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High Frame Rate Ultrasound Flow Vectorgraphy: On the Design of Robust Estimators

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1. Introduction

To be able to unambiguously map multi-directional flow dynamics, it is critical to determine both the flow angle and the velocity magnitude at different spatial positions without uncertainty. Alas, conventional Doppler ultrasound and color flow imaging cannot achieve this purpose because its flow estimation principles are inherently faced with the beam-flow angle dependence problem. In order to use ultrasound to derive velocity estimates that reflect the actual flow characteristics, new flow vector estimation paradigms have been proposed. They generally make use of: (i) multi-angle Doppler; (ii) transverse oscillations; (iii) speckle tracking; (iv) directional cross-correlation. While each approach has its own merit, they are all known to yield erroneous vector estimates under certain scenarios. These frailties are particularly exposed when the imaging frame rate is inadequate.

2. Statement of Contribution

In this research, we have investigated how flow vectors can be estimated consistently at high temporal resolution. We propose to design a least-squares multi-angle Doppler vector estimator that involves a minimum squared error fitting operation. Also, it is our intent to integrate this vector estimator solution with high frame rate imaging methods based on broad-view data acquisition paradigms like plane wave excitation, such that the concept of high frame rate ultrasound flow vectorgraphy can be efficaciously realized.

3. Methods

A. High Frame Rate Data Acqusition

Acquisition of flow vector information at high frame rates is facilitated by performing steered plane wave transmissions. A typical firing sequence comprises a group of M unfocused pulsing events (transmitted in order; see Fig. 1). Between firings, the transmission (Tx) steering angle is changed incrementally so as to cover a span of M angles. For a sequence with M Tx angles and pulse repetition frequency f_{PRF}, the nominal data acquisition frame rate (f_{DAQ}) is in effect equal to f_{PRF}/M. f_{DAQ} can be in the kHz range as long as a limited number of Tx angles is used and f_{PRF} is in tens of kHz.

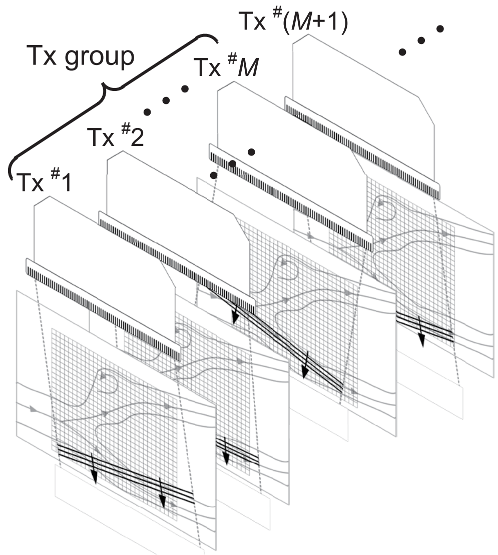

Fig. 1 Illustration of transmission sequence used for high frame rate ultrasound flow vectorgraphy.

B. Beamforming and Slow-Time Sampling

For each Tx event at a given angle, a set of N beamformed data frames is generated in parallel. Each frame is formed with a different receive (Rx) steering angle. The beamformed data value of individual pixel positions in the frame is computed by dynamic receive focusing. With M unique Tx angles, beamformed data frames would be derived for MN combinations of Tx-Rx angle pairs.

To monitor temporal changes in flow dynamics, plane wave transmission and receive beamforming are carried out multiple times for all MN Tx-Rx angle pairs. This process is essentially the same as performing slow-time sampling in conventional Doppler ultrasound. As such, at every pixel position, there would be MN slow-time ensembles (one for each Tx-Rx angle pair) available for flow vector estimation. The slow-time sampling rate is simply equivalent to f_{DAQ}, which in turn is equal to f_{PRF}/M.

C. Robust Flow Vector Estimator

Based on the MN slow-time ensembles for different Tx-Rx angle pairs, flow vector estimation

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is performed independently at all pixel positions and different slow-time instants. After individually applying clutter filtering to each slow-time signal, its corresponding mean frequency is estimated as a function of time. Once the *MN* slow-time frequency shifts are estimated from all Tx-Rx angle pairs, the axial and lateral components of the flow vector are derived using multi-angle Doppler analysis. This process is performed on a timewise basis. At each slow time instant, the computation seeks to solve an over-determined system of equations whereby *MN* mean frequency estimates from all the Tx-Rx angle pairs (at one slow-time instant) are used as input to solve for axial velocity and lateral velocity (i.e. the two unknowns). Specifically, the flow vector can be estimated by carrying out the following matrix operation with each $MN\times 1$ measurement vector $\mathbf{u}$ (consisted of individual frequency shift values):

$$\mathbf{v} = \begin{bmatrix} v_z \\ v_x \end{bmatrix} = (\mathbf{A}^T\mathbf{A})^{-1}\mathbf{A}^T\mathbf{u} \quad (1)$$

where the T superscript denotes a matrix transpose operation. The entity $(\mathbf{A}^T\mathbf{A})^{-1}\mathbf{A}^T$ is well-known in linear algebra as the pseudoinverse of matrix $\mathbf{A}$, which is an $MN\times 2$ constant matrix with the following form based on the trigonometric coefficients of every Tx-Rx angle pair:

$$\mathbf{A} = \begin{bmatrix} \cos\theta_1 + \cos\varphi_1 & \sin\theta_1 + \sin\varphi_1 \\ \vdots & \vdots \\ \cos\theta_M + \cos\varphi_N & \sin\theta_M + \sin\varphi_N \end{bmatrix} \quad (2)$$

4. Experiments

A. Imaging Platform Description

A channel-domain ultrasound research scanner has been used to evaluate the efficacy of the least-squares vector flow estimation technique. The scanner platform is a composite hardware that uses a SonixTouch system for channel-domain Tx operations (to fire steered plane waves), an L14-5 linear array (128 channels), and a SonixDAQ tool for Rx data acquisition. Fast data processing is enabled through our previously developed GPU-based beamformer solutions. Two imaging configurations have been tested: (i) basic dual-angle Doppler with 2 Tx and 1 Rx (–10°, +10°); (ii) generalized multi-angle Doppler with 3 Tx (–10°, 0°, +10°) and 3 Rx (–10°, 0°, +10°). Other key parameters include: 5 MHz frequency; 3-cycle pulse duration; 10 kHz pulse repetition frequency.

B. Flow Well Model

A rotational flow phantom has been designed and fabricated to facilitate performance analysis. It is physically consisted of a cylindrical well (2.5 cm diameter) embedded within a cast of tissue mimicking agar. Placed within the well is blood mimicking fluid (Shelley Medical) with acoustic properties and fluid viscosity matched to those of human blood. To create circulating flow patterns within the well, a magnetic stirrer is placed at the well bottom. Flow vectorgraphy maps are generated with ultrasound data acquired from this setup to analyze estimation consistency at both local scale (whether individual vectors are correct) and global scale (whether rotational flow pattern within the well can be accurately reconstructed).

5. Results and Discussion

Our least-squares flow vector estimator was capable in tracking circulatory flow consistently. Fig. 2a shows representative estimation results for the 3 Tx, 3 Rx imaging configuration. As can be observed, this configuration has depicted rotational flow inside the well more consistently than the primitive 2 Tx, 1 Rx flow vector estimation approach, which gave rise to some spurious vector estimates (Fig. 2b). It is also much more intuitive than conventional color flow imaging (Fig. 2c), which can only depict the circulatory flow as two semicircle zones (red and blue; for flow toward and away from transducer respectively). Corresponding cineloops will be shown at the meeting.

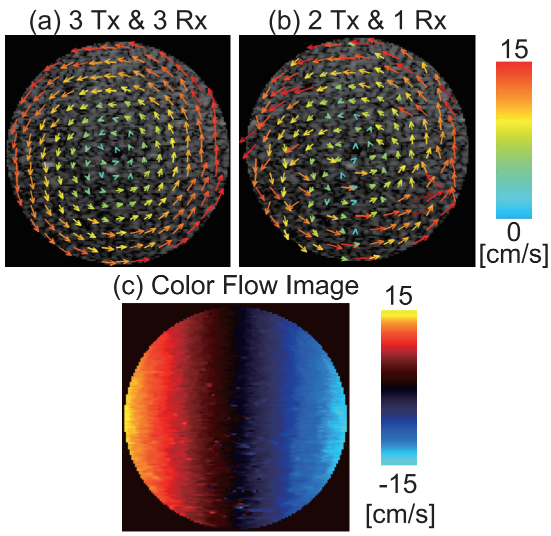

Fig. 2 Flow vectorgraphy maps acquired from a rotational flow phantom using two flow vector estimation configurations: (a) 3 Tx, 3 Rx; (b) 2 Tx, 1 Rx. (c) Color flow image shown for reference.

Acknowledgement

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