Two-dimensional Blood Flow Vectors Obtained with Bidirectional Doppler Ultrasound

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***Abstract*—Precise measurement of blood flow is important because blood flow closely correlates formation of thrombus and atherosclerotic plaque. Among clinically applied modalities for blood flow measurement, color Doppler ultrasound shows two-dimensional (2D) distribution of one-dimensional blood flow component along the ultrasound beam. In the present study, 2D blood flow vector is obtained with high temporal and bidirectional Doppler ultrasound technique. Linear array probe with the central frequency of 7.5 MHz and an ultrasound data acquisition system with 128 transmit and 128 receive channels were equipped. Frame rate of 5 kHz was achieved by parallel receive beam forming with a wide transmitted wave. The flow velocity was measured from two different angles by beam steering. The interval of two measurements was 0.8 msec and it was considered as almost one moment to obtain 2D blood flow vector. B-mode image and 2D blood flow vector of the pulsatile flow in a carotid artery model showed small vortex at the bifurcation area. The method was also applied for visualization of *in vivo* blood flow vector in human carotid arteries. 2D blood flow measurement may predict the risk area of thrombus and plaque formation induced by abnormal blood flow.**

I. Introduction

Cardiovascular disease is the common problem in all over the world. Blood flow measurement in the cardiovascular system is important because it closely correlates with the thrombus or plaque formation. In clinical settings, ultrasound Doppler technique, phase contrast MR angiography and optical flow of contrast media have been used. Among those imaging modalities, ultrasonography has been widely used as a simple and useful tool.

Although the blood flow can be measured with pulsed Doppler ultrasound, only one-dimensional blood flow component along the ultrasound beam is measured by conventional ultrasound systems. Color Doppler ultrasound is often considered to visualize two-dimensional (2D) blood flow, however, color Doppler ultrasound just shows 2D distribution of one-dimensional blood flow. Thus, the conventional color Doppler method is not enough to measure the complex flow dynamics in the stenosis or bifurcation regions. Precise measurement of blood flow is important for understanding local flow dynamics. Especially, turbulent flow or fluid shear stress affect the formation of thrombus or development of atherosclerotic plaque. Thus, the measurement of the 2D blood flow may lead to early diagnosis of the cardiovascular diseases.

The blood flow structure in left ventricle (LV) has been assessed by echo-dynamography (EDG) [1-6]. EDG is a smart visualization technique in echocardiography in which 2D distribution of blood flow vectors in cardiovascular system is deduced by applying fluid dynamics theories into Doppler velocity datasets. EDG has been validated by numerical simulation and particle image velocimetry of model circulation [7]. However, the complete 2D blood flow analysis is impossible in EDG.

In the present study, a method of measuring actual 2D blood flow vector with high temporal and bidirectional Doppler ultrasound technique realized by parallel beamforming is proposed.

II. Methods

A. Data Acquisition Setup

A linear array probe with a central frequency of 7.5 MHz (L7-4, Aloka, Tokyo, Japan) was connected to an ultrasound acquisition system with 128 transmit and 128 receive channels (V-1 Data Acquisition System, Verasonics, Redmond, WA, USA). The pulse repetition frequency was 5 kHz. The number of the point of measurement was 1664 points along the depth direction and 128 points along the transverse direction. The parameters of the ultrasound system used in the present study are shown in Table 1.

TABLE I. Parameters of the Ultrasound System

Parameter	Value
Transducer	Linear array
Number of active elements	128
Pitch	0.2 mm
Kerf	0.02 mm
Center frequency	7.5 MHz
Number of cycles pr. Pulse	3
f_{prf} of pulse	5 kHz
Number of transmit events at one set	23
f_{prf} of set	40 Hz
Sampling frequency	30 MHz
Apodization in receive	Tukey
Focal point	Parallel

B. Parallel Beamforming and Wall Filter

Conventional beamforming generates a single line per transmit, thus demands 128 transmits to form a frame of data in the present setting. Sound speed is assumed as 1540 m/s in the tissue. After one beam is transmitted, there is double time cost relative to the detection depth to transmit another beam.

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This limitation results in slow frame rate and lack of detail temporal resolution. Parallel beamforming generates 128 acoustic lines per transmit and the frame rate of 5 kHz is achieved with a wide transmitted wave.

The parallel beamforming was performed by applying delay time to each signal received by each element. Tukey window of $\alpha = 0.7$ was used as a receive apodization function [8-9].

$$f_l(t) = \frac{1}{I_l}\sum_{i=1}^{I_l}\omega(i) f_i(t_y + t_{yl\_i}) \quad (1)$$

where I_l is the number of receive elements, and $\omega(i)$ is the apodization function, t_y is the reception time for the depth of y and t_{yl_i} is the reception time for the depth of y at line number l.

The blood-flow signal is buried in the noise on acquired RF signal, since the reflective signal from the blood is much smaller than the reflective signal from a blood vessel wall. Therefore, it is necessary to separate motion of the blood vessel wall, and the motion of blood flow by using wall filter [10-11]. In the present study, a wall filter described below was used.

$$\theta_c = \arg\left(\sum_{i=1}^{I-1} a_i^* a_{i+1}\right) \quad (2)$$

$$a_m' = a_m e^{-i\theta_c m} \quad (3)$$

$$a_m'' = a_m' - \frac{\sum_{n=1}^{N} a_n'}{N} \quad (4)$$

where a is the input complex signal, a'' is the signal of which DC component was removed, I is the number of the measurement point along the depth direction, and N is the number of insonification for each measurement. In this experiment, I = 1664 and N = 8. The reflected signal from blood and from a blood vessel wall were separated by this wall filter.

C. Compound Imaging

Traditionally, a linear array transducer sends a sound signal perpendicular to the probe head, then receives the reflection wave. With compound imaging, the transducer sends signals at multiple angles, allowing it to observe tissue from multiple angles and to eliminate the artifacts. In the present study, compound image was made from combination of seven images from different steering angles (+18°, +12°, +6°, 0°, -6°, -12°, -18°) into a single image.

D. Calculation of 2D Blood Flow Vector

In this experiment, autocorrelation technique was used for the velocity estimation [12]. The autocorrelation technique is described below.

$$x = Ae^{\theta_x} \quad (5)$$

$$y = Be^{\theta_y} \quad (6)$$

$$C = xy^* = ABe^{\theta_x}e^{-\theta_y} = ABe^{\theta_x-\theta_y} \quad (7)$$

$$\varphi = \arg(C) \quad (8)$$

where A and B are amplitude in time to differ, θ_x and θ_y are phase in time to differ, and φ is phase difference. Phase difference is obtained by performing autocorrelation technique to the input signal a'' obtained with wall filtered signal (equation (4)). The velocity is calculated from phase difference with the expression (9) described below.

$$v = \frac{1}{2}\frac{c}{F_c}\frac{\varphi}{2\pi} prf \quad (9)$$

where v is the flow velocity, c is the sound speed, F_c is the center frequency and prf is the pulse repetition frequency. In this paper, c = 1540 m/s, Fc = 7.5 MHz, prf = 2.5 kHz, and $-\pi \le \varphi \le \pi$ were used.

The Doppler signal was acquired by insonifying a scanned region with a parallel beamforming. One-directional blood flow velocity component along the ultrasonic beam was measured by averaging 8 ultrasound pulses in one-directional steering. Autocorrelation technique was used for velocity calculation. Velocity measured at +15° and -15° steering were synthesized to obtain 2D blood flow vector. The interval of two measurements was 1.6 msec and it was considered as almost one moment.

Fig. 1 shows the schematic illustration of calculating 2D blood flow. Each one-directional blood flow vector was obtained in the black or red area, two-dimensional blood flow vector in the blue area was calculated by compounding the two components.

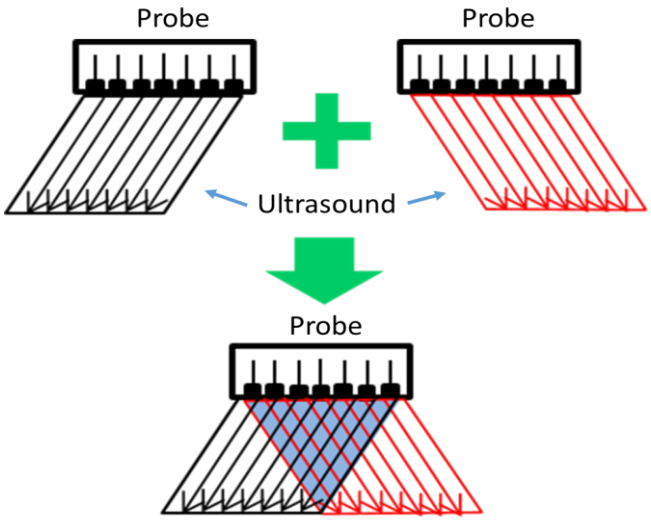

Figure 1. Schematic illustration of the calculation of 2D blood flow vector

2D blood flow vector was calculated from the following equations.

$$V_x = \frac{v_2 - v_1}{2\sin(\pi \cdot \theta / 180)} \quad (10)$$

$$V_y = \frac{v_2 + v_1}{2\cos(\pi \cdot \theta / 180)} \quad (11)$$

$$\vec{V} = (V_x, V_y) \quad (12)$$

where v_1 is the one-directional velocity component measured from the angle of $+\theta$, v_2 is that measured from the angle of $-\theta$, V_x is the calculated velocity component of x-axis, V_y is that of y-axis. $\theta = 15°$ in this study.

E. Carotid Artery Model

Three-dimensional (3D) structure of a human carotid artery was measured by multi-detector row computed tomography (CT) with contrast enhancement. Carotid artery model with bifurcation regions was made based on the CT data. Fig. 2 show the (a) reconstructed carotid artery geometry and (b) macroscopic view of carotid artery model made of PVA (polyvinyl alcohol).

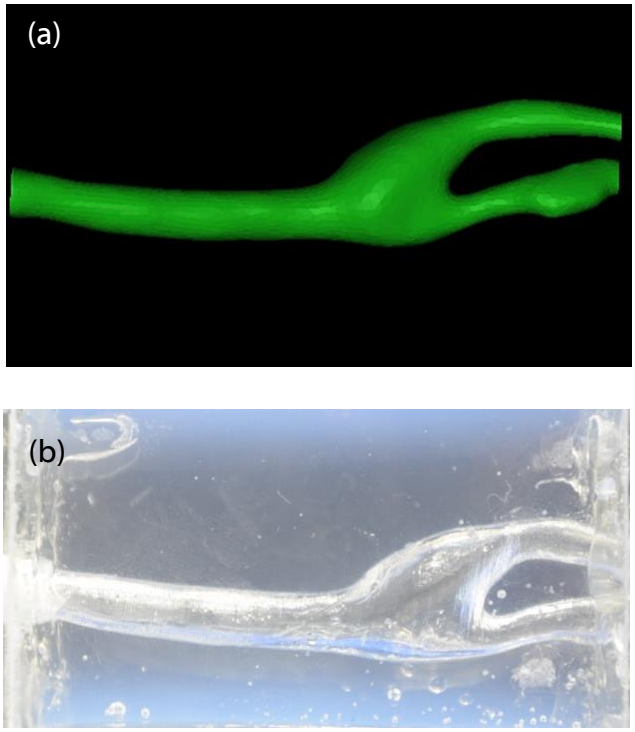

Figure 2. (a) Reconstructed 3D carotid artery geometry, (b) carotid artery model made of PVA (lower)

A pulsatile flow pump (553321, Harvard Apparatus, Holliston, MA, USA) was used in the model circulation. The parameters were stroke volume = 10 ml, heart rate = 60 bpm and systole/diastole ratio = 35 %. Agar particle was added in the water to enhance ultrasound reflection from the flow and it was used as the blood mimicking phantom in the model circulation.

For clinical imaging, ultrasound signal from human carotid arteries were recorded and the same algorithm was applied to obtain high temporal B-mode and 2D velocity vector imaging of *in vivo* carotid artery.

III. Results

A. Imaging of Carotid Artery Model

Fig. 3 (a) shows the 2D blood flow vector at ejection phase and Fig. 3 (b) shows that at late systolic phase, respectively. A vortex and inverse flow are observed at the late systolic phase.

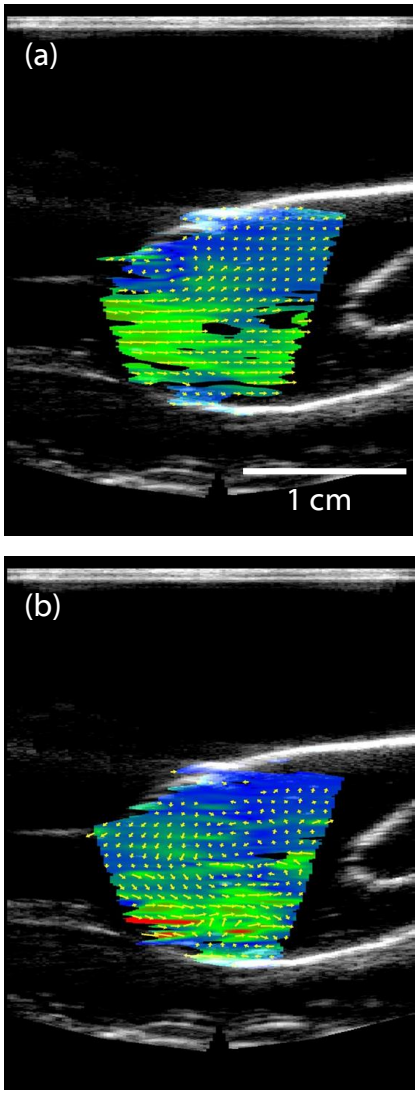

Figure 3. (a) 2D blood flow vector at ejection phase, (b) 2D blood flow vector at late systolic phase

B. Imaging of Human Carotid Artery

Fig. 4 shows the imaging of human carotid artery. The frame rate of (a) B-mode image and (b) 2D vector image were 100 fps. B-mode image shows the dilatation of the carotid artery at systolic phase. 2D Doppler image shows that the peak flow was observed after the dilatation of carotid artery. The finding was only achieved by this high temporal imaging with the parallel beamforming. Parabolic flow profile in the carotid artery is also clearly visualized by this technique (3).

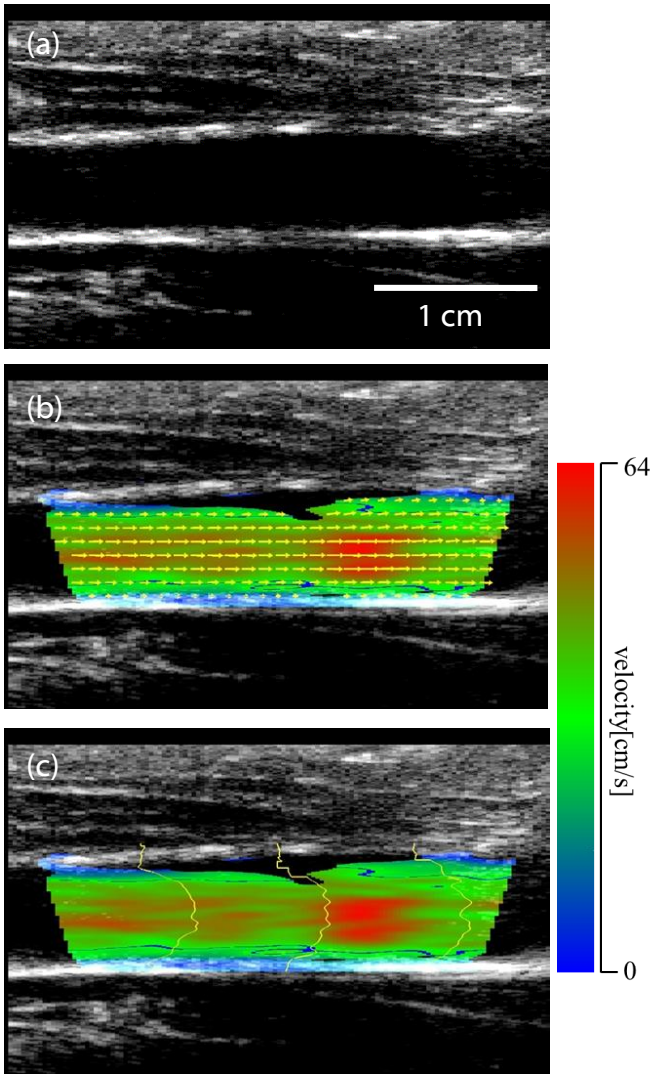

Figure 4. (a) B-mode imaging, (b) 2D blood flow vector and (c) flow profile of human carotid artery

IV. Discussion

Some ideas of 2D velocity measurement by observation of the flow from bidirectional angles were proposed in 1970's [13]. However, the spatial and temporal resolutions were not sufficient to consider the bidirectional measurements focused the same spot at the same moment in those days. In this study, the pulse repetition rate of the parallel beamforming was 5 kHz. One sequence of measurement consists of acquisition of 7 images for compound B-mode imaging (200 μsec x 7 = 1400 μsec), data transfer (2000 μsec), acquisition of 16 images for bidirectional Doppler method (100 μsec x 16 = 1600 μsec), and data transfer (5000 μsec). The total time was 10 msec thus the frame rate of 100 Hz was achieved. The observation depth was 41.7 mm in this setting. As the carotid artery is located in the half depth, the frame rate would be doubled if the observation depth was set at 20 mm.

In the carotid artery model, vortex flow and reverse flow were observed at the bifurcation region. These findings would be important to detect early lesion of atherosclerosis because the plaque or thrombus formation was closely related to the turbulence flow and wall shear stress.

Limitations of the present study is that the accuracy of 2D flow vector was not yet investigated. Future work in comparison with particle image velocimtry is planned for validation of this method.

V. Conclusion

2D blood flow vector is obtained with high temporal and bidirectional Doppler ultrasound technique. B-mode image and 2D blood flow vector of the pulsatile flow in a carotid artery model showed small vortex at the bifurcation area. 2D blood flow measurement may predict the risk area of thrombus and plaque formation induced by abnormal blood flow.

References

[1] Ohtsuki, S. and M. Tanaka. The flow velocity distribution from the Doppler information on a plane in three-dimensional flow. *Journal of Visualization*. 2006; 9(1):69-82.

[2] Tanaka M, Sakamoto T, Sugawara S, Nakajima H, Katahira Y, Ohtsuki S, Kanai H., Blood flow structure and dynamics, and ejection mechanism in the left ventricle: analysis using echo-dynamography. *J Cardiol*. 2008; 52(2): 86-101.

[3] Tanaka M, Sakamoto T, Sugawara S, Nakajima H, Kameyama T, Tabuchi H, Katahira Y, Ohtsuki S, Kanai H., Physiological basis and clinical significance of left ventricular suction studied using echo-dynamography. *J Cardiol*. 2011; 58(3):232-44.

[4] Nakajima H, Sugawara S, Kameyama T, Tabuchi H, Ohtsuki S, Tanaka M, Saijo Y. Location of flow axis line in the left ventricle and its interaction with local myocardial motion. *J Echocardiogr*. 2011; 9(1): 24-27.

[5] Kojima T, Kameyama T, Nakajima H, Khmyrova E, Kurokawa T, Saijo Y. Evaluation of vortex flow in left ventricle by echo-dynamography and phase contrast magnetic resonance angiography. *Conf Proc 34th IEEE Eng Med Biol Soc*. 2012: 2676-9.

[6] Kojima T, Omori A, Nakajima H, Kurokawa T, Kameyama T, Saijo Y. Validation of Echo-Dynamography by virtual color Doppler echocardiography generated from phase contrast magnetic resonance angiography datasets. *Conf Proc 35th IEEE Eng Med Biol Soc*. 2013: 105-8.

[7] Uejima T, Koike A, Sawada H, Aizawa T, Ohtsuki S, Tanaka M, Furukawa T, Fraser AG. A new echocardiographic method for identifying vortex flow in the left ventricle: numerical validation. *Ultrasound Med Biol*. 2010; 36(5): 772-788.

[8] Shattuck DP, Weinshenker MD. Explososcan: A parallel processing technique for high speed ultrasound imaging with linear phased arrays. *J Acoust Soc Am*. 1984; 75(4): 1272-1282.

[9] Montaldo G, Tanter M, Bercoff J, Benech N, and Fink M. Coherent plane-wave compounding for very high frame rate ultrasonography and transient elastography. *IEEE Trans Ultrason, Ferroelectrics Freq Control*. 2009;56(3): 372-382.

[10] Lai X and Torp H. Effects of the wall filter on the estimation of high blood velocity. *Proc IEEE Ultrason Symp*. 1997; 1263-1267.

[11] Zhang Y, Gao Y, Wang L, Chen J, and Shi X. The removal of wall components in Doppler ultrasound signals by using the empirical mode decomposition algorithm. *IEEE Trans Biomed Engin*. 2007; 54(9): 1631-1642.

[12] Kasai C, Namekawa K, Koyano A, and Omoto R. Real-time two-dimensional blood flow imaging using an autocorrelation technique. *IEEE Trans. Son. Ultrason*. 1985; SU-32: 458-464.

[13] Fox MD. Multiple crossed-beam ultrasound Doppler velocimetry. *IEEE Trans. Son. Ultrason*. 1978; 25(5): 281-286.