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Velocity Vector Imaging

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Abstract

Nowadays, ultrasound Doppler imaging is widely used in assessing cardiovascular functions in the human body. However, a major drawback of ultrasonic Doppler methods is that they can provide information on blood flow velocity along the ultrasound beam propagation direction only. Thus, the blood flow velocity is estimated differently depending on the angle between the ultrasound beam and the flow direction. In order to overcome this limitation, there have been many researches devoted to estimating both axial and lateral velocities. The purpose of this article is to survey various two-dimensional velocity estimation methods in the context of Doppler imaging. Some velocity vector estimation methods can also be applied to determine tissue motion as required in elastography. The discussion is mainly concerned with the case of estimating a two-dimensional in-plane velocity vector involving the axial and lateral directions.

Keywords: *Multiple beams, Spatial quadrature, Speckle tracking, Spectral broadening, Velocity vector*

I. Introduction

Doppler imaging is based on the principle that moving scatterers in blood cause an upward or downward shift in the frequency of a received ultrasound echo. First, we introduce various methods of estimating Doppler shift frequency. Early methods were the zero-crossing detector, phase detector, instantaneous frequency detector, etc., but their performances were not quite satisfactory [1-10].

In the phase detector, the mean Doppler shift frequency, f_{mean}, is estimated using

$$f_{mean} = \frac{PRF}{2\pi} \frac{\sum_{n=1}^{N} [i(n-1)q(n) - i(n)q(n-1)]}{\sum_{n=1}^{N} [i^2(n) - q^2(n)]}, \qquad (1)$$

where $i(n)$ and $q(n)$ denote the demodulated in-phase and quadrature components, respectively, n is the sample time index, N is the ensemble data length, and PRF stands for the pulse repetition frequency.

The instantaneous frequency detector employs the following expression:

$$f_{mean} = \frac{PRF}{2\pi N} \sum_{n=1}^{N} \left[\tan^{-1}\frac{q(n)}{i(n)} - \tan^{-1}\frac{q(n-1)}{i(n-1)} \right]. \qquad (2)$$

In 1985, we saw a breakthrough in real-time two-dimensional (2-D) Doppler imaging. Note, however, that here the word 2-D does not mean vector Doppler but refers to a 2-D region over which 2-D velocity is to be estimated. Kasai *et al.* [11] proposed an autocorrelation method for real-time estimation of Doppler shift frequency. The method is also referred to as the Kasai method, or 1-D autocorrelator because it uses only slow-time data. The algorithm for estimating the mean Doppler

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shift frequency is expressed as follows:

$$f_{mean} = \frac{PRF}{2\pi} tan^{-1} \left\{ \frac{\sum_{n=1}^{N} [i(n-1)q(n) - i(n)q(n-1)]}{\sum_{n=1}^{N} [i(n)i(n-1) + q(n)q(n-1)]} \right\}. \tag{3}$$

The Kasai method is very important in that it is a practical real-time algorithm that makes it possible to build a commercial 2-D color flow mapping system. Nowadays, most commercial ultrasound scanners use this algorithm because the implementation is simple and the performance is good. However, its main drawback is that the mean frequency estimation capability deteriorates for low signal-to-noise ratio (SNR) signals and that aliasing errors occur for high velocity flows. On the other hand, maximum entropy methods derived from autoregressive models were reported for determining the Doppler velocity from the estimated mean frequency [12-14].

Recently, improvements on the Kasai algorithm have been reported [15-24]. Wilson [15] paved the way to improvements in its performance by looking at the mean frequency estimation problem from a different perspective, i.e., in the frequency domain. Loupas *et al.* [16-18] proposed a method of using the slow-time, as well as the fast-time, data to improve Doppler frequency estimates. They presented a 2-D autocorrelator which always outperforms the Kasai's method, i.e., 1-D autocorrelator. It estimates the Doppler velocity as the slope of a 2-D Fourier spectrum of Doppler echo data passing through the origin in the 2-D Fourier domain representing the temporal frequency as abscissa and the ensemble length frequency as ordinate. They compared three Doppler velocity estimators such as the 2-D autocorrelator, the crosscorrelator, and the 1-D autocorrelator through experiments. The 2-D autocorrelator turned out to be consistently better than the 1-D autocorrelator in terms of both velocity and power estimation.

Another drawback of the autocorrelation method is that it assumes that the center frequency is a constant and thus it cannot take into the variation of center frequency due to frequency-dependent attenuation and speckle with the result that the estimation performance deteriorates at poor SNRs. Thus, the estimated mean Doppler frequency will not be accurate. In order to overcome this problem, Munk and Jensen [19] proposed the use of 2-D Fourier transform and Radon transform. Here, the 2-D refers to the pulse emission (ensemble length) direction (frequencies related to PRF) and the RF-line sample (axial, i.e., depth) direction (frequencies related to the sampling rate in the axial direction). They simulated their new approach to estimate axial blood flow velocity with both synthetic (based on a cosine waveform expression including axial sample index and pulse emission index) and simulated (generated using Field II) data. The data used were 500 realizations of 16 by 16 RF data matrix for SNRs ranging from -6 to 12 dB in steps of 3 dB. For the case of simulated data, the proposed method was better than the autocorrelation method for all noise conditions, but for the case of synthetic data, the autocorrelation method was better, except for an SNR of -3 and 0 dB. However, they did not carry out experiments. The method is related to the work of Wilson [15], and of Allam *et al.* [20,21].

The Torp group [22-24] extended the autocorrelation method so that the mean Doppler frequency as well as RF center frequency can be estimated to reduce the bias and variance of estimation. Their modified autocorrelation method was shown to achieve lower estimation bias and variance than the conventional autocorrelation method and the cross-correlation method with parabolic interpolation.

Meanwhile, Ferrara and Algazi [25-27] performed the maximum likelihood estimation of Doppler velocity by taking into consideration the effect of Doppler on both envelope and carrier. Schlaikjer and Jensen [28] combined the maximum likelihood blood velocity estimator with the properties of flow physics.

Hein and O'Brien [29] compared the performance of various Doppler estimation methods. Shariati and colleagues [30] compared the autocorrelation and crosscorrelation methods in terms of mathematics, processing step, and performance. Pinton and coworkers [31] conducted performance comparison of the normalized crosscorrelation method, Kasai's method, and Loupas' method with different parameter selections in terms of displacement errors. The Kasai's method turned out to have the fastest computation time and the worst performance. Loupas' method performed almost as well as the normalized cross correlation, except when the kernel length is small and the SNR is low, or except when the bandwidth is large.

It is important to reject clutter signal as much as possible before proceeding to Doppler mean frequency estimation. The clutter refers to a large amplitude echo reflected from stationary or slowly moving tissues. The block which is responsible for reducing clutter is known as a clutter or wall filter. Thomas and Hall [32] moved the clutter frequency to the dc frequency by complex demodulation, following its identification. Then the previous version of the complex baseband Doppler data is subtracted from the present version to yield only the Doppler frequency shift data from moving scatterers in the blood. Zheng *et al.* [33] applied the ideas of Wilson as well as Allam and Greenleaf, to remove clutter. The filtering is done in the frequency domain. Kadi and Loupas [34] assessed the performances of a regression filter and a step-initialized IIR filter. Their simulation results show that the regression filtering is generally better in clutter suppression than the IIR filtering. Jensen [35] studied the effect of including clutter filtering on velocity estimation.

A major drawback of 1-D Doppler imaging methods is that the estimated velocity depends on the angle, θ, between the insonifying beam and the blood flow direction, making the Doppler shift frequency, f_d, dependent on the beam-to-flow angle, θ, as can be seen in the following relationship:

$$f_d = \frac{2f_0 v}{c} \cos\theta, \tag{4}$$

where c is the speed of ultrasound, v is the blood flow velocity, and f_0 is the transducer center frequency. Thus, to obtain Doppler velocity estimates that are independent of the beam-to-flow angle, multidimensional methods for estimating Doppler velocity have been pursued. In this article, we confine the discussion to 2-D velocity estimation, even though some methods can be applied to three-dimensional (3-D) velocity estimation as well. In general, 2-D blood velocity estimation methods can broadly be divided into five major categories: 1) multiple beams, 2) spectral broadening, 3) speckle tracking, 4) lateral beam modulation, and 5) others. Now, we will explore and discuss each of them in detail.

II. Multiple Beam Methods

Methods that belong to this category transmit and receive multiple beams to interrogate Doppler velocities in distinct directions. The basic idea of using two or more beams is to obtain more information by interrogating a heart or a blood vessel in distinct directions. The change of beam propagation directions can be achieved either mechanically using separate individual transducers or electronically using array transducers as shown in Fig. 1. Transmitting and receiving in different directions is the same as increasing the number of equations so that

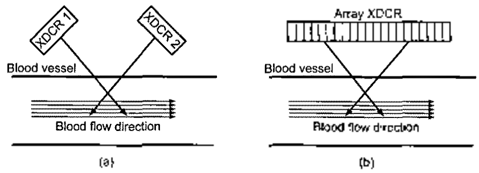

Fig. 1. Multiple beams can be transmitted and received in distinct directions using either (a) separate individual transducers or (b) a single array transducer.

the latter becomes not fewer than the number of unknowns so that the relevant system of equations can be solved for a unique solution. Needless to say, it is desirable that those additional equations be as independent as possible. This means that it is in general better to have a large angle between multiple beams. The velocity magnitude and direction are obtained by taking into account the geometry and computing the Doppler frequency shift in each direction.

The multiple beam method was first suggested by Fahrbach [36,37]. He used two transducers arranged to make a right angle to each other in order to determine the magnitude and direction of blood flow. Because the two transducers are positioned to make an angle of 90°, one measurement is given as $2f_0v\cos\theta/c$, and the other measurement is given as $2f_0(-v)\cos(\theta-90^\circ)/c=-2f_0v\sin\theta/c$. Taking the square root of the sum of the squares of the two terms, we have $2f_0v/c$, which is now independent of θ.

Steinman *et al.* [38] studied the effects of beam steering on the change of Doppler sample volume.

Daigle *et al.* [39,40] showed that by making simultaneous Doppler measurements with beams at known angles to each other, the absolute velocity could be obtained. Their paper is considered to be a seminal work that ensuing papers by Fei and Vilkomerson are based in part on. Also, using a pulsed Doppler velocity meter and a triangulation procedure, they accurately measured the beam−to−flow angle from three velocity components obtained at different beam−to−flow angles with three pulsed Doppler transducers on an esophageal probe inserted into beagle dogs.

Fox [41] showed that with the arrangement of two crossed−beam transmit transducers and a single receive transducer, a 2−D velocity could be estimated, and presented a method of determining 3−D velocity using three sets of crossed−beam transducers, each operating at different offset frequencies for spectral separation. A revolving turntable was used to provide experimental results of 2−D velocity estimation. The estimated velocity agreed well with the predicted one. Using three orthogonal Doppler transmit−receive probes that are spatially arranged, Fox and Gardiner [42] formulated equations relating three orthogonal velocity components and two Doppler shift frequency ratios, and showed through turntable and flow phantom experiments that their 3−D angle−independent theory correctly predicted velocity with an error of less than 4%.

Wang and Yao [43] used two pairs of transmit and receive transducers, with each pair used to obtain one of the two beams. The optimum probe position is where the two Doppler outputs can be canceled out exactly, so that the blood velocity at that position can be determined by adding both outputs.

Previously, a number of single−element, i.e., piston−type transducers were used to insonify a sample volume so that the beams can intersect each other. With the advancement of technology, a linear array transducer has come to be used. A full aperture or subaperture firing or receiving scheme can be used.

A study by Tamura *et al.* [44] shows that velocity vectors can be obtained from color Doppler images with a reasonably good accuracy. Real−time implementation of the vector computation method may be achieved, based on the basic scheme described by Furuhata *et al.* [45], using a single transmitter and two receivers created from a linear array transducer. A part of the linear array transducer can be used as both transmitter and receiver while another part located some distance from the first can be used as the second receiver with a Doppler angle that varies with depth. 2−D velocity vector was computed using velocity components from two directions. A color Doppler flow mapping system (Acuson model 128; Acuson, Mountain View, CA, USA) with a 5−MHz linear array transducer was used for this study. Electronic steering was used to obtain flow images from two or three directions at 20° to each other. Careful choice of the pulse repetition frequency just prevented aliasing, and enabled the best velocity resolution to be achieved. Video digitization of color Doppler flow images was used to obtain velocity values. Each video frame was stored on a hard disk, and was transferred to an IBM PS/2 model 80 with a 20-MHz processing speed. The color calibration code was analyzed in order to determine the magnitude of velocity from each pixel in a flow image.

Fei *et al.* [46] established a theoretical background, and presented a formula that can be used to obtain both the magnitude and angle of blood flow velocity using three differently oriented beams. They also introduced a method of reducing the amount of computation using a lookup table, and explained the necessary hardware configuration and computational algorithms.

Also, Fei and Fu [47] obtained three overlapping sector images by changing the transducer direction from left to center to right. The images were processed in a computer to produce *in vivo* color Doppler images of the abdominal aorta of a normal human subject. The major drawback of their method is that the transducer needs to be moved and rotated to acquire three overlapping images and that off-line processing is required.

In another paper [48], Fei *et al.* acquired three images of a flow field whose fields of view overlapped using an ultrasonic scanner with a linear array transducer, and obtained three directional velocities which in turn were used to estimate both the magnitude and direction of the blood velocity in the central region of those three overlapped images. They compared the estimated magnitude and direction to the magnitude calculated from the flowmeter and the direction measured from B-mode image with the results showing a good agreement.

In yet another paper [49], a method was also presented for visualizing 2-D flow fields, where the flow angle is represented by hue and the velocity magnitude is represented by varying the intensity and saturation. Three beams were used with a linear array transducer to obtain both the magnitude and angle of the blood flow velocity. The estimated velocity magnitude and angle agreed well with those calculated from a flowmeter and measured from B-mode images, respectively.

Maniatis *et al.* [50] used multiple beams to determine the vector velocity. For each beam direction, one color flow image was obtained. Depending on the manner in which multiple color flow images are utilized to provide both the velocity and angle at individual pixels of the entire flow field, the multiple beam approach was further divided into four different methods, which are the two-component method, simple average method, weighted average method, and least mean squares method. They compare by simulation the relative merits of the four various schemes that fall into the category of multiple beam vector Doppler velocity estimation in terms of their bias, variance, and computational complexity. The two-component method uses only two beam directions, while the rest three methods use more than two beam directions. It is found that the two-component method is considerably more accurate in magnitude estimation than the other methods and that the least mean squares method demonstrates the most significant improvement in terms of accuracy due to the increase in the number of observations.

On the other hand, Vilkomerson *et al.* [51] presented a method of producing multiple beams at known angles with a single array transducer, and formulated a method of determining both velocity and angle using two beams with either one or two insonation frequencies. Both the interelement spacing and insonation frequency were adjusted to make beams have desired angles. Also, Vilkomerson *et al.* [52] developed a diffraction-grating transducer that can be used to produce multiple beams at known angles so that angle-independent Doppler measurements can be made. Its performance was also characterized. The research team [53] also proposed changing the phase or frequency of a driving signal to produce multiple angle beams with the diffraction-grating transducer. The transducer is useful for contact scanning of blood vessel due to its limited-diffraction beam characteristics. It was also shown that the shape of Doppler spectrum could be used to accurately determine the blood flow. Using diffraction-grating transducers that are based on a configuration of four PVDF/TrFE copolymer piezo - plastic transducers that can lie flat on the skin and that inherently produce angled beams, Vilkomerson *et al.* [54] demonstrated vector Doppler operation. Three transmitters spaced 120° apart were placed around a circle with one transducer receiver placed in the center. They determined the vector velocity at the point where all four beams intersected. The absolute velocity was obtained from the square root of the sum of the squares of the velocity vector components.

Using two independent ultrasound beams simultaneously, Overbeck *et al.* [55] produced complementary Doppler signals. When both Doppler shift frequencies obtained are the same, the blood flow velocity is obtained by adding both of them. They eliminated the influence of the angle between the probe and blood flow.

Phillips *et al.* [56-58] accurately estimated 2-D motion with a phased array transducer and color flow signal processing by adaptively tracking a single transmitted pulse with more than one receive subaperture. Their proposed method has the advantage that it can display 2-D motions without decreasing the frame rate and does not require the alignment of two transducers.

For more details regarding a very comprehensive review of multiple-beam, angle-independent velocity vector methods which more than 30 laboratories and companies had developed from the 1970s to 2000, the interested readers are referred to Dunmire *et al.*'s excellent review article [59]. The review includes most proposed crossed-beam systems reported before its publication, including multiple single-element transducers, an arrangement of two array transducers, or one array transducer with multiple transmit and receive subapertures. It is concluded that despite limitations, the cross-beam technique is a simple and feasible approach to accurate velocity measurements and improved flow rate estimations, leading to a better understanding of blood flow dynamics.

Tortoli *et al.* [60] used two transducers to determine the beam-to-flow angle. One transducer called measuring transducer is used to measure the Doppler frequency shift, while the other called reference transducer is used to make a right angle to the blood flow by rotating it. Whether the angle between the reference transducer and the blood flow equals 90° can be determined from the spectral symmetry of a Doppler signal around the dc fre - quency. If the spectrum is symmetrical with respect to the dc frequency, then the mean frequency is zero, indicating that there is no Doppler frequency shift and thus the beam-to-flow angle is equal to exactly 90°. The Doppler shift frequency obtained from the measuring transducer contains two unknowns, i.e., the angle between the measuring transducer and the blood flow and the velocity. The angle between the measuring transducer and the blood flow can be obtained as 90° minus the angle between the reference and the measuring transducer. The latter angle is already available, because we can measure the angle between those two transducers. One major drawback of this method is that the reference transducer should be rotated in finite steps to find the angle that makes a right angle with the blood flow using spectrum analysis and this procedure has to be performed repetitively at each sampling volume.

Pastorelli *et al.* [61] presented a new 2-D vector Doppler system that was engineered to perform *in vivo* real-time 2-D vector Doppler acquisitions. This new system is based on a FEMMINA platform connected to an Esaote S.p.a. LA523 linear array probe, and uses an Esaote Megas as its analog front-end. Standing for Fast Echographic Multi-parametric Multi Image Novel Apparatus, FEMMINA is a hardware and software platform that can process and display RF echoes in real time [62]. In order to validate results produced by this new system, the 2-D vector Doppler technique has been extensively simulated by means of the Field II ultrasonic field simulator. Both *in vitro* and *in vivo* characterization of the presented 2-D vector Doppler system was carried out in laboratory, the results of which were presented. The system operates by performing two independent 1-D Doppler estimations along different directions in order to completely reconstruct the velocity vector within a scan plane. The reconstructed velocity is presented in real time by FEMMINA as a vector superimposed on the B-Mode image.

An extensive set of simulations was performed in order to establish a gold standard regarding vector Doppler 2-D techniques, and to be able to assess the performance of the 2-D Doppler system by comparing simulated and experimental results. Steering angles between the directions of the two independent 1-D Doppler measurements of around ±12° are found to be sufficient to estimate the vector information.

The experimental results were found to be in good accordance with the simulations. For both *in vitro* and *in vivo* experiments, vector velocity maps consisting of three lines of sight were produced in real time. For *in vitro* experiments, some angles produced worse performance. This can probably be attributed to the combination of the wall filter action on the two independent lines of sight and to the backscattering radiation pattern of the employed Doppler thread phantom. For *in vivo* experiments a particular scanning approach has been used, and the estimated velocity vectors seem to produce good results, even though it is hard to quantify the quality of the velocity estimate *in vivo*. The total time required to perform the 2-D Doppler estimation depends both on acquisition time and on processing time. In particular, the data acquisition time is directly related to the desired accuracy of the 1-D Doppler estimates, and consequently of the final 2-D Doppler estimates. Moreover, scanning a larger area requires a longer acquisition time. In particular, a 10 x 3 velocity vector map can be produced six times per second while showing in the background higher frame rate B-Mode images. The realized real-time 2-D vector Doppler system is fully certified as hospital equipment, and thus can be employed to carry out an experimental characterization of the 2-D Doppler technique in a clinical environment.

Recently, Lu *et al.* [63] have presented the principle and feasibility of estimating a 2-D blood flow velocity with a fast Fourier transform (FFT)-based high frame rate imaging method. From B-mode images reconstructed using FFT at angles of ±15° relative to the transducer beam axis, two velocity components are produced at those angles to produce a 2-D vector velocity map.

The multiple beam method suffers from performance degradation when imaging deep lying blood vessels because the angle between beams tends to decrease. The reduced angle results in poor estimation of velocity. This is similar to the situation that as the condition number of a matrix increases, the system of equations becomes more difficult to solve and more sensitive to measurement noise. When using multiple separate transducers to generate multiple beams, accurate alignment is not a trivial task. Alternatively, an array transducer can be used to generate multiple beams, but the available aperture size needs to be sacrificed, which in turn degrades the estimation accuracy.

III. Spectral Broadening Methods

The second method utilizes the output spectrum broadening effect of ultrasound Doppler signals arising from the transit time and beam geometry. When the incident field is a plane wave of infinite extent, which is in general not true for finite aperture transducers, the Doppler effect is zero for radiation normal to the direction of motion [64]. Consequently, the probing of flows transverse to the axis of finite diameter beams, particularly focused beams, is feasible. Experiments were performed to verify theoretically predicted transverse flow spectrum shapes and to investigate the feasibility of using the spectral width for measuring continuous and pulsatile flows. Newhouse *et al.* [65] reported a technique for estimating the total velocity vector using only two transducers by measuring not only Doppler frequency shift but also its bandwidth.

Newhouse *et al.* [64,65] and Tortoli *et al.* [66] established the relationship that the maximum frequency f_{max} of a measured Doppler spectrum equals the sum of the Doppler shift frequency f_d and one half of the Doppler spectrum bandwidth B_d. Letting F and W denote the effective focal length and the diameter of a circular ultrasound transducer with a wavelength λ, respectively, the above relationship can be mathematically expressed as:

$$
\begin{aligned}
f_d &= \frac{2f_0 v}{c} cos\theta, \\
B_d &= \frac{2Wf_0 v}{Fc} sin\theta, \\
f_{max} &= f_d + \frac{B_d}{2} = f_d + \frac{Wf_0 v}{Fc} sin\theta.
\end{aligned}
\tag{5}
$$

After manipulation, the velocity v and the beam-to-flow angle θ are expressed as follows:

$$
\begin{aligned}
v &= \sqrt{\left(\frac{cf_d}{2f_0}\right)^2 + \left[\frac{Fc}{Wf_0}(f_{max} - f_d)^2\right]^2}, \\
\theta &= \tan^{-1}\left(\frac{2F}{W}\frac{f_{max} - f_d}{f_d}\right).
\end{aligned}
\tag{6}
$$

It follows then that the axial and transverse velocity components can be found as $vcos\theta$ and $vsin\theta$, respectively.

Tortoli *et al.* [66] verified experimentally the theory of Newhouse *et al.* that the transverse Doppler bandwidth is invariant with the range of cell depth and dimensions. Their experiments on a thread phantom showed that both magnitude and direction of a transverse velocity vector could be determined by means of spectral analysis based on FFT.

Chiang *et al.* [67] and Lee *et al.* [68] utilized the above relationship to automatically determine the magnitude and angle of blood flow velocity. They provided experimental results on a UHDC flow phantom (Shelley Medical Imaging Technologies, London, ON, Canada) for generating pulsatile flows and on a carotid artery, using a color Doppler ultrasound system (CFM-725, VingMed Sound, Horten, Norway) with a 2.5-MHz, 1.47-cm diameter, 7.8-cm focal length, four-element annular array transducer. Li *et al.* [69] used the dependence of Doppler bandwidth on Doppler angle to estimate the transverse flow component, where the variance of the bandwidth is estimated from correlation, i.e., using the well-known Kasai's method.

Lu *et al.* [70] proposed the use of limited-diffraction beams to estimate the transverse Doppler velocity, and verified the efficacy by simulation and experiment.

The main drawback of the spectral broadening method is that it is valid only for laminar steady flow, but not for pulsatile or turbulent flows. Also, when there are many different velocities present in a sample volume, the spectral broadening model does not hold true.

IV. Speckle Tracking Methods

The third category is the speckle tracking method that capitalizes on the fact that the speckle pattern translates from one pulse emission to the next due to blood flow [71-79]. Fig. 2 shows that a small region of speckle moves from a position $P(x_0,z_0,t_0)$ in one frame at time t_0 to another position $P(x_1,z_1,t_1)$ in the next at time t_1. The axial and lateral axes are

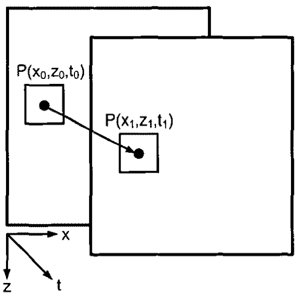

Fig. 2. Translation of speckle pattern from one frame to the next.

denoted by z and x, respectively. This new technique for blood velocity imaging is based on tracking the motion of a speckle pattern produced by blood, and has the advantage that it is angle independent for in-plane flow and that it is not subject to aliasing.

Nakajima *et al.* [71] found that there exists a linear relationship between the blood flow velocity of fluid and the zero-crossing count of speckle intensity fluctuations when the velocity is small.

Trahey *et al.* [72] presented a new technique for blood velocity imaging based on tracking the motion of the speckle pattern produced by blood. Unlike Doppler velocity determination, the technique is angle-independent. Initial *in vivo* experiments provided promising results. They found the motion of two small, nonoverlapping target regions, A and B, in two consecutively acquired image frames by calculating the correlation coefficients between the two rectangular regions. The two representative points $P(x_0, z_0, t_0)$ and $P(x_1, z_1, t_1)$ in Fig. 2 belong to the image segments, A and B, respectively. The normalized correlation coefficients for an axial shift of m and a lateral shift of n are given by

$$\rho(m,n)=\frac{\sum_i\sum_j\left[A(i,j)-\overline{A}\right]\left[B(i+m,j+n)-\overline{B}\right]}{\sqrt{\sum_i\sum_j\left[A(i,j)-\overline{A}\right]^2\left[B(i+m,j+n)-\overline{B}\right]^2}}, \qquad (7)$$

where $\overline{A}$ and $\overline{B}$ are the mean pixel values of the image regions of A and B, respectively, and i and j denote the axial and lateral indices, respectively. The values of m and n that make $\rho(m,n)$ maximum correspond to a new target position, and are equal to quantized versions of $z_1 - z_0$ and $x_1 - x_0$, respectively. In order to track the shift in position of a cloud of moving scatterers in blood, a crosscorrelation method is also used to find the peak position where the correlation between the two regions of interest is maximum. However, the crosscorrelation method requires a large amount of computation with interpolation necessary to attain subsample precision. Also, another requirement is that the speckle pattern for the present pulse emission should be a translated version of that for the previous pulse emission. This necessitates the use of a high frame rate data acquisition system so as to decrease the degree of decorrelation.

In order to alleviate the above computing requirements that are quite demanding, a metric referred to as the sum of absolute differences (SAD) was proposed by the Trahey group as a criterion for angle-independent Doppler velocity estimation to determine the amount of shift between the kernel and search regions in two successive image frames. The algorithm worked well to produce color flow maps of the popliteal vein of a normal female. The SAD measure is simpler to implement than cross-correlation methods, and is defined as follows:

$$\varepsilon(m,n)=\sum_i\sum_j|A(i,j)-B(i+m,j+n)|. \qquad (8)$$

Bohs *et al.* [73,74] described a real-time vector velocity imaging system based on the speckle tracking technique that could determine both the axial and lateral velocity components without aliasing, and presented results from phantom studies showing that the system could accurately estimate velocities of over 2.5 m/s. Their 2-D tracking employed the SAD criterion and a number of SAD printed circuit boards for parallel search processing. However, they did not incorporate in the system static echo cancellation filters whose implementation was a challenging task at that time.

2-D speckle tracking methods are known to overcome the chief limitations of conventional Doppler-based velocity estimators, i.e., angle dependence and aliasing, but the number of lateral lines available is usually very small. This requires the use of interpolation. Geiman *et al.* [75] and Bohs *et al.* [76] proposed a new lateral interpolation technique, termed grid slopes, which is computationally simple and can accurately measure subpixel translations. Using the gradient in SAD coefficients of the first acquisition kernel and the gradient of the SAD coefficients between the kernels of the first and second acquisitions, they determined the subpixel shift without actually performing interpolation. They experimented with a moving string phantom at a transducer angle of 90° to evaluate the ensemble tracking with grid slopes interpolation in the lateral dimension. In order to reduce speckle decorrelation, they used 4:1 parallel receive processing offered by a Siemens Elegra ultrasonic scanner with a 7.5-MHz transducer focused at 3 cm and a PRF of 11.1 kHz. A 2 lateral by 10 axial pixel kernel region and a 4 lateral by 30 axial pixel search region were used for all trials.

To further reduce the amount of computation in a region-based speckle pattern correlation or block matching techniques for estimating 2-D flow velocities from a sequence of ultrasound images, Wang and Shung [77] suggested three-bit pattern correlation algorithms that are efficient in terms of computation, and presented the experimental results on a mock flow loop. The results from the proposed algorithms and crosscorrelation functions showed a good agreement.

Udesen *et al.* [78] used an ultrasound research scanner called RASMUS (Remotely Accessible Software configurable Multichannel Ultrasound Sampling) system where unfocused plane waves were transmitted to estimate blood velocity vectors from the speckle pattern change from one emission to another.

Lu *et al.* [79] used storage correlator arrays to estimate blood flow velocity vector by tracking the speckle pattern in real time.

The speckle tracking method is inherently suitable for 2-D velocity estimation because it tries to find the axial and lateral shifts. The raw echo data need to be collected rapidly. The method works quite well for in-plane translational movements of scatterers, but its performance degrades in the presence of decorrelation between the kernel and search regions due to out-of-plane and/or turbulent flows. It is also quite time-consuming to compute the amount of shift.

V. Lateral Beam Modulation Methods

We will now take a look at the spatial quadrature method which has been relatively recently proposed to estimate velocity vectors. Methods belonging to this category introduces transverse or lateral beam modulation so that the transverse flow which is perpendicular to the beam direction can also be detected. It is to be noted that Lu showed the possibility of transverse Doppler velocity measurement with limited-diffraction beams. In particular, the Bessel beam method may be considered to be the forerunner of lateral beam modulation methods [70].

To distinguish between the forward and the reverse flow relative to a transducer, two lateral beams are used that have a phase quadrature relationship between them. To achieve this relationship, the array aperture is apodized with weighting functions that have the Hilbert transform relationship. By introducing beam modulation in the lateral or elevational direction and making the two beam patterns to be in phase quadrature on receive, Anderson [80] demonstrated the feasibility of estimating the transverse velocity. To obtain the lateral beam modulations that have the property of spatial quadrature, he applied two different apodizations (i.e., one is even and the other is odd) to the transducer aperture so that the resulting far field beam patterns can be in spatial quadrature. Both beam patterns form a Hilbert transform pair in the spatial domain like both cosine and sine functions in the time domain. Some drawbacks to the approach are that the spatial quadrature relationship can be maintained only at the focal point and the lateral velocity can be estimated following the determination of the axial velocity.

Anderson [81] provided experimental results on wire targets and sponges. In order for the spatial quadrature modulation to be distinguished from the axial modulation, the data must first be axially aligned over the ensemble. Given the analytic form of the spatial quadrature modulation, the sign of the lateral velocity component is unambiguous. Because scatterers modulate echoes in the spatial domain, the velocity of blood flow can be estimated. The derivation of the spatial quadrature relationship assumes the use of a continuous wave, which is not the case in real ultrasonic imaging. The newly proposed spatial quadrature concept was experimentally verified with a computer-controlled, calibrated flow pump and a Siemens Elegra scanner [82]. Both the true and estimated lateral velocity profiles computed off- line showed a good agreement.

Anderson *et al.* [83] compared the bias and variance of spatial quadrature and speckle tracking over a range of flow rates. Speckle tracking was found to provide a smaller variance and bias, while the spatial quadrature estimator provided estimates for the entire velocity range at a single PRF.

Anderson [84] presented a method of demodulating the even and odd return echoes, in which each of them is represented as a multiplication of sinusoids with shift frequencies in the axial and lateral directions and the axial and lateral Doppler shift frequencies are separated independently. Experiments were conducted on a vessel phantom, and the axial velocity, flow rate, and flow angle were accurately estimated. The relative errors of estimated flow rates were 3 % and 6 % for flow at Doppler angles of 90° and 60°, respectively.

Jensen and Munk [85] presented a new method for determining the movement of an object from a field with spatial oscillations in both the axial direction of the transducer and in one or two directions transverse to the axial direction. The method is based on Munk's thesis work [86]. Also, a method was presented for making a field that is transversely modulated along with a suitable 2-D velocity estimator. The method was introduced under the name of transverse oscillation. Simulation results for a fixed flow rate of 1 m/s were obtained at various angles. The method requires a total of three receive beamformers, i.e., lateral in-phase, lateral quadrature, and axial beamformers. Blood velocities can be estimated only at a certain depth where the quadrature phase relationship between a pair of transversely modulated beams can be maintained.

Munk and Jensen [87] presented experimental results of an approach published in their May 1998 paper [85], where they introduced a pair of transversely oscillating fields, displaced one quarter of the lateral oscillation period relative to each other, through receive beamforming to generate Doppler frequency shift in the lateral direction. Experiments were carried out on a ruby point reflector that was moved horizontally parallel to the surface of a transducer and a very fine sponge that was fixed relative to the moving transducer. For a flow angle of 90°, the bias turned out to be −18 %.

Jensen and Lacasa [88] manually set the blood flow angle, and focused the ultrasound beam along the direction of the blood flow. The returned echoes were received at some focal point along the flow direction for two consecutive pulse emissions, and were crosscorrelated to estimate the velocity in the flow direction. The major drawback of this approach is that the flow-to-beam angle must be available beforehand so that the beam can be focused parallel to the blood flow direction.

Jensen [89] presented a new method for simultaneously estimating both axial and lateral velocities using a transversely modulated ultrasound field for probing a moving medium under investigation. The approach can find the velocity with a standard deviation of 10 % relative to the maximum velocity, when the velocity is orthogonal to the ultrasound beam. The results were obtained using Field II computer simulations and a modified autocorrelation method. A relative accuracy of 10.1 % is obtained for the transverse velocity estimates for a parabolic velocity profile for flow transverse to the ultrasound beam and an SNR of 20 dB using 20 pulse-echo lines.

As mentioned before, drawbacks to the spatial quadrature are that maintaining the spatial quadrature relationship at other than the focal point is difficult and estimation of the lateral velocity needs to be preceded by that of the axial velocity. Also, application of Hilbert transform across an active transducer is not an easy task especially for wideband excitations. The downward shift of center frequency with increasing depth is also another concern that needs to be taken into account.

VI. Other Methods

In this category, we discuss aperture domain methods proposed by Walker [90] and by Wang *et al.* [91] along with directional crosscorrelation methods proposed by the Jensen group [92~97]. We note that the latter method may also be categorized into the speckle tracking methods because the crosscorrelation operation inherently relies on the speckle pattern change.

Walker [90] proposed using individual echoes received on each array element to estimate the vector time shift between two consecutive firings. The scheme was proposed to overcome the problem that different parts of an array lie at different angles with respect to the blood flow vector. The possibility was also suggested of using signals summed over groups of elements rather than signals from individual elements to simplify the hardware implementation.

The basic idea of Wang *et al.*'s paper [91] is to determine both the axial and lateral velocities simultaneously by representing the arrival-time difference across the array aperture as a linear function with a slope and an intercept, and determining both of the latter two parameters using linear regression. The basic idea is similar to that of Walker in that both utilize aperture domain data, but differs in introducing subaperture processing as well as polynomial modeling. The aperture data refer to the data from individual array elements before applying the delay and sum operation for beamforming. Some theoretical bounds including the Cramer-Rao lower bound are established, followed by Field II-based simulation and DiPhAS (Digital Phased Array System) data acquisition system-based experimental results. Both simulation and experiment are considered to be preliminary. Nevertheless, they help corroborate the methodology. When a target moves from one position to another, the arrival time of the echo received on some fixed array transducer element will also change from one value to another.

By taking the difference between the two arrival times, referred to as the arrival-time difference, and making some approximations, they have been able to represent it as a function of both the axial and lateral velocities. Clearly, this is a novel idea. Actually, the function is a first-order polynomial with a slope and an intercept. The slope corresponds to the lateral velocity, and the intercept to the axial velocity. So they applied linear regression to simulated and experimental data to estimate two parameters, i.e., the slope and the intercept. In contrast, Walker employed crosscorrelation using the 2-D velocity as a lag variable so that the 2-D velocity can be determined as that value which maximizes the crosscorrelation function. At the end of their paper, the possibility is mentioned of estimating a 2-D velocity using a 1-D array and a 3-D velocity using a 2-D array. In simulation, the Field II was used, and the scatterers were moved at a constant velocity. The focal point was 30 mm from a 64-element, 5-MHz transducer.

Experiments on real flow were not carried out, but a linear array transducer was moved along the axial and lateral directions relative to a fixed gelatin phantom containing glass beads for speckle generation. Note that this arrangement is different from the real situation where the scatterers inside a blood vessel move relative to an imaging transducer. The number of firings was set to 16 to get good results by averaging over multiple pulse transmissions in computing the Kasai's autocorrelation method, but the number of firings needs to be further reduced to make it amenable to real-time implementation of 2-D color flow imaging. For actual implementation, the receive aperture needs to be divided into a number of small subapertures, resulting in poor lateral beam focusing and sensitivity. After acquiring returned echoes on each of 64 receive elements for each of 64 transmit elements without applying any time delay, the aperture domain data were reconstructed using full dynamic focus on both transmit and receive. A complete dataset consisted of 4096 RF A-scans.

Also, Jensen [93] performed beamformation using a linear array along the direction of the flow for a given depth, and found the correct velocity amplitude for a purely transverse velocity direction. However, the method works under the assumption that the angle between the emitted beam and the flow direction is known in advance. He rationalized the assumption by citing the fact that the angle could be found from a B-mode image as in conventional spectral velocity estimation. Determining the angle manually is cumbersome so the process needs to be automated. In the case that the flow is turbulent, it may be difficult to determine the angle from the B-mode image. A simple method of stationary echo canceling was used, which is just subtracting the mean of all the echo signals from the current echo signal. It is called a mean subtraction algorithm. To estimate the maximum point of the crosscorrelation function, an interpolation method was used to locate it exactly without increasing the amount of computation much.

Using an experimental flow rig, Jensen and Bjerngaard [94] showed that the vector velocity can be estimated by emitting focused directional signals along the flow direction which is assumed to be known and estimating the shift between two consecutive echoes returning from the same depth.

Kortbek and Jensen [96] used a normal transmission of a focused ultrasound field. The principle is that if the beam direction is the same as the flow direction, then the current echo is a delayed replica of the previous echo from the same scatterer site; otherwise, the current echo is a delayed, as well as decorrelated, version of the previous echo. Hence, the flow angle can be determined by finding the beam direction that maximizes the correlation between a pair of ensemble echoes that are usually one pulse emission period apart.

Previously, the Doppler beam angle was manually determined from the B-mode image. Jensen and Oddershede [97] presented a method for estimating both velocity magnitude and angle using crosscorrelation of the received signals. In the crosscorrelation, the angle is determined by interpolation using a parabolic approximation of the peak. The principle of automatic angle determination is based on the fact that when the beamforming direction coincides with the true flow direction of scatterers, the correlation functions attain the highest correlation values.

Katakura and Okujima [98] proposed a new method of measuring the complete vector components of a target velocity using a wide transmission beam and in-phase accumulation procedure on receive, but addressed only the case of measuring the velocity of point reflectors. The method consisted of two steps: Fourier transformation of the received signals in the direction of the time axis (the ultrasonic pulse emission number) and projection integration in the polar-axis direction. It was argued that the method could also be applied to speckle-generating targets if the effect of reverberation were not significant.

VII. Conclusions

In this article, we have surveyed five major categories of vector Doppler methods for estimating 2-D blood flow velocity. To recapitulate, they are multiple beam methods, spectral broadening methods, speckle tracking methods, lateral beam modulation methods, and other methods. Each method has its own advantages and disadvantages. All 2-D methods should assume that the flow lines are parallel to, or confined to, the imaging scan plane. This limitation can be overcome by resorting to a full 3-D vector Doppler approach. Usually, the estimated 2-D blood flow velocity is displayed by superposing a number of arrows on top of a B-mode image with their individual directions and lengths representing the directions and magnitudes of the velocity, respectively. The display method is similar to representing the results of optical flow estimation. Up to now, to the best of our knowledge, no one 2-D vector Doppler method has matured to the level of commercialization into an ultrasound Doppler scanner. Active research is still going on to find better vector velocity estimation methods.

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