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Reducing Registration Error in Cross-beam Vector Doppler Imaging with Position Sensor

Canxing Xu[1], Kirk W. Beach[2], Daniel Leotta[3], Edward Stuzman[2], and Yongmin Kim[1]
[1]Department of Bioengineering, [2]Department of Surgery, [3]Applied Physics Laboratory
University of Washington
Seattle, WA 98195

Abstract—**Various vector Doppler methods have been proposed in the last several decades to overcome the Doppler angle dependency in both conventional spectral Doppler and color Doppler by measuring both the speed and direction of blood flow. However, they have not been adopted for routine use because most of them require specialized hardware, which is not available in commercial ultrasound systems. An alternative approach (cross-beam method) that uses color Doppler images obtained from different steered beam angles is more feasible, but there is error in registering multiple color Doppler images because they are not acquired simultaneously. To alleviate this problem, we have evaluated a cross-beam vector Doppler system that registers spatially with a position sensor two color Doppler images from two different angles and temporally with ECG synchronization. The registration error was reduced to an average of 0.92 mm from 2.49 mm in 9 human subjects. Vector Doppler carotid artery images of a healthy subject and a patient with atherosclerotic plaques are also presented.**

Keywords- Doppler ultrasound; vector Doppler; atherosclerotic plaques; position sensor

I. Introduction

Doppler ultrasound has been used widely for diagnosing vascular diseases, such as atherosclerosis, by measuring the velocity of blood flow. A major limitation of spectral Doppler and color Doppler imaging modes is their dependency on Doppler examination angle in that they only measure the projection of blood velocity along the ultrasound (US) beam direction. Although the velocity can be estimated by assuming that the blood flow is parallel to the vessel wall, this assumption fails when there is non-parallel blood flow in the vessel, such as eddies and turbulence, leading to error [1].

Many researchers have tackled this limitation with vector Doppler to determine both the speed and direction of blood flow. The cross-beam method is commonly used, in which the flow is measured from multiple directions and then the vector velocity is computed accordingly [1]. Most of the cross-beam vector Doppler systems are based on multiple transducers [2] or the subaperture technique [3], by which the elements in an ultrasound transducer are divided into several groups for US beam transmission and/or echo reception. However, the arrangement of multiple transducers is cumbersome, and the subaperture technique is not generally available to the users in commercial US systems. On the other hand, an approach where blood velocity is computed by combining multiple color Doppler images (CDIs) obtained from a single transducer by steering ultrasound beams in multiple directions would be more practical [4, 5]. This method is feasible on most commercial ultrasound systems, but it requires the ultrasound probe to be kept steady to avoid misregistration between CDIs acquired with different steered angles. For vector Doppler research, the time interval between CDI acquisitions can be several minutes in some ultrasound systems because the vector Doppler imaging mode is not available to the users and as a result the acquired CDIs need to be stored for off-line processing. In such cases, it is challenging even for an experienced sonographer to hold the probe still for several minutes. One solution to this problem is to attach a position sensor onto the ultrasound probe and use the position information in registration. In this paper, we present a vector Doppler system that combines steered-beam color Doppler imaging with a position sensor.

II. Methods

A. Vector Doppler system

Our vector Doppler system consists of (1) an US scanner (Hi Vision 5500, Hitachi Medical Systems America, Twinsburg, OH) with an EUP-L53 7.5-MHz linear-array transducer and (2) a magnetic position sensor (Flock of Birds, Ascension Technology Corporation, Burlington, VT). The position sensor consists of a transmitter that is kept stationary as the reference and a receiver mounted on the ultrasound probe to measure its position and orientation in real time [6]. The position sensor transfers the position and orientation measurements to the US scanner via a serial port. In addition, a built-in ECG monitor of the US scanner was used to collect ECG information from the subject.

B. Vector Doppler image reconstruction

The cross-beam method reconstructs a vector Doppler image (VDI) by combining multiple color Doppler images (CDIs) acquired from different directions [4]. Let us assume that the flow in a vessel has a speed v and an angle α to the normal of a linear probe as shown in Fig. 1. In one color Doppler image (CDI A), the US beam direction is steered to the right side of the normal with an angle α_1, while in another CDI (CDI B) the ultrasound beam is steered to the left side with an angle α_2. Each of α, α_1 and α_2 is an angle from the normal of a linear probe (the dotted line in Fig. 1). We define counterclockwise and clockwise angles from the dotted line as positive and negative, respectively. Thus, in Fig. 1, α and α_1 are positive, and α_2 is negative. The measured velocities in CDI A (v_1) and CDI B (v_2) are given by:

$$v_1 = v \cdot \cos(\alpha - \alpha_1);\ v_2 = v \cdot \cos(\alpha - \alpha_2) \tag{1}$$

where the direction of positive flow is away from the US probe. The orientation (α) and the speed (v) can be determined by:

$$\alpha = \begin{cases} \tan^{-1}\left[\dfrac{v_2}{v_1 \sin(\alpha_2 - \alpha_1)} - \dfrac{1}{\tan(\alpha_2 - \alpha_1)}\right] + \alpha_1, & v_1 \geq 0 \\ \tan^{-1}\left[\dfrac{v_2}{v_1 \sin(\alpha_2 - \alpha_1)} - \dfrac{1}{\tan(\alpha_2 - \alpha_1)}\right] + \pi + \alpha_1, & v_1 < 0 \end{cases} \tag{2}$$

$$v = \frac{v_1}{\cos(\alpha - \alpha_1)} = \frac{v_2}{\cos(\alpha - \alpha_2)} \tag{3}$$

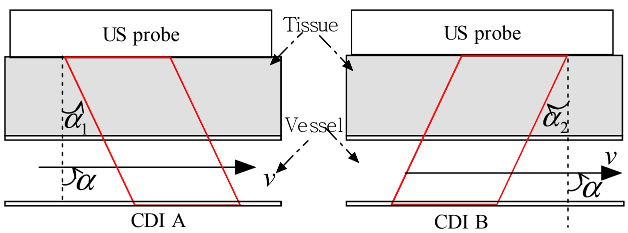

Fig. 1. Cross-beam vector Doppler based on steered color Doppler images (CDIs).

To obtain more accurate results, the beam steering angles, α_1 and α_2, are typically set to the maximum supported by the US system [5]. In Hi Vision 5500, this is 20 degrees on each side.

C. *CDI acquisition*

Data were acquired by the research interface deployed on the Hi Vision 5500 system [7]. About 6 seconds of baseband in-phase (I) and quadrature (Q) data together with position sensor information were acquired for each CDI and stored for off-line processing. Each scanline has a timestamp to record the system time when the scanline was acquired and an ECG tag to flag if the scanline was acquired at the peak of a QRS complex.

To begin data acquisition, the US beam direction was set to the first angle. After adjusting the CDI parameters, e.g., PRF and ROI size/position, scanning for 6 seconds was performed at 20~25 fps. Then, the US system was stopped, and the acquired I/Q data were moved to the hard disk. It took about one and a half minute to store the captured 6-second CDI data set at 20~25 fps. After that, the operator set the beam direction to the second angle and might also adjust the CDI parameters. Finally, another 6-second CDI data were acquired and stored. The time interval between the scanning of two CDI data sets was about 2 minutes.

D. *CDI registration*

As described above, the probe needs to be kept steady for accurate vector Doppler image reconstruction between the two successive CDI acquisitions (for about 2 minutes). Otherwise, the misregistration would occur due to unwanted probe motion. This error could be reduced with the use of a 3D position sensor.

Let us assume that the coordinate of a target is (x_A, y_A) in CDI A and (x_B, y_B) in CDI B, and the coordinate of the target in the reference coordinate system (the "Flock of Birds" transmitter) is (u, v, w). Then,

$$\begin{aligned} \begin{bmatrix} u & v & w & 1 \end{bmatrix}^T &= R_A \cdot P \cdot \begin{bmatrix} x_A & y_A & z_A = 0 & 1 \end{bmatrix}^T \\ &= R_B \cdot P \cdot \begin{bmatrix} x_B & y_B & z_B = 0 & 1 \end{bmatrix}^T \end{aligned} \tag{4}$$

where P, R_A and R_B are the 3D rigid-body transformation matrices that include both rotation and translation [8]. P, the transformation matrix from the US plane to the receiver, was determined by calibration after the receiver was mounted on the US probe [8]. R_A is the transformation matrix from the receiver to the transmitter when the CDI A was captured, while R_B is the transformation matrix for CDI B. Therefore, the relationship between two different CDIs is:

$$\begin{bmatrix} x_A & y_A & z_A & 1 \end{bmatrix}^T = C \cdot \begin{bmatrix} x_B & y_B & z_B & 1 \end{bmatrix}^T \tag{5}$$

where C is the transformation matrix from CDI B to CDI A:

$$C = \begin{bmatrix} c_{11} & c_{12} & c_{13} & c_{14} \\ c_{21} & c_{22} & c_{23} & c_{24} \\ c_{31} & c_{32} & c_{33} & c_{34} \\ 0 & 0 & 0 & 1 \end{bmatrix} = (R_A \cdot P)^{-1} \cdot R_B \cdot P \tag{6}$$

By assuming that the two CDIs are isoplanar, the third axis in the transformation defined in Eq. (5) can be dropped, thus simplifying the transformation between the two CDIs to 2D rigid-body affine warp as:

$$\begin{bmatrix} x_A \\ y_A \end{bmatrix} = \begin{bmatrix} c_{11} & c_{12} \\ c_{21} & c_{22} \end{bmatrix} \begin{bmatrix} x_B \\ y_B \end{bmatrix} + \begin{bmatrix} c_{14} \\ c_{24} \end{bmatrix} \tag{7}$$

If we allow only rotation and translation (no scaling or shearing), then c_{11}, c_{12}, c_{21}, and c_{22} are the rotational coefficients, and c_{14} and c_{24} are the translational offsets.

Finally, due to rotation between the two CDIs, the beam angle of CDI B needs to be adjusted after CDI B is registered to CDI A with Eq. (7), i.e., the angle α_2 used in Eqs. (2) and (3) should be changed to $\alpha_3 = \alpha_2 + \alpha'$, where α' is the 2D rotation angle in matrix C and is equal to $\cos^{-1}(c_{11})$.

E. *Cardiac phase synchronization*

The cardiac phase of the two color Doppler images needs to be synchronized before image registration and VDI reconstruction. The timestamp and ECG tag of each scanline contain the necessary information. A QRS scanline is the scanline that is captured at the peak of a QRS complex, while a QRS frame is a CDI frame that includes a QRS scanline. Moreover, the offset of a QRS frame is defined as the delay time of the QRS peak, which is calculated as the time difference between the first scanline and the QRS scanline in a QRS frame.

There were about 5 to 9 QRS frames in each 6-second CDI data set. Note that the frames had been acquired at a frame rate of 20~25 fps. Therefore, there could be a maximum mismatch of 40~50 ms in cardiac phase between two QRS frames. To reduce this error, the two QRS frames (from two different CDIs) with the most similar QRS offset were selected for registration and VDI reconstruction. Such two QRS frames are called matched QRS frames. In CDI A, for example, QRS frame 1 has an offset of 1 ms, and QRS frame 2 has an offset of 20 ms, while in CDI B the offset is 19 ms for QRS frame 1 and 10 ms for QRS frame 2. In this case, frame A2 (QRS frame 2 in CDI A) and frame B1 become the matched QRS frames because they have the smallest difference (1 ms) in QRS offsets among the four possible combinations (A1/B1, A1/B2, A2/B1 and A2/B2).

III. Results

A. CDI registration

We tested our vector Doppler system on the carotid artery in 9 human subjects. Some landmarks, such as carotid bifurcation or hyperechoic calcified regions, were used to evaluate the registration results. The pixel location of a landmark was determined manually in the B-mode image of each CDI. The registration error between two CDIs was defined as the pixel location difference of the selected landmark (or the mean difference if multiple landmarks were selected) in the matched QRS frames. This difference was then converted to mm using the pixel dimension information (e.g., 0.136 mm both axial and lateral directions in Hi Vision 5500 when the scan depth is 5 cm).

The results are summarized in Table I. The registration error between two original CDIs ranges from 0.85 to 7.42 mm with the mean of 2.49 mm without position sensor, while the error after registration with position sensor ranges from 0.6 to 1.5 mm with the mean of 0.92 mm.

TABLE I
Registration Error

Subject No.	*Without position sensor (mm)*	*With position sensor (mm)*
1	1.53	1.11
2	2.32	1.22
3	1.07	0.67
4	0.85	0.71
5	3.46	1.50
6	2.67	0.68
7	1.96	0.78
8	7.42	1.03
9	1.11	0.60
Mean	2.49	0.92

Figure 2 shows an example (Subject No. 6) where vector Doppler data at the carotid bifurcation with two atherosclerotic plaques were acquired. Figure 2(a) is the B-mode image of CDI with the first beam angle. The original and position-sensor-registered B-mode frames for the second beam angle are shown in Figs. 2(b) and 2(c), respectively. Five landmarks were selected to mark the shoulders and caps of plaques in Fig. 2(a), which were used in evaluating the registration. The registration error between the two original CDIs was about 20 pixels, which is equivalent to 2.67 mm. This error was reduced to less than several pixels (0.68 mm) after registration with position sensor (Fig. 2(c)).

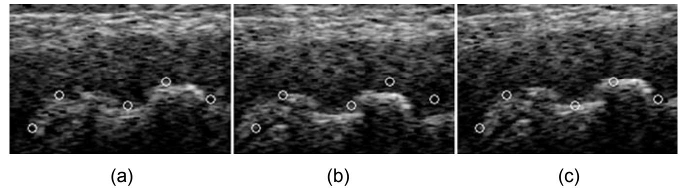

Fig. 2. Registration between different color Doppler images: (a) The B-mode image of CDI A, (b) the B-mode image of CDI B, and (c) the B-mode image of CDI B after registration with position sensor; The circles are the selected landmarks for registration.

B. VDI from a volunteer

CDIs were acquired at the common carotid artery (CCA) proximal to the carotid bifurcation and the jugular vein of a healthy male volunteer. After aligning the two CDIs, a series of VDI frames (each VDI frame consists of color representing speed and arrows indicating blood flow directions in the user-defined ROI) were reconstructed for different cardiac phases. The VDI frame with the maximal speed in the artery was considered as the VDI frame at peak systole. Figure 3 shows the VDI frame at peak systole and another VDI frame after peak systole. The flow directions in the carotid artery and jugular vein are opposite. Shortly (~100 ms) after peak systole, a flow reversal can be seen at a site close to the bifurcation (Note that discontinuous flow patterns in the jugular vein in Fig. 3(b) are most likely due to the fact that some of the blood flows might be rejected by clutter filtering because their velocities decrease after peak systole). Similar results, such as the opposite flow directions in the carotid artery and jugular vein and the flow reversals near the bifurcation after peak systole, were observed with other vector Doppler systems [9].

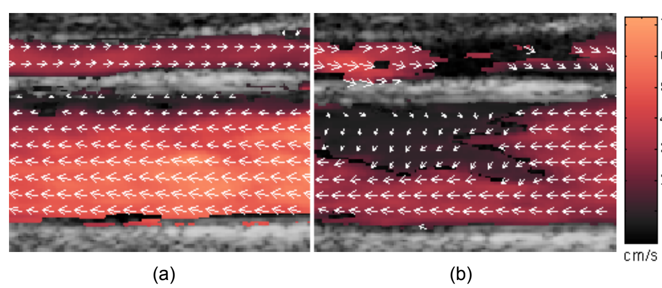

Fig. 3. VDIs in the carotid artery and jugular vein of a volunteer: (a) At the time of peak systole and (b) shortly (~100 ms) after peak systole.

C. VDI from a patient with plaques

The VDI frames were also reconstructed from the internal carotid artery (ICA) and carotid bifurcation of a patient with two small atherosclerotic plaques. Figure 4(a) shows the VDI frame during peak systole. Reversed flow is seen at the bifurcation region (enclosed by a dotted ellipse). The flow direction returns to normal after passing the bifurcation region. The reason for a flow reversal at the bifurcation region could be found from the positional relationship between carotid bifurcation and the ultrasound imaging plane, as shown in Fig. 4(b). The ultrasound imaging plane is perpendicular to the bifurcation plane. Also, Fig. 4(b) schematically shows a flow profile pattern at carotid bifurcation where the CCA branches into the ICA and the external carotid artery (ECA) [10]. Note that around the carotid sinus there is a recirculation zone [10], which could appear as the reverse flow in the ultrasound images acquired from the imaging plane shown in Fig. 4(b). These findings are qualitatively consistent with the results of computational flow dynamics simulation and MRI studies, such as flow reversals around the bifurcation and laminar flow downstream to the bifurcation [11].

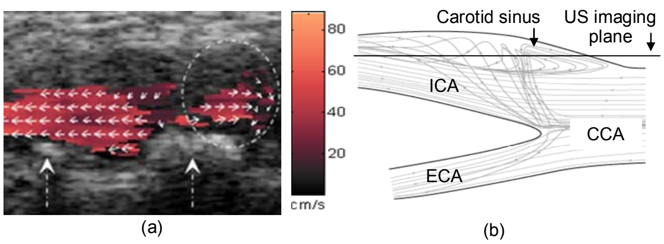

Fig. 4. VDI from a patient with plaques: (a) VDI in the carotid bifurcation with atherosclerotic plaques (pointed by the dashed arrowheads) during peak systole and (b) a schematic view of a possible flow profile in the bifurcation plane.

IV. DISSCUSIONS

An ideal cross-beam vector Doppler system should be able to acquire multiple CDIs quickly by steering the beams along the preselected directions and reconstruct the VDI frames in real time. In most commercial ultrasound systems, this feature is not currently available. Thus, the beam directions need to be selected by the operator, and the CDIs from different beam angles need to be acquired sequentially. For research, furthermore, the VDIs need to be reconstructed off-line after storing the acquired CDI data. Therefore, to reduce the registration error caused by motion, the probe needs to be held still, and the subject should not move during CDI acquisition. However, the registration error could be large when the time to acquire CDIs is long. In Hitachi Hi Vision 5500, this time was about 2 minutes, which contributed to a mean registration error of 2.49 mm (ranging from 0.85 to 7.42 mm) in 9 subjects even though the sonographer tried to keep the probe steady. By integrating a "Flock of Birds" position sensor, the mean registration error was reduced to 0.92 mm.

Note that in subjects 1, 2, 3, 4 and 9 in Table I, the registration error was from 0.85 to 2.32 mm with an average of 1.38 mm without position sensor and from 0.6 to 1.22 mm with an average of 0.86 mm with position sensor. The reduction in registration error is less than 50% with position sensor when the registration error is small (e.g., less than 1.8 mm). That is probably due to the fact that the positional accuracy of the position sensor used is 0.07 inch, which is about 1.8 mm [6]. In subjects 5, 6, 7 and 8, the error was from 1.96 to 7.42 mm (an average of 3.88 mm) without position sensor and from 0.68 to 1.50 mm (an average value of 1.00 mm) with position sensor. The error reduction rate was from 56.6% to 86.1%. This suggests that the position sensor is more effective in reducing the large registration error caused by probe motion between CDIs.

A major limitation of the proposed method is that only 2D motion is accounted for with the position sensor. This is because the two CDIs for 2D VDI reconstruction need to be isoplanar. Otherwise, they cannot be registered even though the position sensor provides the information about 3D position and orientation. The operator could adjust the scan plane according to the B-mode image appearance to satisfy the isoplanar condition, which makes the process operator-dependent.

Another disadvantage is that the motion of a subject cannot be corrected by the position sensor. An assumption in Eq. (5) is that the subject coordinates in the reference coordinate system (the "Flock of Birds" transmitter) are identical during the acquisition of CDI A and CDI B, which can be satisfied only if the subject does not move. The assumption would be violated if the subject moves during scanning, making the registration with the position sensor incorrect. Such movement could be detected from the registration error. If the error after registration is still large (e.g., more than 2 mm), it might be reasonable to infer that the subject moved. In these cases, the acquisition of CDI A and CDI B could be repeated or other registration methods, such as manual registration, should be used.

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