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In Vivo 3-D Vector Velocity Estimation with Continuous Data

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Abstract—In this study, a method for estimating 3-D vector velocities at very high frame rate using continuous data acquisition is presented. An emission sequence was designed to acquire real-time continuous data in one plane. The transverse oscillation (TO) method was used to estimate 3-D vector flow in a carotid flow phantom and *in vivo* in the common carotid artery of a healthy 27-year old female. Based on the out-of-plane velocity component during four periodic cycles, estimated flow rates in an experimental setup was $2.96 \text{ ml/s} \pm 0.35 \text{ ml/s}$ compared to the expected $3.06 \text{ ml/s} \pm 0.09 \text{ ml/s}$. In the *in vivo* measurements, three heart cycles acquired at 2.1 kHz showed peak out-of-plane velocities of 83 cm/s, 87 cm/s and 90 cm/s in agreement with the 92 cm/s found with spectral Doppler. Mean flow rate was estimated to 257 ml/min. The results demonstrate that accurate real-time 3-D vector velocities can be obtained using the TO method, which can be used to improve operator-independence when examining blood flow *in vivo*, thereby increasing accuracy and consistency.

I. INTRODUCTION

The cardiovascular blood flow is very complex as it propagates in all three dimension, where short lived vortices can arise and vanish within 100-200 ms [1]. So, to provide the true dynamics of the blood flow, 3-D vector velocities at a high frame rate is therefore necessary. However, current methods for velocity estimations are mainly restricted to 1-D or 2-D methods.

For estimating 2-D vector flow various methods have been proposed, e.g., speckle tracking [2], directional beamforming [3], plane wave emissions as suggested in [4] and transverse oscillation (TO) [5]. 2-D vector flow gives a more realistic picture of the actual flow, but does not provide information about the out-of-plane velocity component.

For realtime out-of-plane velocity estimation, a 2-D transducer is needed. A 2-D piezoelectric transducer was used in previous work, which showed that all three velocity components can be obtained for two crossing planes [6] using a 3-D implementation of the TO method [7] [8] and Provost *et al* with a similar transducer, estimated the axial velocities in a full volume using Doppler techniques [9].

One method to increase the frame rate significant, is by designing an emission sequence for continuous data acquisition. Continuous data acquisition offers several interesting options; for instance, the frame rates for velocity estimates can be increased to the kHz range, which helps to capture complex flow patterns. Second, adaptive velocity estimation algorithms can be applied to create a more complex and dynamic echo

cancelling or increase the detectable velocity range under the rapid changing conditions during a cardiac cycle.

This paper presents the results from 3-D vector velocity estimation using continuous data in both an experimental setup and for an *in vivo* measurement. The out-of-plane velocity components is in both cases used to derive the flow rates. With the applied emissions sequence, ultrafast frame rates up to 2.1 kHz was obtained. In combination with the TO method, both slow flow in the end-diastole and fast flow in the peak-systole could be detected.

II. MATERIALS & METHODS

This section describes the continuous 3-D method. It introduces the equipment, emission sequence, data processing, and theoretical properties of the TO velocity estimator. The basic principles of the TO method is described in [5][10].

A. Emission sequence

In Fig. 1 an illustration is seen of how the applied sequence was designed to yield continuous data. The idea is that the duration between emissions of two identical flow lines F_i should be the same at all times. This could for instance be achieved by emitting the flow lines F_1 - F_5 consecutively, followed by emitting the B-mode line B_1 . Next, the flow lines F_1 - F_5 would be emitted again, followed by the B-mode line B_2 etc. When the desired amount of B-mode lines B_n has been reached, the sequence repeats itself from the beginning. The number of unique B-mode emission B_N was set to 64 in all measurements. A schematic representation of an emission sequence is

$$\begin{array}{cccccccc} F_1 \rightarrow F_2 \rightarrow F_3 \rightarrow F_4 \rightarrow F_5 \rightarrow B_1 \rightarrow & & & & & & & \\ F_1 \rightarrow F_2 \rightarrow F_3 \rightarrow F_4 \rightarrow F_5 \rightarrow B_2 \rightarrow & & & & & & & \\ \vdots & \vdots & \vdots & \vdots & \vdots & \vdots & & \\ F_1 \rightarrow F_2 \rightarrow F_3 \rightarrow F_4 \rightarrow F_5 \rightarrow B_N \rightarrow & & & & & & & \end{array}$$

This would produce continuous flow data and one or multiple continuous B-modes images if desired.

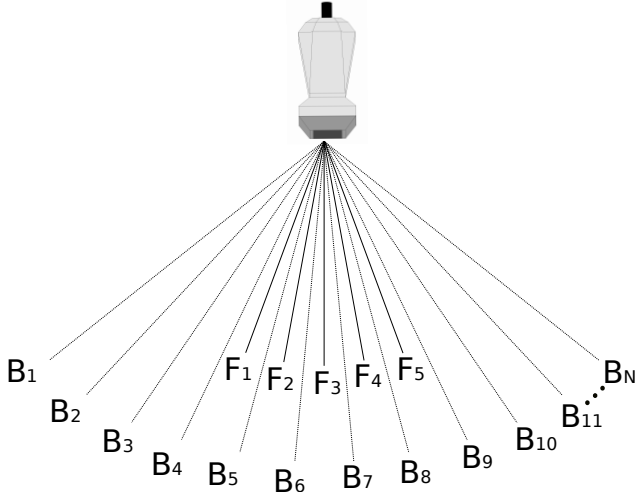


Fig. 1. Illustration of how an emission sequence for continuous data acquisition could be constructed. First, the flow lines from F_1 to F_5 are emitted followed by a B-mode line B_1 . The next cycle contains again all the flow lines from F_1 to F_5 and is ended with the B-mode line B_2 . This continues until B_N is reached, and the first cycle can be repeated again. The angles spanned are only for illustrative purpose and not to scale.

B. Velocity estimators

The axial velocity estimates were based on the autocorrelation approach [15], and the two transverse velocity components were found by using the TO method [5][10].

For each transverse velocity estimate two TO beams were beamformed along the lines separated spatially by their respective transverse wavelengths $\lambda_x/4$ or $\lambda_y/4$, thereby generating two fields phase-shifted by 90° . The two transverse wavelengths are theoretically given by

$$\begin{aligned}\lambda_x(z) &= 2\lambda_z \frac{z}{d_x} \\ \lambda_y(z) &= 2\lambda_z \frac{z}{d_y},\end{aligned}\quad (1)$$

where λ_z is the wavelength of the emitted pulse, z is the axial depth of the beamformed RF-line and d_x , d_y are the distances between the two peaks in the receive apodization in the x - and y -direction. The apodization used in receive was in both directions two rect profiles spanning 8 elements with a spacing of 16 elements between. From [10], the maximum transverse velocities that could be estimated without reaching the aliasing limit were

$$\begin{aligned}v_{x_{max}} &= \frac{\lambda_x}{4k} f_{prf} \\ v_{y_{max}} &= \frac{\lambda_y}{4k} f_{prf},\end{aligned}\quad (2)$$

where k is the lag used in the autocorrelation and f_{prf} , is the effective pulse repetition frequency between two emissions of the same flow line.

C. Measurement setup

The experimental ultrasound scanner SARUS [11] with 1024 channels in receive and transmit was used along with a Vermon 3.5 MHz, 0.3 mm pitch, 0.278 mm height 32×32 element 2-D phased array transducer (Vermon S.A., Tours, France) for data acquisition. See Table I for transducer specifications. The emitted frequency was 3.0 MHz, to avoid grating-lobes and reduce transducer heating. Data was sampled from all 1024 channels and stored for offline processing on a Linux cluster.

The employed 2-D matrix probe contains, due to construction issues, three inactive rows in the x -direction. The dimensions of the actual transducer is therefore 35×32 with the active aperture being 32×32 .

D. Setup for flow pump measurements

For the phantom measurements a predefined pulsating carotid flow profile was generated with a flow pump (CompuFlow 1000 System, Shelley Medical Imaging Technologies, Ontario, Canada) containing blood mimicking fluid with backscattering coefficient equivalent to blood cells. With the included software CompuFlow 1000 System, a carotid flow profile was generated with a cycle time of 0.84 s and a flow rate of $3.06 \text{ ml/s} \pm 0.09 \text{ ml/s}$. A blood mimicking fluid was pumped through a carotid phantom (Danish Phantom Design, Frederikssund, Denmark) with a diameter of 0.8 cm located at a depth of 2.7 cm. Properties for the sequence used for the experimental setup can be seen in Table I.

E. Clinical setup

Intensity measurements to ensure compliance with the FDA limits, were performed for the *in vivo* sequence. The derated mechanical index (MI) was 1.29 and the pulse repetition frequency f_{prf} was scaled to 12.6 kHz to obtain $I_{spta.3} = 720 \text{ mW/cm}^2$ in accordance with FDA limits. The *in vivo* measurement was performed on a healthy 27-year old female who had been resting for 15 minutes before measurements were conducted to ensure a steady flow. The scans were performed by an experienced radiologist. Properties for the *in vivo* sequence are summarized in Table I as well.

III. DATA PROCESSING

The stored data was processed offline. The raw RF data was match filtered and Hilbert transformed before the IQ data was beamformed with the Beamformation Toolbox 3 [12]. In this part, the three velocity components were decoupled, such that one line was beamformed for the axial velocity estimation and two dedicated lines were beamformed for each of the two transverse velocity estimates. In total five unique beamformed lines were used to estimate the 3-D velocity vector for each flow line. Due to the asymmetric geometry of the transducer, two different transverse wavelengths λ_x and λ_y were used for the velocity estimator. The applied transverse wavelengths were $\lambda_x = 3.42 \text{ mm}$ and $\lambda_y = 3.90 \text{ mm}$ for the experimental setup and $\lambda_x = 1.98 \text{ mm}$ and $\lambda_y = 2.18 \text{ mm}$ for the *in vivo* measurements at the depth of 2.7 cm and 1.5 cm respective. For a more extensive description of the employed 3-D TO method, see previous work [13],[7],[8]. Echo cancellation of the beamformed data was subsequently performed with a low frequency Doppler filter algorithm [14].

TABLE I. TRANSDUCER AND EMISSIONS SEQUENCE SETUP

Transducer		Flow Parameters		
Parameter	Value	Parameter	Phantom	<i>In vivo</i>
Transducer type	2-D phased array	Excitation signal	8 cycles Hanning weighted	
No. of elements in x	35 (3 inactive rows)	Frequency	3.0 MHz	3.0 MHz
No. of elements in y	32	Number of flow lines F_N	11	5
Pitch in x and y	0.3 mm	f_{prf}	7.5 kHz	12.6 kHz
Height	0.278 mm	Steering angles	$[-12^\circ:12^\circ]$	$[-10^\circ:10^\circ]$
Kerf	0.022 mm	Δ line separation	2.4°	5°
Sampling frequency	70 MHz	Receive apodization	Rect	Rect
Decimation factor	2	No. of elements between TO apodization peaks	24	24
Center frequency	3.5 MHz	No. of elements in TO apodization	8	8

IV. RESULTS

In this section, results from the two conducted experiments are presented. First, results from the flow pump experiment are presented, and thereafter results from the *in vivo* experiment.

A. Flow pump measurements

From the experimental setup with the flow pump, four seconds of continuous data at 625 Hz was acquired corresponding to 4.8 cycles. Based on a B-mode image showing the cross-sectional view of the vessel, a mask was manually drawn to segment out an area used for velocity estimates. Out-of-plane velocities were calculated in the selected area and the flow rate through it was estimated. The calculated flow rates are seen in Fig. 2. From the known cycle time of 0.84 s, the data set was divided into four phase coherent cycles, and the mean and standard deviation (SD) were calculated. The average flow rate of the profile in Fig. 3 was 2.96 ± 0.35 ml/s compared with the theoretical average flow rate of $3.06 \text{ ml/s} \pm 0.09 \text{ ml/s}$.

B. In vivo measurements

From the 3.07 seconds of continuous data obtained at 2.1 kHz, three complete heart cycles could be identified. As expected the highest peak velocities were found for the center flow line, which propagated closest to the center of the vessel. Peak out-of-plane velocities in each cycle were 83 cm/s ($t = 0.69$ s), 87 cm/s and 90 cm/s ($t = 2.55$ s) based on an M-mode view. Based on the average time between these peak velocities, a heart cycle time of 0.93 s giving a heart rate of 64 beats/min was found. This heart cycle time was used in future calculations.

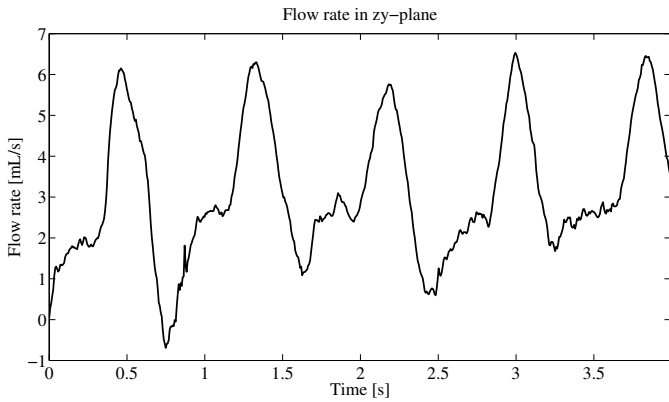


Fig. 2. Flow rates estimated from the out-of-plane velocity component in the experimental setup.

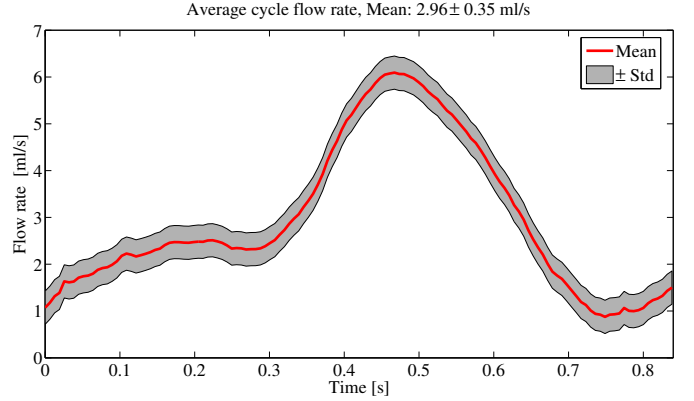


Fig. 3. Mean flow rate averaged over four complete cycles (red line) and one standard deviation (grey area). Mean average flow rate obtained in the phantom experiment was 2.96 ± 0.35 ml/s.

The discriminated velocity estimates for all three components were then scan converted and interpolated yielding a complete 3-D vector velocity map at each time instance. The data set was divided into three subsets each of length 0.93 s. The mean 3-D vector velocities during one cycle was then found by calculating the average velocity estimate for all three velocity components across the three sub-samples. In Fig. 4 3-D vector velocities averaged over three heart cycles at the identified end-diastole are presented. The magnitude of the velocity vector $|\mathbf{v}| = \sqrt{v_x^2 + v_y^2 + v_z^2}$ was found to be 17.0 cm/s in the center of the vessel. The highest center velocity for $|\mathbf{v}| = 81.6$ cm/s was identified as the peak-systole. Spatial variance in peak velocity between each heart cycle and a slightly variance in the heart cycle time results in an average $|\mathbf{v}|$ during peak systole that is lower than the peak v_x velocities during each of the three heart cycles. The 3-D vector velocity map is seen in Fig. 4. Based on the cross-sectional area and the out-of-plane velocity component, the estimated mean average flow rate was 257 ml/min. In the literature the corresponding flow rate from the carotid artery found with spectral Doppler for a healthy person is 334 ml/min [16].

V. DISCUSSION & CONCLUSION

The presented results showed that 3-D vector velocities at very high frame could be obtained when acquiring continuous data. In combination with the TO method, both the slow and fast flow could be detected in pulsating flow setups. The TO method was applicable for estimating accurate flow rates in an experimental setup with a flow pump generating a carotid flow profile.

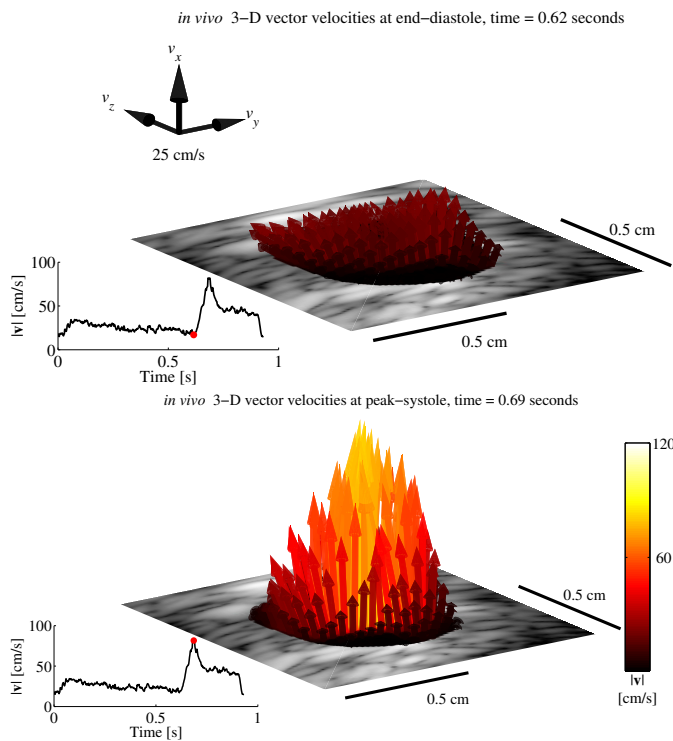


Fig. 4. 3-D vector velocities at the end-diastole, $|\mathbf{v}| = 17.0$ cm/s. 3-D vector velocities at the peak-systole, $|\mathbf{v}| = 81.6$ cm/s. 3-D vector velocities averaged over three complete and consecutive heart cycles. The bottom-left graph shows the magnitude of \mathbf{v} in the center of the vessel at the time instance of the cycle indicated by the red dot. The colored arrows depicts the direction of the flow and its magnitude. The velocity arrows are plotted on top of a B-mode image and only values within the discriminator are shown. Mean flow rate over the averaged cycle was 257 ml/min.

The presented method offers a unique possibility to study complex flow pattern in 3-D, even when slow and fast flow is present. Furthermore, accurate angle-independent flow rates and peak velocity estimates can be derived from the 3-D vector flow. However, further validation of the method against e.g. similar MRI estimates are needed to clarify the accuracy and potential for clinical use.

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