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Estimation methods for flow imaging with high frequency ultrasound

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Abstract

This article proposes to estimate slow blood flow with high frequency ultrasound imaging. The proposed technique combines 2 methods. First, a statistical method, called Speckle Flow Imaging (SFI) based on the analysis of changes in the speckle pattern along time, gives an index directly related to the total velocity vector. Secondly, a block matching approach estimates the in-plane velocity components. Results on calibrated flow sequences of blood mimicking fluid have shown good agreement with the statistical model. The quantification of flow is achieved with pulsed flow and is also angle independent when the flow is perpendicular to the ultrasound beam. Speckle Tracking has been evaluated on the same data and has shown good estimation of the in-plane velocity vector when the component of velocity perpendicular to the imaging plane is inferior to 1 mm/s. The results of these two methods permit the evaluation of the total 3D velocity field and the orthogonal velocity component relative to the imaging plane. This allows the quantification of blood flow (volumetric per time unit across the sequence).

Keywords: Blood flow; Speckle Tracking; High frequency imaging; Decorrelation; Microcirculation

1. Introduction

Extraction of quantitative information from ultrasonic images is generally hard to achieve due to low signal to noise ratio. The ultrasound image formation models and the statistical properties of the speckle have to be taken into account in order to estimate the motion of blood from ultrasonic sequences. This work presents a method for estimating the velocity field of blood flow in microcirculation from high frequency ultrasound imaging.

Recent biological studies and research on animal models such as mice have shown the necessity to quantify vascular flow dynamics in microcirculation of healthy and pathological tissues. The analysis of tissue vascularization permits the tracking of the evolution of tumour cells for example. It has been shown that high frequency ultrasound imaging is a suitable modality for the visualisation of the microcirculation [1]. Technological advances in high frequency ultrasound imaging and methods such as the one proposed in this paper could contribute to improve biological mechanism understanding [2].

For cardiovascular applications, Doppler methods are usually considered to be the standard. However, for slow blood flow quantification in vessels with diameter less than 1 mm, such methods are limited. Also, with Doppler methods the angle between the orientation of flow and the probe must be known, the spatial resolution is limited and a specific acquisition mode must be chosen. All these points are limitations that make Doppler methods unsuitable for the present situation.

This work presents a new real-time non-Doppler method based on the spatiotemporal analysis of the speckle pattern from high frequency ultrasound image sequences. The next section presents the two complementary methods: SFI and Speckle Tracking. We show how volumetric flow is calculated from the results obtained simultaneously by the two methods. Section 3 presents results obtained on experimental data acquired with calibrated flow. This method does not encounter the problems of Doppler method and does not require any specific settings of the imaging system.

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2. Materials and methods

To estimate the 3D blood flow velocity components, two complementary approaches are developed. The Speckle Flow Index (SFI) algorithm estimates the total velocity for each pixel of the sequence and a block matching approach (2D Speckle Tracking) estimates the 2D velocity components in the imaging plane (apparent velocity).

2.1. Speckle Flow Index (SFI) algorithm

The method is based on a spatiotemporal analysis of the changes of the speckle pattern. The statistical properties of the speckle have been widely studied [3]. It has also been shown that the amount of change in the morphology of the speckle along time is directly related to the second order statistics. This approach has been used by Lupotti [4] to measure speckle decorrelation for transverse blood flow estimation with an intravascular catheter.

In this work, the estimation is based on the statistical models of the speckle of biological tissues in movement proposed by Wear et al. [5] and applied by Rubin et al. [6]. This model stipulates that the decorrelation law during time of the speckle depends on the flow velocity, on the point spread function of the imaging system (PSF) and on the framerate.

The normalized auto-covariance C of a pixel across time is modelled by a Gaussian function:

$$C \propto \exp\left[\frac{-\Delta t^2}{2}\left|\frac{\vec{V}}{\vec{\sigma}(y)}\right|^2\right] \tag{1}$$

where Δt is the framerate, $\vec{V}$ is the fluid velocity, $\vec{\sigma}(y)$ is the beam correlation width of the PSF in the velocity direction (for a given flow orientation, it depends on the depth y).

The principle of the SFI method consists in the estimation of the normalized variance of one pixel across the direction z (Fig. 1) directly from image sequences.

We propose a real-time recursive fast formulation of this estimator in the form of a space–time linear filtering of the sequence of the images differences [7]. This formulation takes into account the qualities of the images (high frequency, high space resolution, low flow, high temporal sampling rate):

$$\mathrm{SFI}(x,y,n) = \sum_{s\in\Omega(x,y)} \beta_s \sum_{n=1}^{K} \alpha_k |I(x,y,n) - I(x,y,n+1)| \tag{2}$$

where n is the number of the image in the sequence, (x,y) are the coordinates of the pixel, α and β are the coefficients of the spatiotemporal filter, $\Omega(x,y)$ is a space window centred on (x,y). Spatial smoothing (7×7 median filter) and temporal smoothing (exponential decay α coefficients) are performed. The intensity term of the image difference is related to the changes of morphology of the speckle and contains information on the movement according to the model of Wear and Popp [5]. This formulation also recalls the same differential techniques of estimation of movement as the approaches by optical flow [8,9].

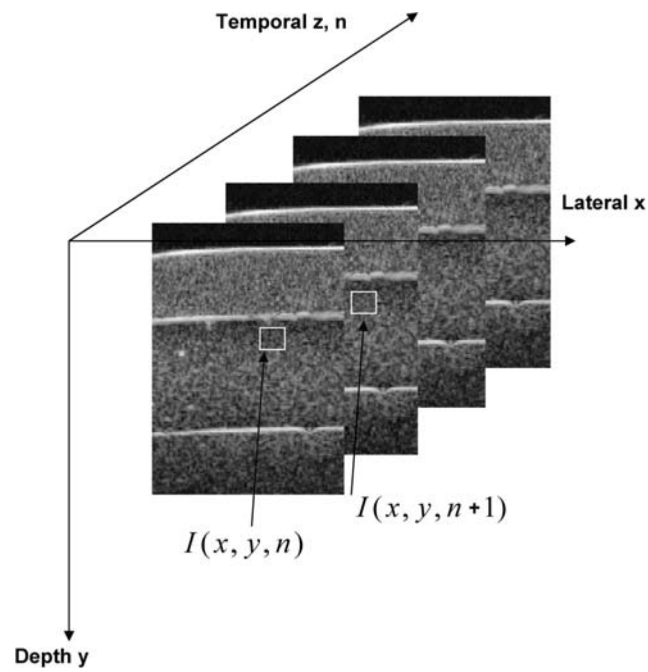

Fig. 1. B-mode image sequence. The pixel (x,y) is analyzed across temporal direction z.

2.2. Speckle Tracking

In parallel to SFI, a block matching method (2D Speckle Tracking, ST) estimates the velocity components in the imaging plane (plane (XY) in Fig. 1). The principle of ST is to find, for an initial block in an image, the best corresponding block in the following image of the sequence. The estimation is made by the maximization of a similarity criterion. This method has been applied with success for motion estimation with ultrasonic image sequence [10]. It has been shown that taking into account the statistical properties and block deformation during the movement improve the estimation [11].

2.3. Flow estimation

The in-plane velocity components are estimated with ST and the total velocity vector magnitude is estimated from SFI. The velocity component orthogonal to the imaging plane can be easily computed and so the 3D velocity vector in each pixel of the sequence. By definition, the total volumetric flow through the vessel can be computed using the Gauss's theorem by summing up the orthogonal component over the area of interest which consists on the intersection of the vessel and the imaging plane.

2.4. Materials

The method is evaluated on B-mode scans acquired on a gelatine phantom containing a vessel of 1 mm diameter. A blood mimicking fluid is injected by a syringe controlled by a linear motor. Calibrated velocities are between 0.1 mm/s and 30 mm/s. A high frequency ultrasonic system (Vevo 660, Visualsonics, Toronto) operating at 40 MHz is used to acquire the sequences of experimental images on the flow phantom. The axial resolution of the system is 40 μm and the lateral resolution is 80 μm. The framerate is 30 images per second and sequences of 300 images are acquired. During this study, the elevational angle is 0°. i.e. the flow is perpendicular to the ultrasound beam.

3. Results

3.1. SFI results

The experimental results obtained on a set of sequences of calibrated flow showed that:

- SFI gives a colour-coded mapping of the total velocity magnitude superimposed on the B-mode images (Fig. 2).
- The calibration of the estimator as a function of the calibrated velocity, carried out on a region of interest placed at the center of the vessel, is in agreement with the statistical model. Mean and standard deviation of SFI are calculated and plotted according to the calibrated velocities. The index increases in a strictly monotonous way with velocity up to 2 mm/s. A plateau is reached for larger velocities, the estimator acts like a motion detector. There is a good agreement with the model of Wear and Popp (continuous line compared to the measurement points) (see Fig. 3).
- SFI permits the evaluation of the dynamics of pulsed flow. The estimator follows the fluctuations of flow when the calibrated velocity is not constant but increases abruptly at regular intervals of time (see Fig. 4). A vertical line describes the distribution of SFI in the ROI placed at the centre of the vessel. This representation is similar to pulsed Doppler mode of conventional ultrasonic systems.
- The SFI estimator does not depend on the orientation of the vessel as predicted by Eq. (1). Mean and standard deviation of SFI are computed for different orientations of the vessel (the angle between the vessel and the imaging plane varies from 0° to 90°). Fig 5 shows that the value of SFI is constant as a function of the orientation angle.
- SFI preserves spatial resolution. For example, a flow speed profile of a laminar flow within a vessel of diameter as small as 1 mm can be estimated. Fig. 6 shows the theoretical profile corresponding to a laminar flow

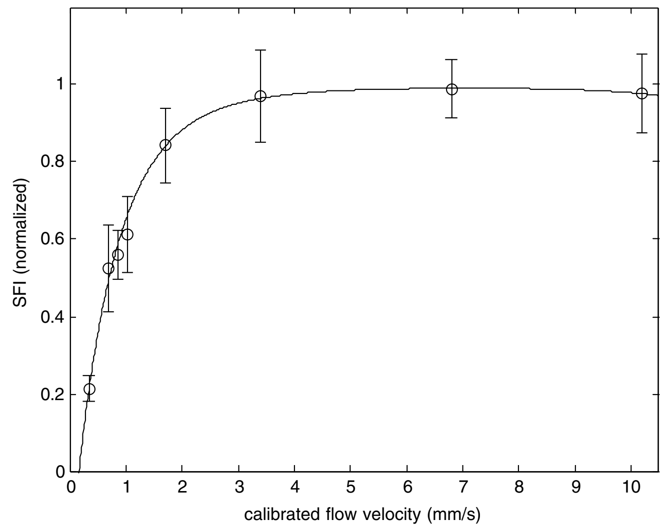

Fig. 3. SFI as a function of calibrated velocity.

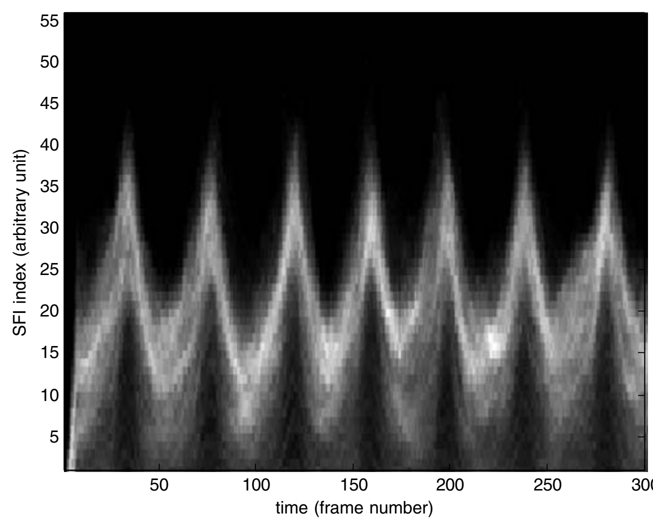

Fig. 4. Sonogram of pulsed flow. Period and magnitude of velocity variations corresponds to experimental conditions fixed on the pulsed flow.

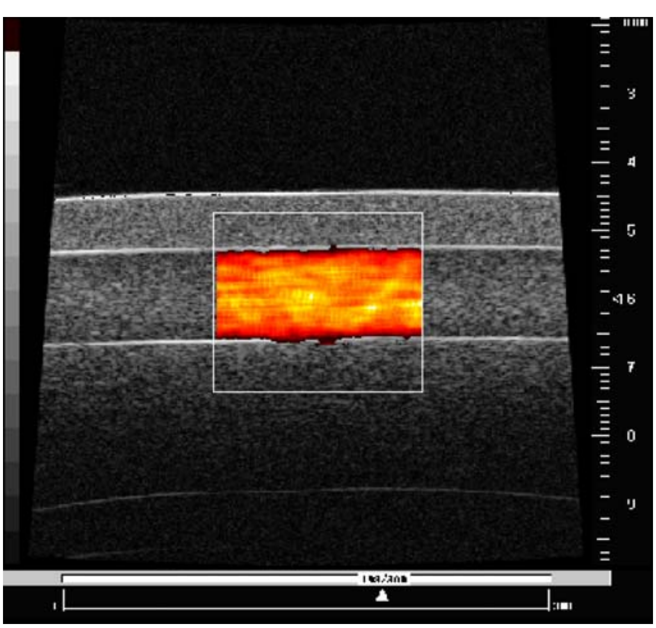

Fig. 2. A real-time dynamic cartography of the velocity estimated in the Region of Interest by SFI.

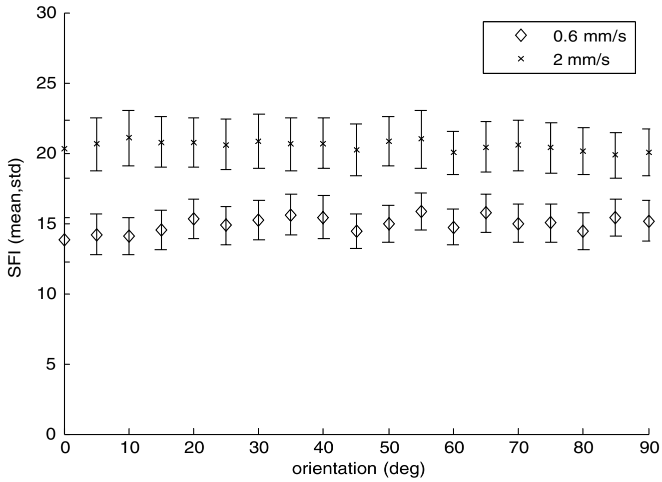

Fig. 5. SFI is sensitive to the magnitude of the velocity and independent with the orientation angle when the vessel is perpendicular to the imaging plane.

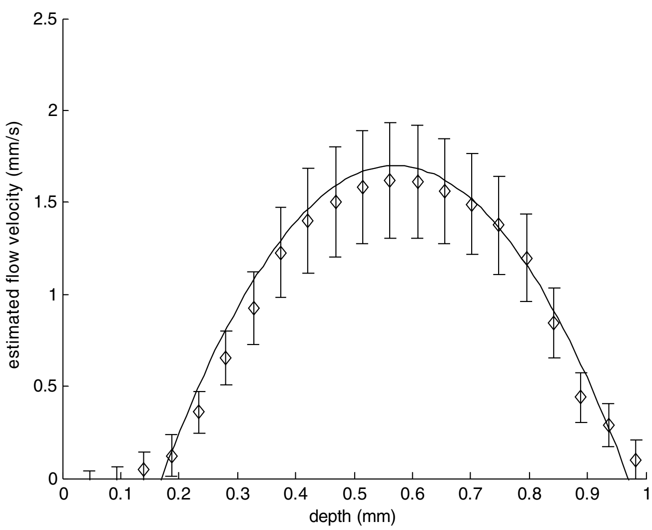

Fig. 6. Laminar parabolic flow and estimated velocities with SFI. Mean calibrated flow was 1 mm/s.

as well as the estimated velocities carried out with SFI, according to calibration curve presented in Fig. 3. As expected, the mean value at the center of the vessel is 1.7 mm/s when mean velocity is 1 mm/s.

3.2. ST results

The Speckle Tracking algorithm has been implemented on the same data described previously in order to compute the in-plane velocities.

The block matching approach used a sum of absolute difference criterion between blocks as a similarity measurement. The tracking region width was 30 pixels wide; the kernel size was 10 by 10 pixels (150 μm). These parameters have been chosen according to the 2D autocorrelation

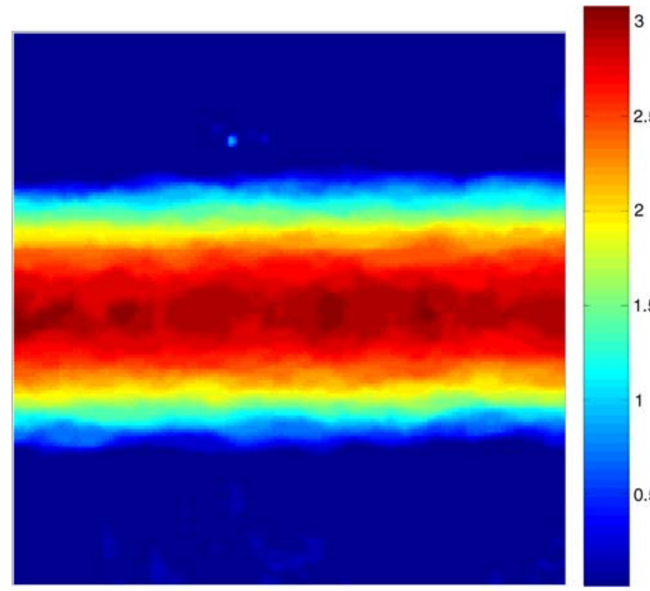

Fig. 7. Color mapping of the velocity estimated with ST. The actual size of the represented area is 2.65 mm × 2.65 mm.

function of B-mode images and the calibrated velocities to estimate.

Fig. 7 shows velocity mapping for a mean calibrated flow velocity of 1.6 mm/s with flow direction parallel to the transducer displacement. The velocity map has been smoothed by a 3 by 3 average filter for display in Fig. 7. Orientation and magnitude of the velocity are correctly estimated (Fig. 8).

For different values of calibrated velocities from 0.1 to 8.5 mm/s, the mean and standard deviation of the mea-

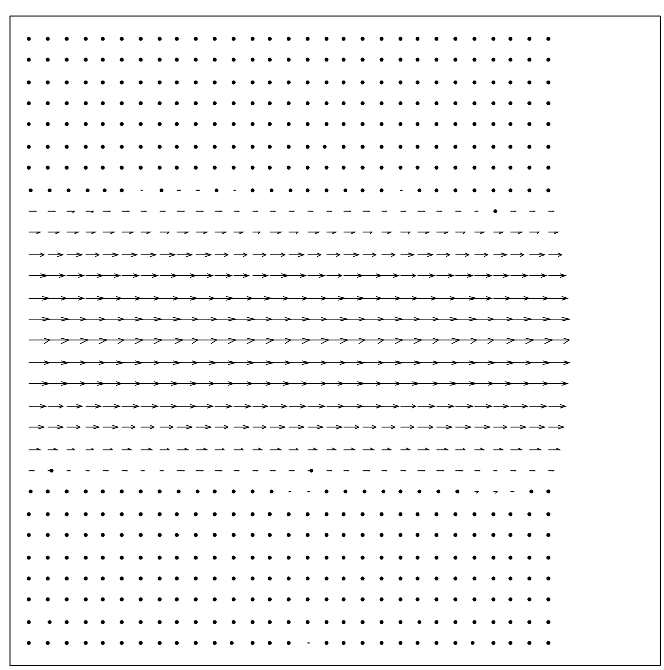

Fig. 8. Value and orientation of the displacement estimated with ST. The actual size of the represented area is 2.65 mm × 2.65 mm.

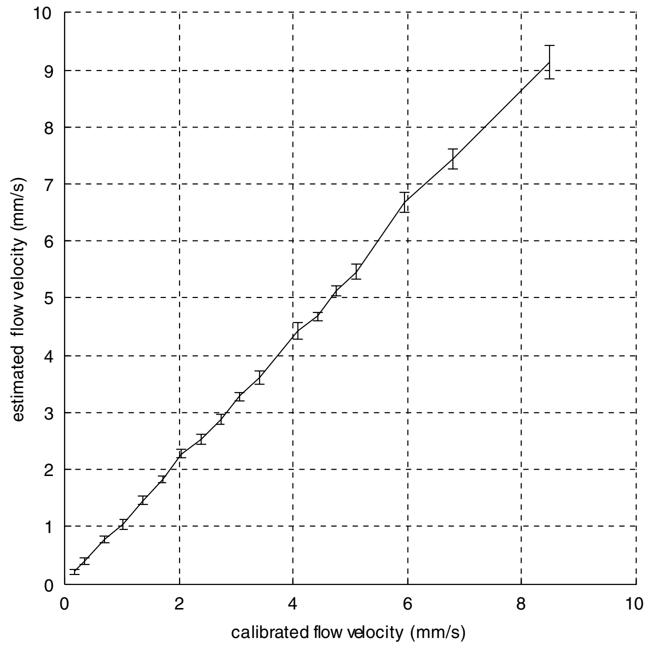

Fig. 9. Velocity estimated with ST method as a function of experimental calibrated velocity.

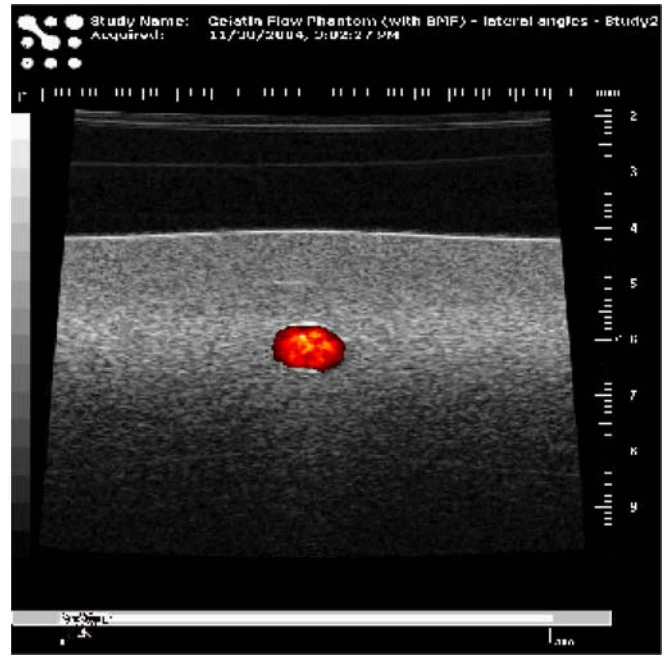

Fig. 11. Detection of the enclosed area of flow with SFI.

sured velocities are computed. ST shows good agreement between calibrated and measured velocities as shown in Fig. 9.

The Fig. 9 is obtained with sequences where the flow is parallel to the transducer displacement. For different orientations of the vessel, ST estimates the in-plane velocities. Fig. 10 shows theoretical velocity values and in-plane velocities estimated with ST when the angle between the vessel and the imaging plane varies from 0° to 90°. The mean calibrated flow velocity is 1 mm/s. Meanwhile, the decorrelation between frames increases when the orientation angle increases. Fig. 10 shows a downward trend of the velocity values as expected.

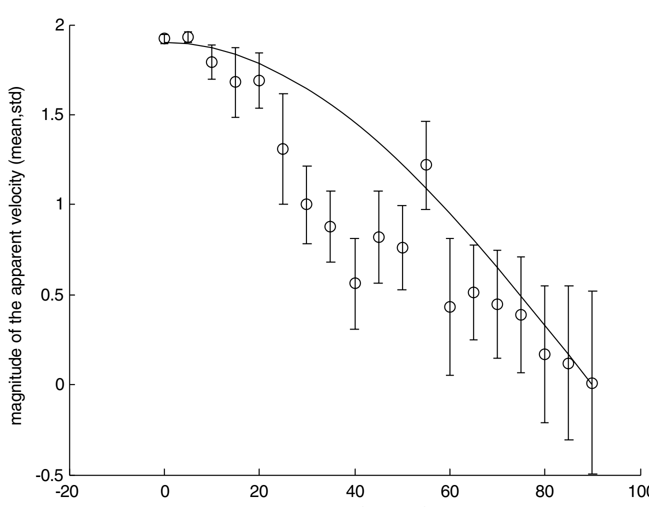

Fig. 10. Magnitude of apparent velocity. Theoretical value of the in-plane velocity (line) as a function of the orientation. The in-plane velocity is estimated with ST.

3.3. Flow quantification

Flow quantification needs the estimation of the area of the intersection between the vessel and the imaging plane. As shown in Fig. 11, the cross sectional area is estimated by thresholding the SFI values. The surface of the image containing flow is then estimated. ST is processed on the same sequence in order to get information about the magnitude and the direction of the in-plane velocities in the region of interest where the flow has been detected with SFI. Then, the third component of the velocity, orthogonal to the imaging plane, is computed from the previous results. By summing up the velocity component orthogonal to the imaging plane over the region containing flow, the total flow is calculated.

Fig. 12 presents the difference between estimated area of flow obtained by the segmentation of the sequence with SFI and theoretical elliptic area of the intersection of the vessel with the imaging plane for different orientations. For different orientations of the vessel, the total volumetric flow is processed along the sequence. Fig. 13 gives the estimation of the mean total flow for different orientations of the vessel. The value of the calibrated mean velocity is 1 mm/s, and the diameter of the vessel is 1 mm. The calibrated flow is approximately 0.8 mm^3/s. Qualitatively; Fig. 13 shows a steady estimation of the flow as a function of orientation angle. The total flow remains constant regarding the orientation of the vessel, but there is a slight over estimation of the value of flow.

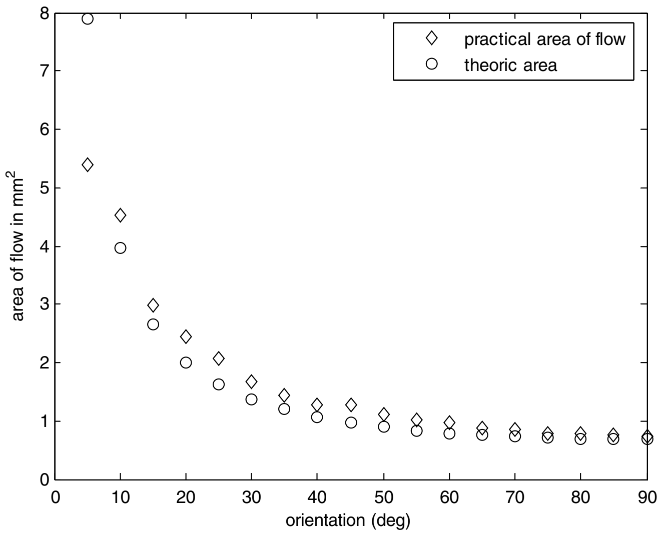

Fig. 12. Theoretical area corresponding to the intersection of the vessel and the imaging plane (○). Area estimated by SFI thresholding (◇).

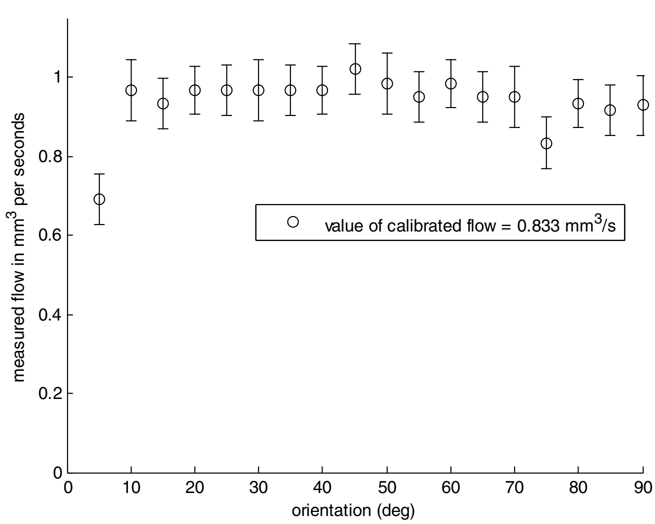

Fig. 13. Mean flow estimated for different orientation angles of the vessel. Calibrated flow is set to 0.833 mm^3/s.

4. Discussion and conclusions

It has been shown that the SFI index is an angle-independent velocity estimator for slow flow imaged with high frequency ultrasound when the vessel is perpendicular to the imaging plane. The SFI velocity index presents saturation for velocities larger than 2 mm/s and acts as a motion detector in that range. This demonstrates the feasibility of mapping slow blood flow in the microcirculation with a non-Doppler method. SFI is processed directly from the envelope image sequences and preserves high spatial resolution. Computations are simple and permit real-time implementation.

In combination with ST, the SFI method permits flow quantification with the estimation of the three components of the flow velocity. Results on flow estimation are quite promising taking into account the simplicity and the possible real-time implementation of SFI. This technique permits computation of volume flow without prior knowledge of the orientation of the vessel. Errors of estimation in the values of flow are essentially due to ST difficulty for the estimation of the magnitude of the apparent velocity when the orientation of the vessel is more than 20° for a mean calibrated velocity of flow of 1 mm/s and a framerate of 30 frames/s.

Regarding these comments, ST estimation is valid only when the component of motion perpendicular to the imaging plane is less than 1 mm/s.

This estimation is corrupted also by different rate of decorrelation in the horizontal and vertical plane. With large angles, the decorrelation increases between frames. This can be improved with larger PSF or a higher framerate to reduce the decorrelation of the speckle pattern during motion.

Moreover, SFI can be used in combination with any other in-plane velocity estimating technique such Doppler imaging, or differential methods to compute the flow.

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