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Resolving the Lateral Component of Blood Flow Velocity based on Ultrasound Speckle Size Change with Scan Direction and Speed

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***Abstract*—Conventional blood flow velocity measurement using ultrasound is capable of resolving the axial component (i.e., that aligned with the ultrasound propagation direction) of the blood flow velocity vector. However, these Doppler-based methods are incapable of detecting blood flow in the direction normal to the ultrasound beam. In addition, these methods require repeated pulse-echo interrogation at the same spatial location. In this paper, we introduce a method which estimates the lateral component of blood flow within a single image frame using the observation that the speckle pattern corresponding to the blood reflectors (typically red blood cells) stretches (i.e., is "smeared") if the blood is moving in the same direction as the electronically-controlled transducer line selection in a 2D image. The situation is analogous to the observed elongation of a subject photographed with a moving camera. Here, we develop a relationship between speckle size, scan speed, and blood flow velocity. Experiments were performed with a blood flow phantom and high-frequency transducer of a commercially available ultrasound machine. Data was captured through an interface allowing access to the raw beam formed data. Blood flow with velocities ranging from 15 to 40 cm/s were investigated in this paper. Results show that there is a linear relationship between the reciprocal of the stretch factor and blood flow velocity. Two scan speeds were used in our experiments. When the scan velocity is 64.8 cm/s, compared with the theoretical model, fitting results based on experimental data gave us a linear relationship with average flow estimation error of 1.74±1.48 cm/s. When the scan velocity is 37.4 cm/s, the average estimation error is 0.65±0.45 cm/s.**

I. Introduction

ULTRASOUND has been widely used as a diagnostic tool in the cardiovascular system. It is known that the distribution of the blood velocities within a vessel contains valuable diagnostic information. Likewise, motion of heart tissue is dependent on the health of cardiac muscle [1]. For example, stroke is a type of cardiovascular disease. It affects the arteries leading to and within the brain. A stroke occurs when a blood vessel that carries oxygen and nutrients to the brain is either blocked by a clot or bursts. It kills more than 143,000 people a year, corresponding to about 1 of every 17 deaths. It is the No. 3 cause of death behind diseases of the heart and cancer. Abnormal blood flow can be counted as one symptom of stroke [2]. Thus, accurate measurement of abnormal blood flow patterns is useful to clinicians.

Currently, most quantitative flow measurement done in commercial ultrasound occurs along the scan axis, i.e., in the direction normal to the transducer face. This is because Doppler-based instruments cannot resolve flow parallel to the transducer face. If a method were devised that measured flow parallel to the transducer face, then the two could be combined to resolve the two-dimensional velocity vector in the scan plane, providing better clinical information to a physician.

In this paper, we develop a previously-proposed relationship between the speckle size and scan speed and direction in a blood flow phantom with different velocities. In our experiments, the flow velocity is parallel to the transducer surface.

II. Background

Several methods have been used to develop ultrasound motion estimators. Conventional methods (available on most commercial ultrasound machines) operate in one dimension (1-D) and estimate the velocity vector projection along the axial dimension of the ultrasound beam. These fall into two main classes. The first are those that derive from the autocorrelation estimator [3], meant to quickly estimate the mean flow velocity over a larger spatial field of view, and now commonly referred to as "color flow." The second are those that display a spectral plot of the (temporal- and wall-filtered) flow signal [4], meant for visualizing a velocity distribution at a single (resolution-limited) small region of interest, now commonly referred to as "spectral Doppler."

However, Doppler is not able to measure the velocity vector projection along the lateral dimension of the ultrasound beam, since there will be no Doppler frequency shift when the transducer is aligned parallel to the blood flow. Some researchers have formed alternate estimation algorithms to solve this problem. For example, estimating the transit time across the ultrasound beam was proposed for measuring flow parallel to the transducer face. One method described by Newhouse and Reid measures the variance of the Doppler signals returned from lateral flow [5]. The spatial quadrature technique was proposed to estimate lateral motion by employing a modulation in the acoustical field in the lateral direction [6; 7]. However, both these methods use no information from multiple ultrasound beam positions or scanning, and therefore differ from the methods discussed in this paper. Direction and magnitude of local blood speckle

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pattern displacement using consecutive B-mode images were measured by Trahey et al, to predict lateral flow [8]. This method requires multiple images and measures speckle position changes, unlike the method in this paper which relies on only one image and estimates speckle size.

In 2001, a patent which one of the present authors (GRB) co-authored suggested a technique of blood flow measurement which takes into account the observed stretching of the speckle pattern when viewed on a scanner whose line order was in the same direction as the moving blood. The patent suggested a transform could be developed by comparing speckle size under conditions of no blood flow movement, with-scan movement, and against-scan movement. Such a transform was not developed in the patent. It is the purpose of this paper to start the development of this transform.

The second-order statistics of speckle in ultrasound B-scans was investigated in [10]. In that paper, a speckle "size" definition was proposed based on the autocovariance function (ACVF) of the speckle. In this paper, we use a similar metric for speckle size: namely the full-width-half-maximum (FWHM) of the ACVF of a region-of-interest (ROI) in the US B-mode image, shown in Fig. 1. It can be estimated by the equation:

$$ACVF_{XX}(lag) = \frac{\sum_{lag=0}^{width}((X(:,1:(width-lag)) - \overline{X}).\times(X(:,(lag+1):width) - \overline{X}))}{N} \quad (1)$$

where X is the matrix of an ROI, $\overline{X}$ is the mean values of X, *lag* is the position shift with ranges from 0 to the width of the ROI, and N is the number of pixels included in the sum operation.

In general, a preferred algorithm would measure the lateral size of speckle corresponding to relative movement of matter in a subject under study, such as blood flow or tissue motion. Depending on the scan sequence direction, speckle corresponding to moving targets or matter will either expand or contract in the direction of motion [9]. Comparing the

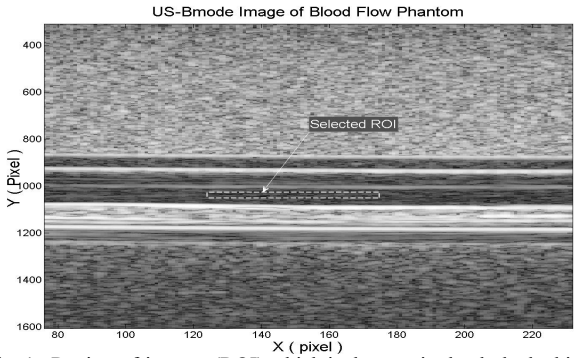


Fig. 1. Region-of-interest (ROI) which is the area in the dashed white line is selected from every US B-mode image of the blood flow phantom. It has a width of 50 pixels and height of 20 pixels, corresponding to 0.617 mm × 0.2468 mm.

expanded/compressed size relative to speckle corresponding to stationary targets potentially allows quantitative lateral flow velocity measurement.

In this paper we call the "scan velocity" the spatial rate at which individual ultrasound A-lines are collected laterally across the transducer. When the scan velocity is greater than the blood flow velocity, the scanning geometry is shown in Fig. 2. The blood flow is parallel to the surface of the transducer. D_2 is the distance between the surface of the transducer and the tube, which is 2 cm. Both scanning and blood flow have the same direction represented by solid arrows. D_1 is the distance traveled by blood flow in one pulse period, which can be calculated as:

$$D_1 = \frac{V_f}{PRF} \quad (2)$$

where V_f is the velocity of blood flow and PRF is the pulse repetition frequency. V_s is the scan velocity, then a stretch factor *SF* for different blood flow velocities can be derived as:

$$SF = \frac{\Delta L}{\Delta L - D_1} = \frac{1}{1 - \frac{V_f}{V_s}} \quad (3)$$

This stretch factor should be the factor the ACVF changes (compresses or expands) due to the relationship between scan rate and blood flow velocity. If we denote the speckle size of stationary blood flow as FWHM-$ACVF_0$, then *SF* of different blood flow velocities can also be represented as:

$$SF = \frac{FWHM - ACVF_v}{FWHM - ACVF_0} \quad (4)$$

Thus, a relationship between the blood flow velocity and speckle size can be derived when we combine (3) and (4) together, which gives us:

$$\frac{FWHM - ACVF_v}{FWHM - ACVF_0} = \frac{1}{1 - \frac{V_f}{V_s}} \quad (5)$$

Furthermore, a linear relationship will exist between the blood flow velocity and the ratio of speckle size between stationary and dynamic blood flow when we take the inverse of both sides of (5), which is:

$$\frac{FWHM - ACVF_0}{FWHM - ACVF_v} = 1 - \frac{V_f}{V_s} \quad (6)$$

When the scan velocity is less than the blood flow velocity, the relationship between speckle size and blood flow velocity can be derived in a similar way. The result is:

$$\frac{FWHM - ACVF_0}{FWHM - ACVF_v} = \frac{V_f}{V_s} - 1 \quad (7)$$

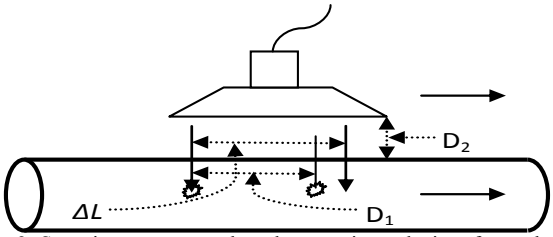


Fig. 2. Scanning geometry when the scanning velocity of transducer above is greater than the blood flow velocity in the tube below.

III. Materials and Methods

The experimental setup is briefly described here. The scanning geometry is shown in Fig. 2. A commercial flow phantom (Optimizer RMI 1425, Gammex, Middleton, WI), was used to simulate blood flow. This phantom contains a tube (5 mm inside diameter, 1.25 mm thickness) through which blood-mimicking fluid is pumped. The fluid has acoustic properties similar to blood (speed of sound 1550 m/s, density 1.03 g/cc). The tube is surrounded by tissue-mimicking material (Speed of sound 1540 m/s, attenuation 0.5 dB/cm/MHz).

The SNR in the experimental setup was 15.0 dB. The SNR was measured in the following manner. One thousand pulse-echo signals were acquired with the flow phantom velocity set to zero. The average of the 1000 signals was calculated to estimate the mean signal. Then for each signal, a noise signal was produced by subtracting the mean signal from the raw signal. The SNR was calculated by dividing the standard deviation of the mean signal by the standard deviation of the noise signal.

The V13-5 transducer (192 elements, 9.5 MHz center frequency) of a SONOLINE Antares Ultrasound Imaging System (Siemens Medical Solutions, Ultrasound Division, Issaquah, WA) was used for data acquisition. The tube is parallel to the surface of transducer, so only lateral flow data were collected. The Axius Direct Ultrasound Research Interface (URI) was employed to transfer ultrasound raw data (post-beamformation but before any processing) to a computer for further analysis.

The "Carotid" exam mode was used to scan the flow phantom. The focal depth is 2 cm, corresponding to where the tube is located in the phantom. The total imaging depth was fixed as 5 cm (starting at 0 cm) to cover the area of the tube in the phantom. In each image, 312 lines were collected. The URI includes header information to allow a researcher access to key parameters of the experimental setup. Using this header information, the frame rate was found to be 16.4 Hz. Since 312 lines were collected in each image, the PRF can then be calculated by multiplying the number of lines with the frame rate, giving 5252 Hz. Furthermore, the number of lines per centimeter was found to be 81.0086 lines/cm. The space interval ΔL between each line can then be calculated as the reciprocal of line density, which is 0.1234 mm. Thus, the "scan velocity", that is, the rate at which new ultrasound lines are formed in space, can be derived as:

$$V_s = \Delta L \times PRF \tag{8}$$

which gives V_s as 64.8 and 37.4 cm/s for the two different PRFs studied.

For each PRF, B-mode images of the flow phantom with velocity ranging from 15 to 40 cm/s were collected for study where the scan direction was the same as the blood flow. Ten images were collected for each velocity. Furthermore, 10 images of stationary blood flow were collected at the same time. In each image, a region of interest (ROI) was selected from the area of tube, with a width of 20 pixels and a length of 50 pixels, which is shown in Fig. 1. This ROI, which is represented as X in (1), was then used to calculate the mean and standard deviation of speckle size. The measured speckle size was used to estimate the blood flow velocity from (6).

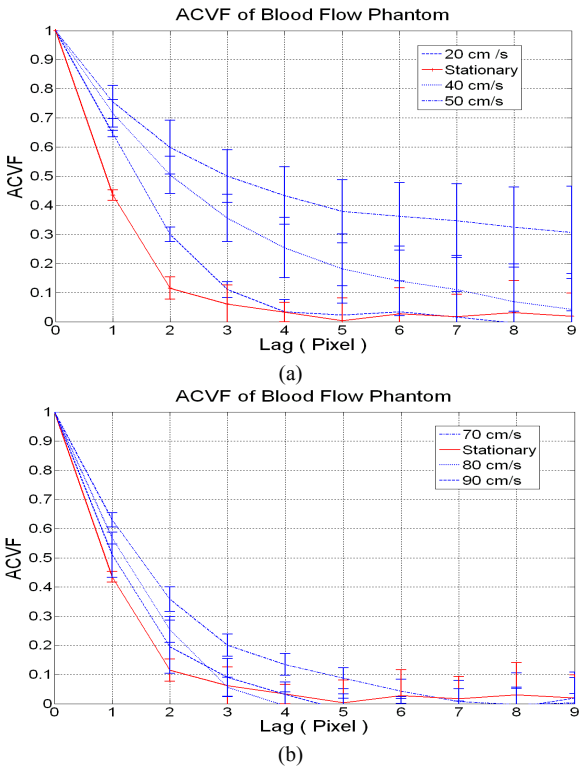


Fig. 3. ACVF of blood flow phantom (scan velocity equals 64.8 cm/s). (a) Scan velocity is greater than the blood flow velocity; (b) Scan velocity is less than the blood flow velocity.

IV. Results

The ACVF were calculated for velocities ranging from 20 to 90 cm/s. These results are shown in Fig. 3. From the graph, it can be seen that the speckle pattern of moving blood flow is stretched significantly compared with stationary flow. Using (6) and (7), the results in Fig. 3 show stretch factors of 1.6056, 2.29, 3.38, 1.67, 1.3718 and 1.1695 for blood flow with 20, 40, 50, 70, 80 and 90 cm/s.

Furthermore, the stretch factor decreases as the absolute difference between blood flow velocity and scan velocity increases. This relationship can be explained by (5).

To investigate the linear relationship between the reciprocal of stretch factor and blood flow velocity indicated by (6), the values of the reciprocal of the stretch factor and blood flow velocity are plotted in Fig. 4.

It can be seen that for part of the theoretical curve, a linear relationship similar to (6) exists between stretch factor inverse and velocity. When the scan velocity is 64.8 cm/s, a straight-fit line has slope as -0.0174 and intercept as 1.046, which are both close to theoretical values of -0.0154 and 1, respectively. When the scan velocity is 37.4 cm/s, the straight-fit line has slope as -0.0281 and intercept as 1.001, which are close to theoretical values of -0.0261 and 1, respectively. When the scan velocity is 64.8 cm/s, compared

with the theoretical model, fitting results based on experimental data gave us a linear relationship with average blood flow estimation error of 1.74±1.48 cm/s. When the scan velocity is 37.4 cm/s, the average estimation error is 0.65±0.45 cm/s.

V. Discussion

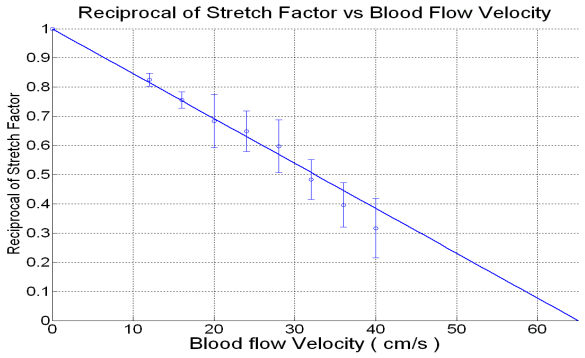
Results in Fig. 4 show that a linear relationship, as indicated by (6), exists between blood flow velocity and speckle size over part of the theoretical curve. In these experiments, the scan velocity was greater than the blood flow velocity. Experiments where the blood flow velocity is greater than the scan velocity are indicated for future work. The relationship is very close to the theoretical one, where the scatter of reciprocal of stretch factor distributed around the theoretical line within around one standard deviation..

As the blood flow velocity approaches the scan velocity, the stretch factor deviates from the theoretical line (data not shown). We believe the reason for this is decorrelation of the scatters. Theoretically, when the blood velocity equals the scan velocity, the same acoustic information will be obtained for every ultrasound A-line, making the stretch factor infinite. In fact, decorrelation of the scatters preventing this from happening. This would cause an upward (positive) bias on the reciprocal stretch factor measurements as the blood velocity approaches the scan velocity. As a result, near this transition point (6) and (7) are not suitable to describe the relationship. Further study will be carried on to investigate the scanning under this condition.

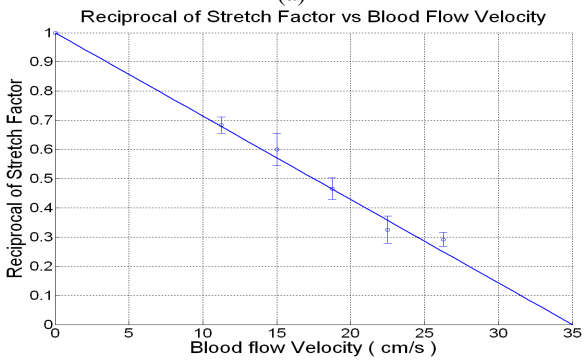
Furthermore, only two PRFs (corresponding to two scan speeds) were used in this paper. Further study can also include blood flow velocities that exceed the scan velocity, thereby testing equation (7).

VI. Conclusion

This paper investigated the relationship between speckle size and blood flow velocity. A linear relationship between the reciprocal of stretch factor and blood flow velocity was derived. Two scan velocities were examined, and compared with the theoretical model, fitting results based on experimental data gave us a linear relationship with low error. Further study should include the extension of this analysis for a wider range of blood flow velocities and PRF. Investigation for a more general relationship between speckle size and blood flow velocity, including scatterer decorrelation (when the scan velocity and blood flow velocity are close to each other) should also be included in future study.



(a)



(b)

Fig. 4. Scatter plot of reciprocal of stretch factor with standard deviation. Solid lines are theoretical lines represented by (6). (a) Scan velocity is 64.8 cm/s, (b) Scan velocity is 37.4 cm/s.

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