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Tiantian Xu

University of Nebraska-Lincoln

Gregory R. Bashford

University of Nebraska-Lincoln, gbashford2@unl.edu

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Further Progress on Lateral Flow Estimation Using Speckle Size Variation with Scan Direction

Tiantian Xu¹ and Gregory R. Bashford¹, *Senior Member, IEEE*

¹Biomedical Imaging and Biosignal Analysis Laboratory, Department of Biological Systems Engineering, University of Nebraska-Lincoln, Lincoln, NE, USA

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 report additional data on a new method recently introduced. This method 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. 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 beamformed data. Blood flow with velocities ranging from 50 to 110 cm/s were investigated in this paper. Previously, we showed results indicating a linear relationship between the reciprocal of the speckle stretch factor and blood flow velocity when the scan velocity is greater than the blood flow velocity [1]. 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. The new experiments reported here include blood flow velocities that are close to and greater than the scan velocity. Results show that the linear relationship degrades under these conditions, which we hypothesize is due to speckle decorrelation and flow gradients.

Keywords—blood flow; velocity measurement; speckle size

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 [2].

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.

Previously, we showed that there is a linear relationship between the speckle size and blood flow velocity when the scan velocity is greater than the blood flow velocity [1]. In this paper, the relationship between speckle size and scan speed and direction was investigated when the blood flow velocity is close to and greater than the scan velocity.

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 is no Doppler frequency shift when the transducer is aligned parallel to the blood flow. Some researchers have formed alternative 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 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 [9]. 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. Previously, we showed that there is a linear relationship between the velocity and speckle size when the scan velocity is greater than the blood flow velocity. In this paper, the relationship was investigated when the blood flow velocity is close to and greater than scan velocity.

Techniques in our previous paper [1] were used in this paper. Speckle size was defined as the full-width-half-maximum (FWHM) of the ACVF of a region-of-interest (ROI) in the US B-mode (detected) data. Scan velocity is the spatial rate at which individual ultrasound A-lines are collected laterally across the transducer. The same scan geometry in [1] was used and the relationship between speckle size and blood flow velocity is as follows, where the “0” subscript denotes a stationary measurement and the “v” subscript denotes a moving measurement.

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

III. MATERIALS AND METHODS

The experimental setup is equivalent to that used previously [1] and briefly described here. 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 Axis 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 preset 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 6 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 14.7 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 4595 Hz. Furthermore, the number of lines per centimeter was found to be 81.01 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:

$$V_s = \Delta L \times PRF \quad (2)$$

which gives V_s as 56.8 cm/s.

B-mode images of the flow phantom with velocity ranging from 50 to 110 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 under no

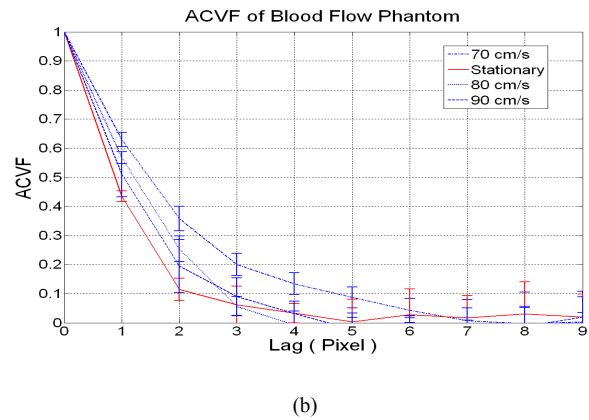
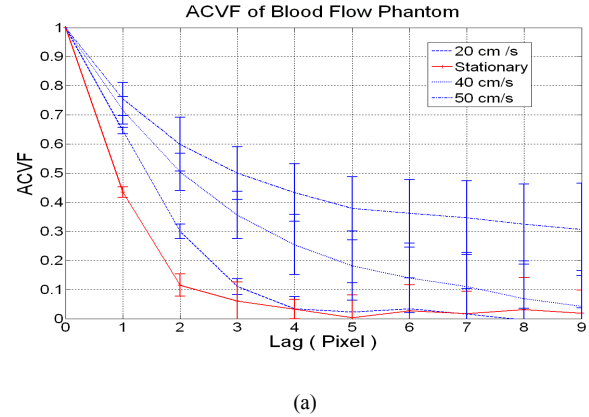
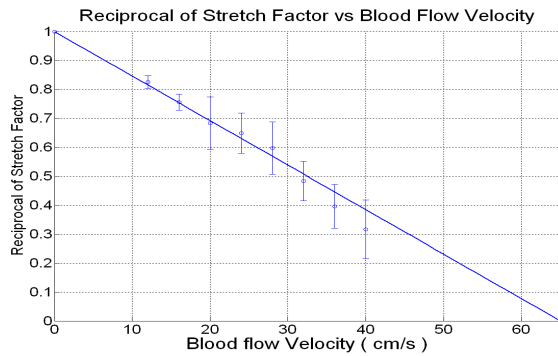


Fig. 1. ACVF of blood flow phantom. (a) Scan velocity is greater than the blood flow velocity; (b) Scan velocity is less than the blood flow velocity.

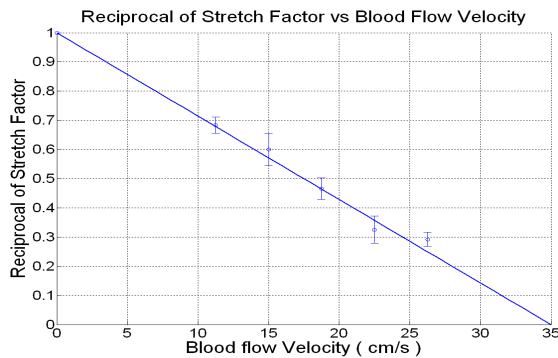
flow condition were collected at the same time. In each image, a region of interest (ROI) was selected from the area of tube and used to calculate the mean and standard deviation of speckle size. The measured speckle size was used to estimate the blood flow velocity from (1).

IV. RESULTS

As shown in our previous paper [1], it can be seen that the speckle pattern of moving blood flow is stretched significantly compared with the no-flow condition, which is shown in Fig. 1. Furthermore, the stretch factor decreases as the absolute difference between blood flow velocity and scan velocity increases. The linear relationship between the reciprocal of stretch factor and blood flow velocity when the scan velocity is greater than blood flow velocity is shown in Fig. 2. It can be seen that for part of the theoretical curve, a linear relationship exists between stretch factor inverse and velocity. 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.



(a)



(b)

Fig. 2. Scatter plot of reciprocal of stretch factor with standard deviation. Solid lines are theoretical lines represented by equation (1). (a) Scan velocity is 64.8 cm/s, (b) Scan velocity is 37.4 cm/s.

To test equation (1), the values of the reciprocal of the stretch factor and blood flow velocity are plotted in Fig. 3. In this condition, the scan velocity is 56.8 cm/s. It can be seen that when the blood flow velocity is close to and greater than the

scan velocity, a linear relationship exists between the reciprocal of stretch factor and blood flow velocity, but it has a significant upward (positive) bias compared with theoretical line.

V. DISCUSSION

Previous results in Fig. 2 show that a linear relationship 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. 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.

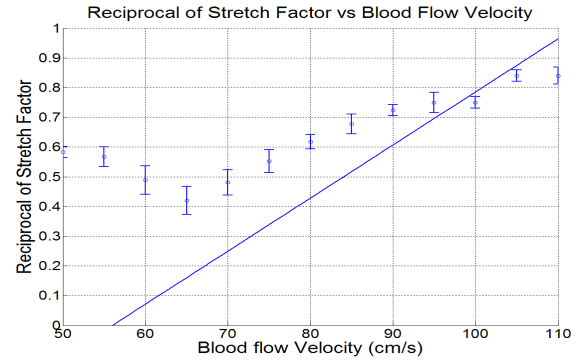


Fig. 3. Scatter plot of reciprocal of stretch factor with standard deviation. Solid lines are theoretical lines represented by (1). Scan velocity is 56.8 cm/s.

As the blood flow velocity approaches and exceeds the scan velocity, the stretch factor deviates from the theoretical line, which can be seen in Fig. 3. We believe that 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 scatterers prevents this from happening. The decorrelation causes an upward (positive) bias on the reciprocal stretch factor measurements as the blood velocity approaches the scan velocity. When the blood flow velocity is greater than the scan velocity, the observed relationship between the reciprocal of stretch factor and blood flow velocity does not fit the theoretical line well. This error could result from increased decorrelation, possibly due to larger flow shear gradients. Further studies will be conducted to investigate speckle size changes under these conditions.

VI. CONCLUSION

This paper investigated the relationship between speckle size and blood flow velocity when the blood flow velocity is close to and greater than scan velocity. Our previous study derived a linear relationship between the reciprocal of stretch factor and blood flow velocity. Compared with the theoretical model, fitting results based on experimental data gave us a linear relationship with low error when the scan velocity is greater than blood flow velocity. Decorrelation causes an upward (positive) bias error when the blood flow velocity is close to the scan velocity. A more general relationship between the speckle size and blood flow velocity including the effects of decorrelation are indicated for future study.

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