Ultrasonic Speckle Tracking with an Adaptive Frame Interval for the Measurement of Blood Flow Velocities

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Abstract. Blood flow velocity profile (BFVP) is commonly used to calculate hemodynamic parameters. Traditional speckle tracking (TST) estimates the displacements of image blocks between adjacent frame images to measure the BFVP. However, quantized errors caused by the transformation from image pixels to physical distances negatively impact on the accuracy of the BFVP measurement. In the present study, an improved speckle tracking (IST) is proposed to adaptively determine the frame intervals between two frame images based on the correlation of speckles. Firstly, the maximum normalized correlation coefficients (MNCCs) are calculated by finding the matches of the kernel blocks in subsequent comparison frame images (CFIs) after the reference frame image (RFI). Then, the largest frame interval with the MNCC greater than 0.6, is chosen to determine the velocity of the kernel block. In the experiments, the ultrasound transducer is used to scan a vessel-mimicking phantom with the peak velocities of 0.1 m/s and 0.2 m/s, respectively. The echo signals are off-line processed to yield B-mode images. The normalized root mean square errors of the TST-based results are 23.06±3.36 % and 19.92±3.08 %, and reduced to 8.36±1.86 % and 7.76±1.90 % by the IST method for the two different peak velocities. In conclusion, the IST method can improve the measurement accuracy of the BFVP, from which accurate diagnosis information could be hopefully acquired in clinics.

1. Introduction

Vascular diseases caused by atherosclerosis are affecting the health of millions of people all over the world [1]. Hemodynamic parameters based on blood flow velocity profile (BFVP) in arteries have been proven to be closely related to the progression of atherosclerosis [2]. Thus, it is important to accurately measure the BFVP in vessels.

As a non-invasive method, the pulse wave Doppler (PWD) technique has been widely used in clinic [3]. However, this method is limited by the flow angle. To circumvent the limitation, traditional speckle tracking (TST) with measuring the displacements of image blocks between adjacent frame images was proposed, and had been demonstrated to be capable to detect the complete vector velocity due to the superiority of the angle independence [4]. Instead of the entire kernel block, Chen et al. used only two diagonal lines or two bisectors in the kernel block for speckle tracking [5]. This method can significantly reduce the computation time, but the measurement accuracy of blood flow velocities is not significantly improved. In the TST method, the shortened duration between adjacent frame images makes the small displacements of speckles and the large quantized errors. The accurate
performance of the method is limited to the quantized errors caused by the transformation from the number of image pixels to physical distances. Thus, two adjacent frame images is not suitable for speckle tracking to estimate the BFVP.

Therefore, it can obtain an increasing displacement to improve the BFVP measurement that the interval (IF) between the reference frame image (RFI) and the comparison frame image (CFI) for speckle tracking is reasonably increased. In this study, the improved speckle tracking (IST) is proposed. It adaptively determines the IFs between two frame images for speckle tracking based on the correlation of speckles. It is firstly necessary to calculate the maximum normalized correlation coefficients (MNCCs) by finding the matched blocks of the kernel blocks in subsequent CFIs after the RFI. Then, the largest IF with the MNCC greater than 0.6 (the strong correlation of speckles), is determined to calculate the velocity of the kernel block. If the MNCC is always less than 0.6 (the weak correlation of speckles), the results are estimated by the IF used in the adjacent radial positions. The feasibility of the IST method is verified in the experiments based on vessel-mimicking phantom with peak velocities of 0.1 m/s and 0.2 m/s. The L14-5w/60 linear-array transducer with the SonixTouch research platform (Ultrasonix Medical Corporation, Richmond, B.C., Canada) is used, and then the echo signals are off-line processed to yield a series of B-mode images.

2. Methods

2.1. Traditional speckle tracking

In the ultrasound diagnostic system, the transducer is used to transmit ultrasound beams and then receive the echoes. RF signals obtained by beamforming those echoes are processed with the envelope detecting, dynamic compression, downsampling and interpolation to produce ultrasound B-mode images with the speckle patterns. The method by tracking speckle patterns in the different ultrasound frame images is called speckle tracking.

As shown in figure 1, the TST uses adjacent two ultrasound frame images [6]. The kernel block with the size of \(X \times Y\) is defined in the \(n\)-th frame image, and the best match to this kernel block is found according to the maximum of the normalized cross-correlation (NCC) between the kernel block and all candidate blocks in the search region of the \((n+1)\)-th frame image

\[
NCC = \frac{\sum_{x=1}^{X} \sum_{y=1}^{Y} (s(x,y) - \bar{s})(s'(x,y) - \bar{s}')} {\sqrt{\sum_{x=1}^{X} \sum_{y=1}^{Y} (s(x,y) - \bar{s})^2 \sum_{x=1}^{X} \sum_{y=1}^{Y} (s'(x,y) - \bar{s}')^2}},
\]

where \(\bar{s}\) and \(\bar{s}'\) are the mean of all pixel values in two blocks \(S\) and \(S'\), respectively. The offset of the best match related to \(S\) represents the displacement of \(S\) which is divided with the time between adjacent frame images to calculate the velocity of the kernel block.

![Cross-correlation](image-url)
2.2. Improved speckle tracking

Theoretically, for the fast velocities, speckles in the RFI produce a large displacement in the subsequent frame images, and thus the correlation between the kernel block and its matched blocks in the subsequent frame images decays quickly. Consequently, the IF referred as the number of the frames between the RFI and CFI, with the strong correlation is small. On the contrary, the slow velocities makes speckles produce a short displacement in the subsequent frame images. In this case, the correlation between the kernel block and its matched blocks in the subsequent frame images decayed slowly. Thus, the IF with the same strong correlation is large.

To realize this idea, an algorithm to adaptively determined the IF for speckle tracking is designed. This algorithm includes two portions. The first portion is to calculate the MNCCs between kernel blocks in the RFI and their matched blocks in the subsequent CFIs. The matrix variables \( C(k, l) \) and \( D(k, l) \) denote MNCCs, and the matched displacements (MDs) between the kernel blocks in the RFI and their matched blocks in the subsequent CFIs, respectively, where \( 1 \leq k \leq K \) is the radial index of kernel blocks, and \( K \) is the maximum number of kernel blocks located along the radial direction of the lumen; \( 1 \leq l \leq L \) is the index of the CFIs, and \( L \) is the maximum number of the CFIs. The sizes of kernel block and search region are \( X \times Y \) and \( M \times N \) pixels, respectively. \( MNCC_{(k,l)}(m_b, n_b) \) is defined as the normalized correlation coefficient between the \( k \)-th kernel block \( S_k \) in the RFI and its matched block \( S'_l \) with the pixel indexes of \((m_b, n_b)\) in the search region of the \( l \)-th CFI. Finally, the matrixes of MNCCs and MDs are obtained by:

\[
C(k, l) = MNCC_{(k,l)}(m_b, n_b) \tag{2}
\]

\[
D(k, l) = \sqrt{m_b^2 + n_b^2} \tag{3}
\]

Figure 2. Flow chart to adaptively determine the IF in the IST method
In the second portion, as shown in figure 2, the IF is adaptively selected according to the calculated $C(k,l)$ based on two criterions: (i) the most IFs with the MNCC greater than 0.6 are selected to calculate the displacement; or (ii) when all MNCCs are less than 0.6, the IFs are replaced by the valid results at the adjacent radial positions. For the $k$-th kernel block, as the CFI $Op_f(k)$ is found by the adaptively selected the IF, the displacement of this kernel block is obtained by:

$$Op_d(k) = D(k, Op_f(k))$$

(4)

Finally, the velocities of $K$ kernel blocks are estimated by their MDs divided by the durations of the IFs adaptively selected.

3. Experiments

To demonstrate the feasibility of the IST method, experiments based on a vessel-mimicking phantom are performed as shown in figure 3. In the schematic diagram for the setup of the vessel-mimicking phantom shown in figure 3(a), the vessel-mimicking made of a silicone tube with a radius of 5 mm traverses an acrylic box filled with cooling and solidified 2 % agar gel served as the tissue-mimicking. The blood-mimicking fluid made of 6 % aqueous starch solution (6 g starch per 100 ml distilled water) flows from the inlet to the outlet of the vessel-mimicking. The transducer is positioned on the surface of the tissue-mimicking above the tube parallely. To minimize acoustic artifacts caused by the reflection, acoustic absorbers are placed at the bottom of the acrylic box. According to the schematic diagram, the physical experimental configuration for the data acquisition is built, and shown as figure 3(b). The blood-mimicking is placed in a flume, and circularly flows in the silicone tube to continuously pass the vessel-mimicking phantom. The impetus is provided by a constant current pump (Shanghai HuXi Analysis Instrument Factory Co., Ltd.).

![Figure 3](image_url)

**Figure 3.** The experiments: (a) Schematic diagram for the experimental setup and (b) photo of the experimental setup

In the experiments, the IST method is evaluated with the two different peak velocities of 0.1 m/s and 0.2 m/s, respectively. The DPP versions are calculated by

$$v(r) = v_{\text{max}} \left(1 - \frac{r^2}{R^2}\right),$$

(5)

where $v_{\text{max}}$ is the peak velocity at the centerline of the vessel-mimicking detected by the PWD, $R = 5$ mm is a radius of the vessel-mimicking, and $v(r)$ is the blood-mimicking flow velocity at the radial position $r$. Experimental parameters are listed in Table 1. Then, the echo signals is delayed and summed to carry out beamforming. With the envelope detection, dynamic compression, downsampling and interpolation, ultrasound image sequences are produced with the frame rate of 400 Hz.

In the RFI, $K = 30$ kernel blocks with the size of $X \times Y = 10 \times 10$ pixels are used to divide the lumen along the radial direction to compute the velocity profile. Total $L = 9$ frame images after the RFI are used as the CFIs. In the CFIs, search regions with the size of $M \times N = 20 \times 150$ pixels are
set to find the matched blocks. The velocities of the 30 kernel blocks are estimated by their optimal MDs divided by durations of the most IF. Moreover, the BFVPs are also estimated with the CST method, and compared with those based on the IST method.

| Type            | Parameter name | Value       |
|-----------------|----------------|-------------|
| Centre frequency | 10 MHz         |             |
| Depth           | 40 mm          |             |
| Width           | 60 mm          |             |
| Frame rate      | 200 Hz         |             |
| Gain            | 60%            |             |
| Dynamic range   | 70 B           |             |

### 4. Results and Discussions

To evaluate variations in the correlation of the kernel blocks with the increasing IF, figure 4 shows the MNCC curves plotted as the function of the IF between the RFI and the CFIs, where different colors denote different distances away from the wall. Each subplot shows the results for a given peak velocity. MNCCs decrease with the increasing IF, which indicates that the correlations between the kernel blocks and their matched blocks reduce with the increasing duration between the RFI and CFI. However, the reduced degrees are different for various peak velocities and distances away from the tube wall. For the two peak velocities, the MNCCs gently decrease with the increasing IF for a slow peak velocity. By contrast, they descend sharply for a fast peak velocity. For the different distances away from the wall, MNCCs descend to different degrees with the increasing IF. The correlation of moving speckles exhibits the maximum decrement at the centreline of the vessel, but the highest stability close to the wall.

![Figure 4. Curves of MNCC plotted as the function of the IF, for blood-mimicking flow with peak flow velocities of 0.1 m/s (a) and 0.2 m/s (b).](image)

![Figure 5. Curves of MD plotted as the function of the IF, for blood-mimicking flow with peak flow velocities of 0.1 m/s (a) and 0.2 m/s (b).](image)

In order to explain variations in the MDs corresponding to the MNCCs, figure 5 shows the curves of MD plotted as the function of the IF. Each subplot shows the results for a given peak velocity. In the IST method, the MD is calculated by the equation (3). With the increasing IF (the time duration), the moving kernel blocks produce an increasing displacement. Theoretically, displacement of uniform
motion has a linear growth with an increase of the moving duration. Due to the blood-mimicking flowing with the constant velocities, the MD curves should linearly rise. As expected, all estimated MDs in figure 5 present a strong linearity when the corresponding MNCCs in figure 4 are larger than 0.6, and MNCC more than 0.6 represents the strong correlation. However, for those fast blood flow velocities, the MNCCs decay quickly as shown in figure 4(b). When the MNCCs are less than 0.6, the correlations between the kernel and matched blocks are weak due to the too large moving distances. For these cases, the estimated MD curves are no longer linearly growing with the increasing IF. As an example, the MNCC curve of $d = 5$ mm (black curve) in figure 4(b) drops below 0.6 as the IF is greater than 5. The corresponding MD of $d = 5$ mm (black curve) in figure 4(d) produce a nonlinear fluctuation.

For those MDs with linear growth, the largest IFs with the MNCCs greater than 0.6 are selected to calculate the reliable MDs. For instances, MNCCs at all seven radial positions in figure 4(a) (the peak velocity is 0.1 m/s) are greater than 0.6. Accordingly, all MD curves in figure 5(a) possess the strong linearity. In these cases, IFs of 9 are selected to compute the velocities for all kernel blocks. In figure 4(b) (the peak velocity is 0.2 m/s), for the radial position of $d = 4.2$mm and 5.0 mm, the largest IFs with the MNCC greater than 0.6 are 5 and 4, respectively. Hence, the IFs of 5 and 4 are selected to compute the velocities.

In order to evaluate the accurate performance of the IST method, the BFVPs of the blood-mimicking flow with the two peak velocities are estimated with the IST method. The results are compared with the TST-based those and the DPP versions, respectively. Figure 6 shows the IST-based BFVPs (dash-dotted lines) plotted with those based on the TST (dashed lines) and DPP (solid lines) results. Two subplots show the estimates for the peak velocities of 0.1 m/s (a) and 0.2 m/s (b), respectively. In general, the BFVPs based on the IST and TST methods are both parabolic and in agreement with the DPP versions. However, the IST-based velocities close to the walls are more consistency with the DPP versions than those based on the TST method. Moreover, the quantized errors caused by the transformation from image pixels to physical distances are relatively lager in TST-based MDs than those based on the IST method. Thus, the TST-based BFVPs possess larger random fluctuations than the IST-based results. Overall, the IST method can improve the measurement accuracy for the BFVP estimation.

5. Conclusions
In the present study, the IST method is proposed to adaptively select the frame interval based on the correlation of speckles. Adaptive frame intervals make displacements of speckles increasing, and thus the quantized errors caused by the transformation from image pixels to physical distance are reduced, particular for slow flows close to the vessel wall. In the experiments, the normalized root mean square errors of the TST-based results are $23.06\pm3.36 \%$ and $19.92\pm3.08 \%$, and reduced to $8.36\pm1.86 \%$ and
7.76±1.90 % by the IST method. It has been concluded that the IST method can provide the more accurate BFVP estimations, and could improve the diagnosis for vascular diseases in clinics.

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References
[1] PR Hoskins, PJ Fish, and Wn Mcdicken. “Developments in cardiovascular ultrasound. Part 2: Arterial applications”. Medical & Biological Engineering & Computing 1998; pp: 259-269.
[2] AM Malek, SL Alper, S Izumo. “Hemodynamic shear stress and its role in atherosclerosis”. Jama 1999,pp: 2035-2042.
[3] T Cao, L Yuan, and R Litao. “Cardiac motion signals contained in the Doppler flow tracing and their influence on the measurement of blood flow velocity”. Chinese Journal of Uitrasonography 2001.
[4] LN Bohs, BJ Geiman, and ME Anderson. “Speckle tracking for multi-dimensional flow estimation”. Ultrasonics 2000; pp: 369.
[5] H Chen, JY Lu. “A fast method for speckle tracking”. Ultrasonics Symposium. IEEE 2013, pp: 2579 - 2582.
[6] GE Trahey, JW Allison, and OT von Ramm. “Angle Independent Ultrasonic Detection of Blood Flow”. Biomedical Engineering, IEEE Transactions on 1987, pp: 965-967.