DOPPLER BEAM STEERING FOR BLOOD FLOW VELOCITY VECTOR IMAGING

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ABSTRACT

Angle independent blood flow velocity vector measurements could be of a great value for the diagnoses of many vascular and cardiac diseases, such as stenotic carotid and coronary lesions. Although many methods such as speckle tracking and multiple transceivers have been proposed, they have limitations. In this paper, a blood flow velocity vector measurement method called Doppler beam steering is proposed for either 2D or 3D imaging. In the method, a narrow ultrasonic beam is steered at three or more angles in transverse directions. The changes in frequency of the returned echoes provide radial components of blood flow velocity along these directions. Blood flow velocity vector is then calculated from these components and images of the velocity vector distribution can be obtained. Both *in vitro* and *in vivo* experiments were carried out to verify the method. Velocities measured with this method agree well with those measured directly with a flowmeter. Results show that this method is capable of estimating blood flow velocity vectors and thus may provide useful diagnostic information that is difficult to obtain with conventional Doppler imaging method.

INTRODUCTION

Blood flow velocity measurement is very important in diagnosing various cardiac and vascular diseases. Although conventional Doppler blood flow imaging methods are widely used, they have a major limitation: only the velocity components along the lines of sights of ultrasound beams are measured [1,2]. Because of this limitation, conventional Doppler methods are not only unsuitable for quantitative evaluation, but also prone to misinterpretation of complicated blood flow patterns presented in curves, branches, bifurcations, and tortuous vessels where the direction of flow velocity may be helical

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in nature [3,4]. In addition, the presence of diseases and its associated turbulence in the arteries cause the blood to move in many different directions, e.g. non-radial jets. These complicated changes in the velocity vectors are not obvious and cannot be detected by conventional Doppler methods including color Doppler imaging.

To overcome these problems, various techniques for measuring two- or three-dimensional (2D or 3D) velocity vectors (true velocity) have been proposed. These techniques fall roughly into three categories: speckle tracking, multiple transceivers, and projection computed velocimetry. In speckle tracking [5-10], the direction and magnitude of the displacement of local blood speckle pattern in consecutive B-mode images are measured. Because the time between image acquisitions is known, velocity vectors can be obtained. This technique overcomes the angle dependence and aliasing limitations of current Doppler velocity measurement methods [11]. Multiple transceiver method uses multiple piezoelectric transducers mounted on the top surface of a lens to measure blood flow velocity vector [12]. Ultrasound beams from the transducers are focused to produce multiple parallel and closely spaced beams. Motion of blood cells is tracked with RF correlation techniques as the cells move along the beams and from beam to beam, and thus 3-D flow velocity vectors can be estimated. Finally, projection computed velocimetry can also be used for blood flow velocity vector measurement [13]. Target velocity components both along and perpendicular to the beam axis are obtained with this method in two steps. The first is to perform the Fourier transform of received signals in time domain (pulse-echo signals). The second is to get projection integration in polar direction. These two steps provide two velocity components along and transverse to the beam axis, respectively.

The methods mentioned above for flow velocity vector measurement suffer from several limitations. The Speckle tracking techniques have problems due to relative motion of ultrasonic scatters within a sample volume in a low frame rate flow measurement system, the pixel-based pattern correlating and matching algorithms employed, and a low signal-to-noise ratio (SNR) of acquired images. Multiple transceiver techniques complicate the scanning system, and may lead to an unjustifiable cost increase. Although the projection computed velocimetry method has a merit of separating the composite velocity vectors with a linear processing, it requires a lot of computation time for either 2D or 3D Fourier transforms for each of the measured points.

In this paper, we propose a blood flow velocity vector measurement method called Doppler beam steering for 2D or 3D imaging. In this method, a narrow ultrasonic beam is produced and steered at either three or four angles. The changes in frequency of the returned echoes provide radial components of the blood flow velocity and are used to calculate velocity vector. Selecting sample volumes over the entire region of interest, 2D or 3D velocity vector images can be constructed.

PRINCIPLE

In conventional pulse and color Doppler methods, velocities at different positions of an image are obtained from the frequency shift with the equation:

$$|\vec{v}| = \frac{f_d c}{2 f_0 \cos\theta}, \tag{1}$$

where $|\vec{v}|$ is the velocity magnitude, c is the speed of sound in tissues, f_d is the Doppler frequency shift, f_0 is the center frequency of the transducer, and θ is the angle between transducer axis and the direction of velocity. Because θ is usually unknown, conventional

color Doppler method can not obtain a true velocity (both magnitude and direction) and only the radial component of the velocity can be measured.

To overcome this problem, a Doppler beam steering method is proposed (Fig. 1). In this method, three focused beams (the principle is the same for four or more beams) are steered at steering angles, γ_i, $(i = 1, 2, 3)$, and azimuthal angles, ϕ_i, $(i = 1, 2, 3)$. The center of the aperture of the ultrasonic transducer is located at the origin of the coordinates, (0,0,0). The radial components of the velocity at three points of a volume (a small volume that covers the three points) of blood are measured.

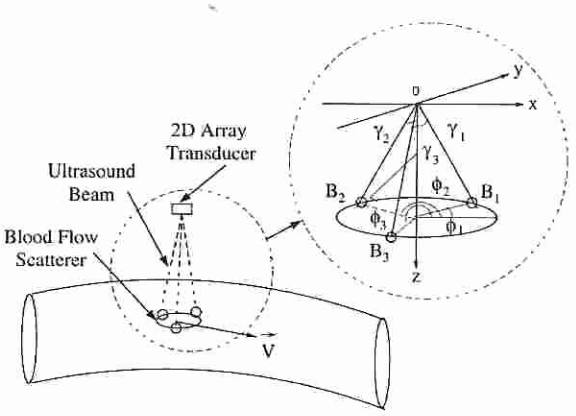

Figure 1. An illustration of Doppler beam steering method.

In Figure 1, it is assumed that the volume of blood have no relative random motions. If $\gamma_i = \gamma$, the three components and the magnitude of the velocity vector of blood volume can be obtained with the matrix [14,15]:

$$\begin{bmatrix} \sin\gamma\cos\phi_1 & \sin\gamma\sin\phi_1 & \cos\gamma \\ \sin\gamma\cos\phi_2 & \sin\gamma\sin\phi_2 & \cos\gamma \\ \sin\gamma\cos\phi_3 & \sin\gamma\sin\phi_3 & \cos\gamma \end{bmatrix} \begin{bmatrix} v_x \\ v_y \\ v_z \end{bmatrix} = \begin{bmatrix} v_{r_1} \\ v_{r_2} \\ v_{r_3} \end{bmatrix}, \tag{2}$$

where v_{r_i} $(i = 1, 2, 3)$ are radial velocities along different beams. Because each velocity component is a simple linear function of the radial components of the velocity, velocity calculations could be performed in real time.

IN VITRO AND IN VIVO EXPERIMENTS AND RESULTS

Both *in vitro* and *in vivo* experiments were conducted at The Jobst Vascular Center at Toledo Hospital, Toledo, OH. *In vitro* experiment was performed with a peripheral vascular Doppler Flow Controller and Pumping System (Model 700, ATS Laboratories, Inc., Bridgeport, CT) as shown in Figure 2. The system consists of a test fluid reservoir, a positive displacement variable-speed pump, a "flow integrator" to control the amount of flow rate, two in-line flowmeters, a pressure gage, and a flow phantom of 13.8 cm wide × 21 cm long × 10 cm deep. The system has the capability to detect flow at varying depths. The phantom is made of rubber-based tissue-mimicking material. Its speed of sound is

in the range between 1450 and 1473 m/s, and the frequency-dependent attenuation is about 0.5dB/cm/MHz at room temperature (23° C). Four cylindrical flow channels, drilled through the phantom material, were available for flow rate measurement. The diameters of the channels are 2, 4, 6, and 8 mm, respectively, and their centers are located at 25 mm from the top surface of the phantom. Experiments were conducted for all channels. ATS 707-G Doppler fluid was used in the experiment. The system can provide steady-state flow rates in the range of 20 to 950 ml/minute. In the experiment, the pump speed was set manually to provide flows around 200 ml/minute. Actual flow rate was read from the flowmeters display during data collection. Using these in-line flowmeters, continuous monitoring of the flow rate during the testing was performed to guarantee a stable flow measurements. Standard ultrasound gel was placed over the phantom for acoustic coupling.

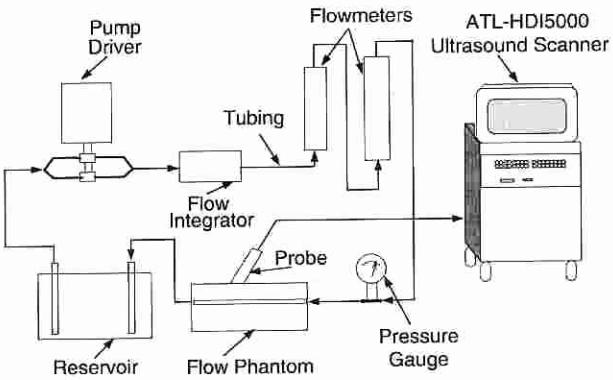

Figure 2. *In vitro* experiment system for flow measurement with both flow meters and the Doppler beam steering methods.

A wideband (4-7 MHz) ultrasound probe (L7-4) was used. It was connected to an ATL (Advanced Technology Laboratories, Bathell, WA) HDI 5000 ultrasound scanner. The probe was manually held to insonify the flow channel. To guarantee that flow sample was taken at the center of the channel without lateral distortion, B-mode images were used to measure the channel diameter to match the actual diameter. The diameter of the beam was about 1.0 mm, and the steering angle, γ, was set to 5° off the vertical axis in four directions.

The result of the *in vitro* experiment with three Doppler beams was obtained from the radial velocities measured within the channels and with Eq. (2). The mean value of the magnitude of the velocity vector of a small volume of the Doppler test fluid was calculated. The result (29.76 mm/s) is compared well with that measured with the flow meter (26.24 mm/s).

In vivo experiment was conducted on the carotid artery of a patient using the same ultrasound scanner as that used for the phantom. B-mode Doppler flow carotid ultrasonography was performed with the patient supine. The head was extended and rotated away from the side to be studied. The anatomic portion of the study was performed first. A cursory survey of the internal and external carotid arteries was done prior to store digital images of the study. This was done in a relative short time period to familiarize us with the patient anatomy and to assess the scanning approach (anterior, lateral or posterial) that

the patient anatomy and to assess the scanning approach (anterior, lateral or posterial) that provided optimal access to the carotid artery. Using a low viscosity gel as acoustic coupling agent, the artery was scanned in cross-sectional plane, proximal to the distal internal carotid artery, using the same probe used in the *in vitro* experiment. Using conventional duplex scan, a pulsed Doppler was superimposed on a two dimensional grey-scale image and a single adjustable sample volume was used to obtain quantitative and qualitative information regarding the blood flow in that particular segment. Such spectral information allows us to calculate maximum velocities and use this information to ascertain the measurements. Scanning sequence was recorded from different angles and stored digitally. Results were obtained for both maximum (65.62 mm/s) and minimum (37.63 mm/s) velocities during a cardiac cycle.

DISCUSSION

The results obtained from both *in vitro* and *in vivo* experiments have shown that the Doppler beam steering technique is capable of estimating 3-D blood flow velocity vector with either three or four beams. This technique can obtain the true velocity that current Doppler blood flow system can not.

In the *in vitro* experiment, there is a small difference between velocities obtained with the experiment and flowmeters. This difference is due to errors in the estimation of steering angles. Using an accurate control of steering angle such as electronic steering, the accuracy can be improved. If the errors in Doppler beam steering measurements are random and independent, the standard deviation can be estimated[16]. If the steering angles are reduced to minimize the sampling volume and increase the number of velocity vectors measured within a region of interest, the errors of the three components of blood flow velocity vectors will be higher.

To compare the Doppler beam steering with the multiple transceiver method, the radial velocity components of one point in center of the channel of the phantom was measured with four beams incident from different angles. The results are also compared well with those obtained with the Doppler beam steering method. The advantage of the Doppler beam steering method is that it is easier to implement and thus may simplify imaging systems.

Because the speed of sound is finite in tissues, it takes time to get echoes for each transmit beam. Therefore, image frame rate will be reduced if more beams are used in the Doppler beam steering method to increase velocity accuracy.

CONCLUSION

Conventional Doppler ultrasound methods do not provide complete information about blood flow velocity vectors. In addition, these methods are prone to errors when the beam-to-flow angles is near 90°. To overcome these limitations, a new method for measuring 2D or 3D blood flow velocity vector distribution has been proposed in this paper. The principle of this method for measuring velocity vectors of moving blood flow cubes in three dimensions has been described. Both *in vitro* and *in vivo* experiments have been carried out with the method. Three and more steered beams have been used to measure the velocity vector of a small cube of blood volume. Results show that the new method can be used to obtain flow velocity vector information that are unobtainable with conventional methods. The usefulness of the proposed technique and its potential for practical use in 3D

blood flow velocity vector estimation are demonstrated. The method is especially useful for large blood vessels where the flow velocity in a small volume of blood is approximately uniform. Combining standard vascular and cardiac imaging techniques with the proposed method for the analysis of flow velocities in the arteries, the diagnosis and management of diseased lesions related to blood flow can be improved.

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