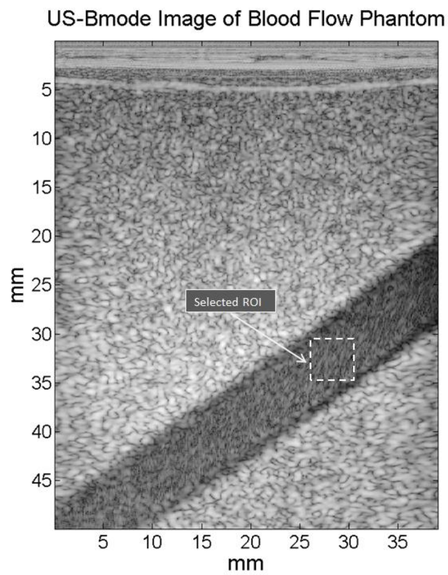

Fig. 1. A Region-of-interest (ROI) is selected from every US B-mode image of the blood flow phantom, which covers an area corresponding to 3 mm × 4.2 mm.

where x and y represent two successive A-lines from the ROI and s is the lag between these two A-lines. The cross-correlation coefficient was calculated for each s, and the value s which gives the maximum cross-correlation coefficient between x and y was recorded as their distance shift d_{xy}. This process was done for every pair of successive A-lines in the ROI and the d_{xy} with the highest frequency of occurrence (i.e., the mathematical mode) was considered to be the overall distance shift of the ROI, which is d_{ROI}. Given the line increment of scanning, which is represented by d_{incre}, the apparent angle θ_a of speckle pattern was calculated as:

$$\theta_a = \tan^{-1}\left(\frac{d_{incre}}{d_{ROI}}\right) \tag{2}$$

As described above, the apparent angle of speckle pattern is estimated by line-to-line cross-correlation in the fast time (axial) direction on the speckle pattern of the blood flow images. However, our experiments showed that the apparent angle θ_a is not always the same as the actual blood flow angle θ. They only equal each other when the lateral component of blood flow is the same as scan velocity. The relationship between these two angles can be shown as follows:

$$\begin{cases} \theta_a < \theta & (V_{lateral} > V_{scan}) \\ \theta_a = \theta & (V_{lateral} = V_{scan}) \\ \theta_a > \theta & (V_{lateral} < V_{scan}) \end{cases} \tag{3}$$

The reason for this phenomenon is visually demonstrated in Fig. 2, depicting a single scatterer moving from left to right across an ROI in which successive A-lines are also being collected. The θ is the actual angle of blood flow, and θ_a is the apparent angle of the blood flow. When the lateral component of blood flow is slower than the scan velocity, as is the case in Fig. 2, the ultrasound beams are collecting data laterally faster than the scatterer is moving. Due to the width of the ultrasound point-spread-function (PSF), the transducer receives a reflected ultrasound echo even if the scatterer is not precisely at the focal position of each beam. The correct depth information of the scatterers is more accurately preserved due to dynamic receive focusing; however, errors will occur in the lateral positions of this scatterer. In Fig. 2, since the lateral component of blood flow velocity is less than the scan velocity, the apparent lateral position of the scatterer will be shifted to the right. As shown by the circles, these positions will be interpreted as the scatterer positions by the transducer. A similar, accumulated effect occurs with many scatterers in the ROI. As a result, the apparent angle of the speckle pattern will be greater than the actual blood flow angle. The opposite phenomenon will occur when the lateral component of blood flow velocity is greater than the scan velocity. Under this condition, the apparent angle of the speckle pattern will be less than the actual blood flow angle.

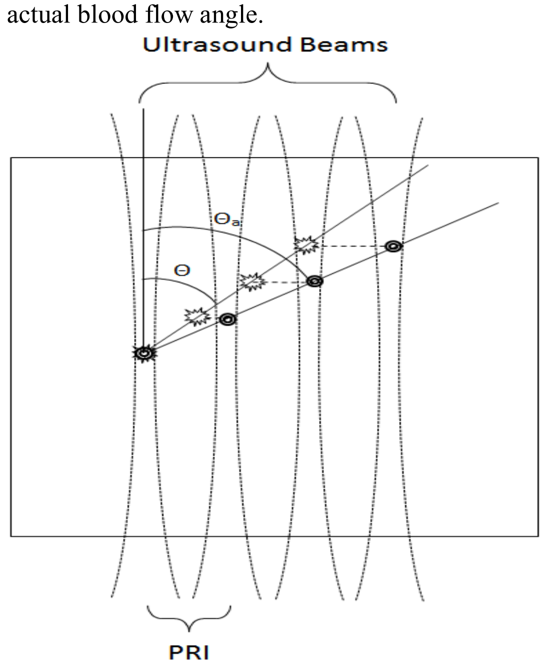

Fig. 2. Representation of the difference between apparent blood flow angle and actual blood flow angle. “Burst” makers show actual position of an individual scatterer, and round circles show where the scatterer is interpreted as existing in the space.

As discussed in our previous studies [21], the apparent lateral speckle size increases when the scan velocity is close to the lateral component of blood flow since the lateral rate of ultrasound beam movement becomes synchronized with the movement of scatterers. To compare the apparent speckle size of blood flow images collected by different scan velocities, the ROI is first aligned by the apparent angle of the speckle pattern, which means shifting each A-line by d_{ROI} to make the speckle pattern in the ROI purely lateral, as shown in Fig. 3. Then the speckle size is calculated by the full-width at half-maximum (FWHM) of the autocovariance function (ACVF) of the ROI [20]. The speckle size will change for different scan velocities. When the speckle size reaches its maximum value, the lateral component of blood flow equals the scan velocity and the apparent speckle angle equals the actual blood flow angle. Thus, in this method, two-dimensional blood flow velocity estimates are made.

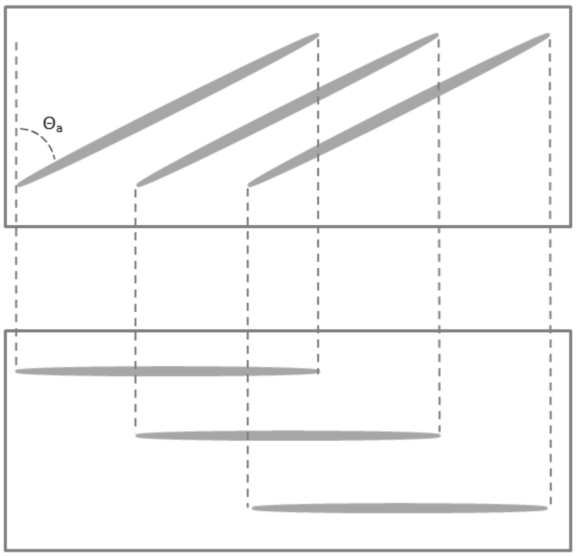

Fig. 3. Each ROI is aligned by cross-correlation, and the apparent flow angle θ_a is calculated during alignment.

III. Materials and Methods

A commercial flow phantom (Optimizer RMI 1425, Gammex, Middleton, WI), was used to simulate blood with an approximately parabolic flow profile. Blood-mimicking fluid which has acoustic properties similar to blood (speed of sound 1550 m/s, density 1.03 g/mL) was pumped through a tube (5 mm inside diameter, 1.25 mm thickness, 40^o from horizontal) in the phantom, and the tube was surrounded by tissue mimicking material (speed of sound 1540m/s, attenuation 0.5 dB/cm/MHz).

A SONOLINE Antares ultrasound imaging system (Siemens Medical Solution, Ultrasound Division, Issaquah, WA) was used for data acquisition. The VF7-3 linear array transducer (192 elements, 3.33 MHz center frequency) was set to focus on the tube located in the blood flow phantom, with a total imaging depth of 5 cm.

The Axius Direct Ultrasound Research Interface (URI) was employed to transfer ultrasound data (post-beamformation but before any down-stream processing) to a computer for further analysis. Imaging parameters can be accessed from the header information given by the URI, including frame rate, number of A-lines and beam spacing. Thus, the scan velocity can be calculated given the values of these parameters. In our experiments, nine scan velocities were used, which were 22, 26, 31, 37, 46, 52, 63, 75 and 91 cm/s. Three different flow velocities were set to the flow phantom, which were 41, 65 and 98 cm/s.

IV. Results

Fig. 4 shows the speckle size of the aligned blood flow image with different scan velocities. It can be seen that the speckle size of the aligned blood flow image is changing with different scan velocities and it reaches the maximum value when the scan velocity equals the lateral component of blood flow.

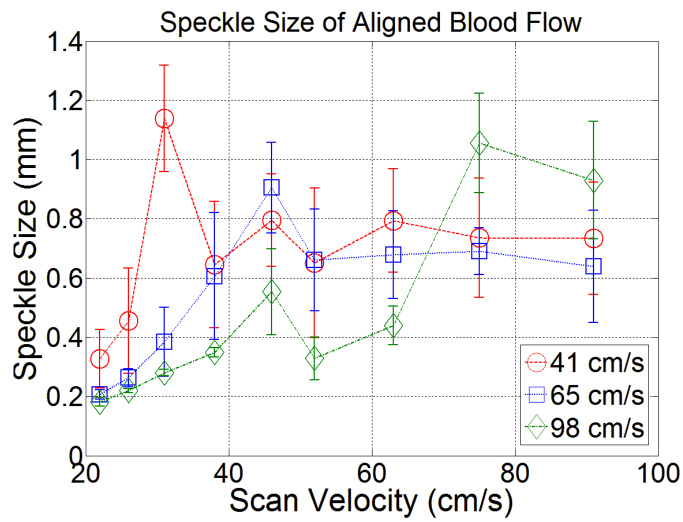

Fig. 4. The speckle size of the aligned blood flow image collected by different scan velocities.

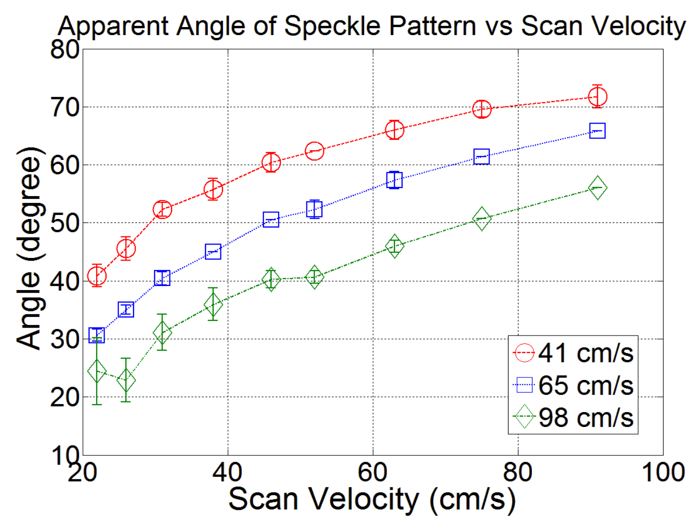

Fig. 5. Apparent angle of the speckle pattern of the blood flow data (41, 65 and 98 cm/s) collected by different scan velocities.

Fig. 5 shows the apparent angle of the speckle pattern with different scan velocities. It can be seen that the apparent angle of the speckle pattern is less than the actual blood flow angle when the scan velocity is less than the lateral component of blood flow and greater than the actual blood flow angle when the scan velocity is greater than the lateral component of blood flow. When the scan velocity equals the lateral component of blood flow, the apparent angle of the speckle pattern equals the actual angle of blood flow.

V. DISCUSSION

The speckle size of the aligned blood flow images changes with different scan velocities, which can be seen in Fig. 4. Generally, the maximum speckle size of aligned blood flow is 1.0 mm, which occurs when the scan velocity equals the lateral component of blood flow. Thus, searching for the maximum speckle size of aligned blood flow image can resolve the lateral component of blood flow velocity. In Fig. 4, the maximum speckle size of aligned blood flow image occurs when the scan velocities are 31, 52 and 75 cm/s. These scan velocities are equal to the lateral component of blood flow with an angle of 50^o relative to the ultrasound beam axis ($41 \times \sin50^o = 31$, $65 \times \sin50^o = 50$ and $98 \times \sin50^o = 75$ cm/s).

Fig. 5 shows that the apparent angle of speckle pattern increases with scan velocities. The reason is the mis-registration between the ultrasound beam scan and scatterer movement. When the scan velocities are 31, 52 and 75 cm/s, which give the maximum speckle size of aligned blood flow image, the estimated apparent angle of speckle pattern is 52.7, 51.6 and 53.8^o, which is greater than 50^o. The reason is that the flow gradient causes speckle decorrelation in the axial direction, which will reduce d_{ROI} during cross-correlation. Since the apparent angle of speckle pattern is calculated by (2), when d_{ROI} decreases, the estimated angle of blood flow will be greater than the actual angle of blood flow.

Combining the results in Fig. 4 and Fig. 5, the amplitude of the two-dimensional blood flow velocity is calculated to be 40.2, 61.8 and 96.8 cm/s for actual velocities of 41, 65 and 98 cm/s. All the estimated velocities are slightly lower than the actual blood flow velocities; a possible reason is that axial speckle decorrelation causes overestimation of blood flow angle as discussed above. Given the lateral component of blood flow velocity, when the angle of blood flow is overestimated, the blood flow velocity will be underestimated. Future studies may quantitatively investigate and compensate the underestimation to improve the accuracy of velocity estimation. Furthermore, only nine scan velocities were used in this experiment. In future studies, more scan velocities may be applied by an improved algorithm to increase the estimation accuracy.

VI. CONCLUSION

This paper investigated a new algorithm for two-dimensional blood flow velocity estimation using the apparent speckle pattern angle. The apparent angle of speckle pattern changes with different scan velocities due to mis-registration between the ultrasound beam and scatterers. Blood flow data from a flow phantom were investigated. Our results showed that the proposed algorithm can resolve the amplitude and angle of the blood flow simultaneously. Future proposed work includes using more scan velocities and algorithm optimization, including the effects of optimal ROI selection and signal-to-noise ratio.

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