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Transverse oscillation vector flow imaging for transthoracic echocardiography

David Pierson Bradway^a, Kristoffer Lindskov Hansen^b, Michael Bachmann Nielsen^b and Jørgen Arendt Jensen^a

^aCenter for Fast Ultrasound Imaging, Department of Electrical Engineering, Technical University of Denmark, DK-2800 Lyngby, Denmark;

^bDepartment of Radiology, Rigshospitalet, Copenhagen University Hospital, DK-2100 Copenhagen, Denmark;

ABSTRACT

This work presents the development and first results of *in vivo* transthoracic cardiac imaging using an implementation of Vector Flow Imaging (VFI) via the Transverse Oscillation (TO) method on a phased-array transducer. Optimal selection of the lateral wavelength of the transversely-oscillating receive field is described, and results from Field II simulations are presented. Measurements are made using the SARUS experimental ultrasound scanner driving an intercostal phased-array probe. The acquisition sequence was composed of interleaved frames of 68-line B-mode and 17-direction, 32-shot vector velocity flow images. A flow pump was programmed for constant flow for *in vitro* acquisitions at varying depths in a tissue-mimicking fluid. Additionally, mitral, aortic, and tricuspid valves of two healthy volunteers were scanned from intercostal acoustic windows. The acquired RF data were beamformed via the TO method, and fourth-order estimators were employed for the velocity estimation. The resulting images were compared with those from conventional spectral Doppler and color flow mapping sequences. VFI is shown to be a clinically-feasible tool, which enables new flexibility for choosing acoustic windows, visualizing turbulent flow patterns, and measuring velocities.

Keywords: Medical Ultrasound, Vector Flow Imaging, Cardiac Imaging, Blood Flow, Transverse Oscillation

1. INTRODUCTION

Velocity estimation of cardiovascular blood flow is an important clinical tool. Conventional color flow mapping (CFM) based on the autocorrelation approach¹ allows the display of the 1-D axial velocities in a 2-D region of interest (ROI). This single component of velocity along the ultrasound beam does not provide a full representation of complex *in vivo* flow. With the introduction of 2-D and 3-D vector velocity algorithms, like the Transverse Oscillation (TO) method suggested by Jensen, Munk and Anderson,^{2,3} velocity direction and magnitude can be more accurately estimated without the need for angle correction or manual intervention.

A recent commercial implementation of the TO method on the ProFocus Ultra View scanner (BK Medical, Herlev, Denmark) runs in real-time for clinical vascular imaging on a linear array. Those tools have been utilized for studies measuring volume flow in arteriovenous fistulas⁴ and visualizing complex flow phenomena during intraoperative US examinations of the heart and aorta,⁵ where peak systolic velocity measurements were compared to conventional spectral Doppler velocities.

Expanding from the initial implementation on a linear array, recently a phased-array transducer was used for measurements in a flow rig,⁶ and a convex array was used for *in vivo* VFI and spectral measurements of flow in the vessels of the liver.^{7,8} This present work is the first known demonstration of *in vivo* cardiac VFI using the TO method from a noninvasive, transthoracic approach.

In this paper, the theory and optimization of TO fields are described in Section 2. Section 3 describes the methods and materials used for the development and pre-clinical testing. In Section 4, the flow phantom and *in vivo* results are presented and discussed, and Section 5 discusses the conclusions of this work.

Further author information: (Send correspondence to J. A. Jensen) E-mail: jaj@elektro.dtu.dk

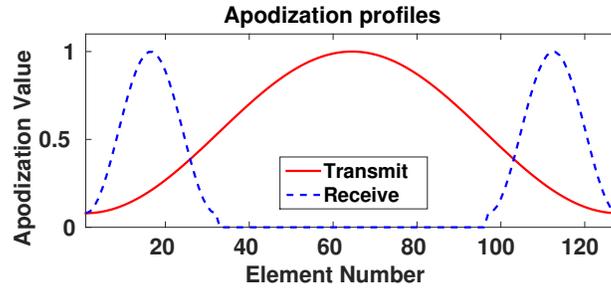


Figure 1: Transmit (red solid) and receive (blue dashed) apodization waveforms used during the simulation.

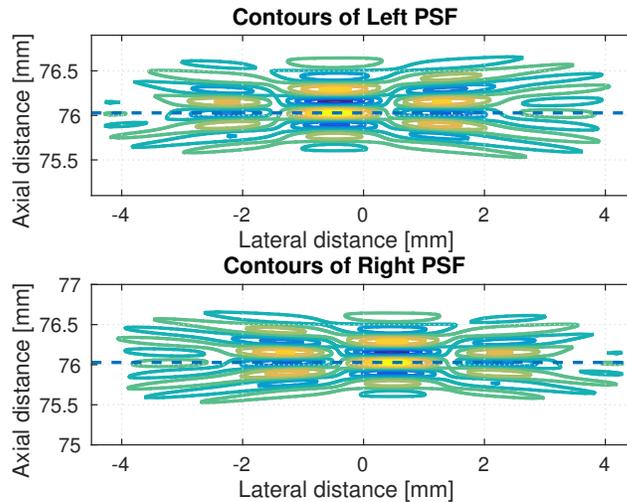


Figure 2: Linear contour plots of the left (top) and right (bottom) point spread functions at a depth of 75 mm.

2. THEORY AND OPTIMIZATION OF TO FIELDS

For estimating velocity components transverse to the direction of propagation of the ultrasound beam, a weakly-focused transmit field is repeatedly emitted at one or more positions in the ROI. A transversely-oscillating receive field can be constructed by applying apodization across the array composed of two Gaussian-weighted peaks, utilizing the Fourier relationship between the apodization function and the lateral response in the far field or at the focus.⁹ The receive apodization function is seen in Fig. 1 with two peaks separated by a distance D generates a laterally-oscillating field at the focus with a wavelength λ_x of²

$$\lambda_x = 2\lambda_z \frac{z}{D}, \quad (1)$$

where λ_z is the axial wavelength and z the depth. The received lateral frequency is then

$$f_x = v_x / \lambda_x, \quad (2)$$

where v_x is the lateral component of the velocity.

Two beams are formed for each lateral position where velocity estimates are to be calculated. The linear contour plots of the left and right point spread functions are seen in Fig. 2. The fields are slightly displaced to either side of the center of the transmitted wave. The angle between focal directions for forming the left and right beams is adjusted so that the lateral Hilbert transform of the left beam is nearly equivalent to the lateral profile of the right beam, as seen in Fig. 3. The imperfect match between the profile of the right beam and the Hilbert transform of the left beam will cause leakage of energy into the negative frequencies in the 2-D Fourier transform

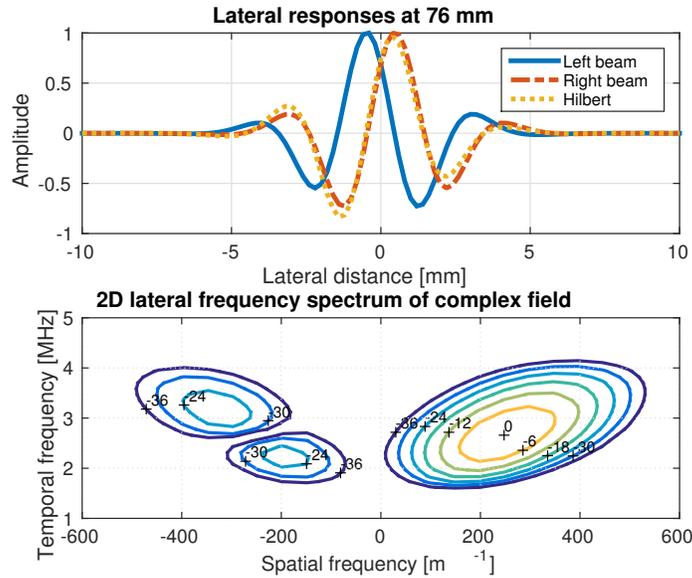


Figure 3: In the top plot, the lateral profiles of the PSFs are plotted for the left beam (solid blue), right beam (dashed red), and Hilbert transform of the left beam (dotted orange). The bottom plot shows the two-dimensional Fourier transform of the complex TO field.

Table 1: Parameters used for the optimization simulations.

Transducer center frequency f_0	3.0 MHz
Number of cycles in emitted pulse	4
Wavelength	0.49 mm
Element pitch	0.22 mm
Kerf	22 μ m
Height of element	15 mm
Fixed elevation focus	85 mm
Number of physical elements of the transducer	128
Electronic focus in transmit	113 mm
Depth for optimizing the transmit focusing	75 mm
F# in transmit for flow	4
Number of receive elements	2 \times 32
PRF for flow	6000 Hz
Emissions for flow	32
Directions for flow	17

of the complex TO field. This unwanted spectral leakage can be used as a criterion for the optimization routine. Samples from the left and right beams are used as in-phase and quadrature pairs in the lateral direction.

The TO field was optimized for imaging at 75 mm by adjusting the beamforming and receive apodization function using the approach described by Jensen.¹⁰ The transducer specifications and imaging parameters used in the optimization study are shown in Table 1.

In Fig. 4, the desired oscillation period is plotted against the achieved simulated lateral wavelength. Setting the lateral wavelength to 4 mm to determine for the receive beams' focus will result in a match between desired and actual wavelength for this focal position.

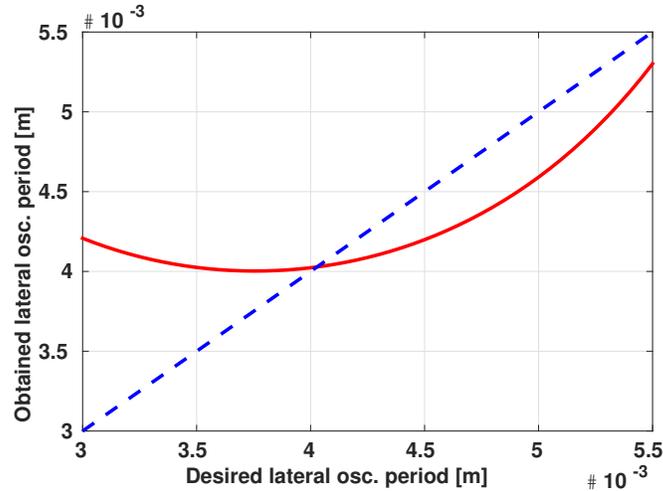


Figure 4: This plots the obtained lateral oscillation period in solid blue as a function of the desired lateral oscillation period in dashed red. Near 4 mm the lines cross, indicating good agreement between desired and achieved lateral period.

This approach is applied to a phased array receive field, and vector flow data are acquired in a flow phantom and *in vivo* as described in the next section.

3. EXPERIMENTAL SETUP

The SARUS experimental ultrasound scanner has been used for acquiring data¹¹ utilizing a 3 MHz STV4 phased-array probe with 128 elements (Sound Technology, Analogic Ultrasound Group, State College, PA). The acquisition sequence consisted of alternating frames of a 68-line B-mode image and a 17-direction, $N_l = 32$ flow shot VFI ensemble. The transmit focus was 56 mm for B-mode and 113 mm for VFI.

3.1 *In vitro*

Preliminary experimental measurements were conducted on a flow phantom containing a 3.6 mm radius tube embedded 10 mm within a tissue-mimicking material. Tissue-mimicking fluid was added atop the phantom and the scanning depth was varied from 45 mm to 85 mm.

3.2 *In vivo*

To ensure FDA compliance for *in vivo* scanning, hydrophone intensity measurements were acquired with an AIMS III scanning system and software (Onda, Sunnyvale, CA). An experienced radiologist acquired *in vivo* measurements of two healthy volunteers' mitral, aortic, and tricuspid valves. A ProFocus scanner was used for ROI guidance and comparison measurements and images. Cine loops and still images were saved for B-Mode, CFM, and spectral Doppler imaging sequences.

3.3 Data processing

Data from the SARUS research scanner were sampled for off-line processing in MATLAB (The MathWorks, Inc., Natick, MA, USA) under Linux. The stored raw RF data were matched-filtered and beamformed using the Beamformation Toolbox 3.¹² Mean stationary echo canceling (clutter filtering) was performed by subtracting the mean value of N_l ensemble lines before velocity estimation. Algorithms outlined in previous work⁶ were used for estimation of axial and transverse flow directions. Estimated velocity components were rotated by the steering angle and were scan converted to a Cartesian grid using bilinear interpolation.

ROIs were marked in the B-Mode by a radiologist, and the corresponding flow estimates were extracted. The magnitude of the flow was found for each point in the ROI, and the mean and standard deviation were calculated.

4. RESULTS AND DISCUSSION

4.1 *In vitro*

Results from the deepest flow phantom acquisition are shown in Fig. 5. The flow within the vessel is masked with a marked ROI. The measured velocity had mean velocity 20.8 ± 8.8 cm/s. The angle of flow in the ROI had a mean angle of 3.3 ± 10.2 degrees. There was some uncertainty in alignment of the transducer fixture with the angle of flow in the rig, but qualitatively there was good agreement with the estimated direction.

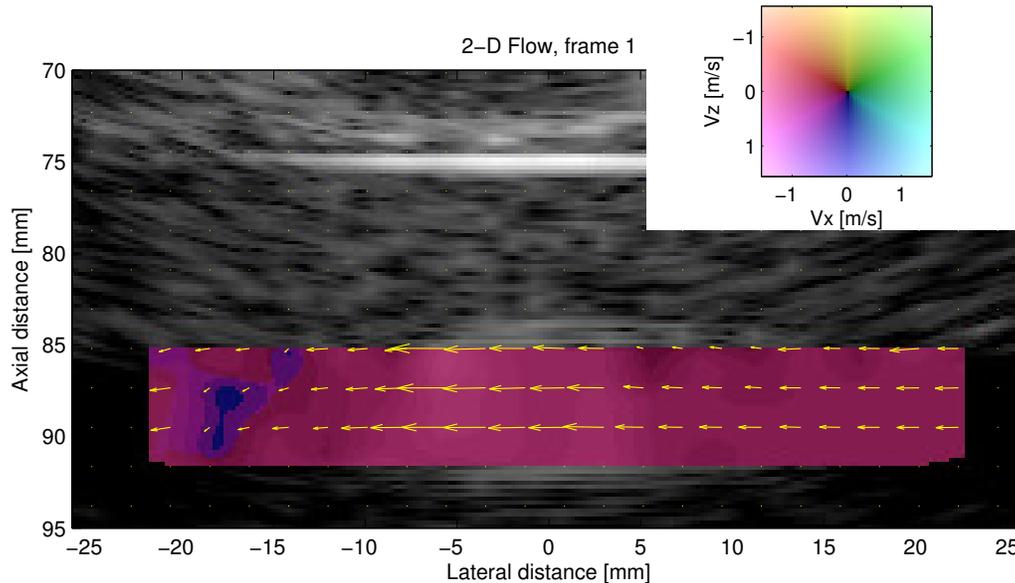


Figure 5: 2-D velocity vectors in a flow rig at 85-92 mm depth, with arrows that indicate the velocity direction and relative magnitude with corresponding 2-D color wheel map

4.2 *In vivo*

The derated intensity values were $MI = 0.8$ and $I_{spta.3} = 66$ mW/cm². Thirty measurements from an *in vivo* VFI acquisition from the second volunteer's tricuspid valve are shown in Fig. 6. The dataset was obtained from right parasternal long axis view. The shape and pattern of the magnitude of the VFI-derived velocity profile appears comparable to the absolute value of the angle-corrected estimates obtained conventional spectral Doppler imaging using the ProFocus scanner, with peak velocity magnitude around 100 cm/s.

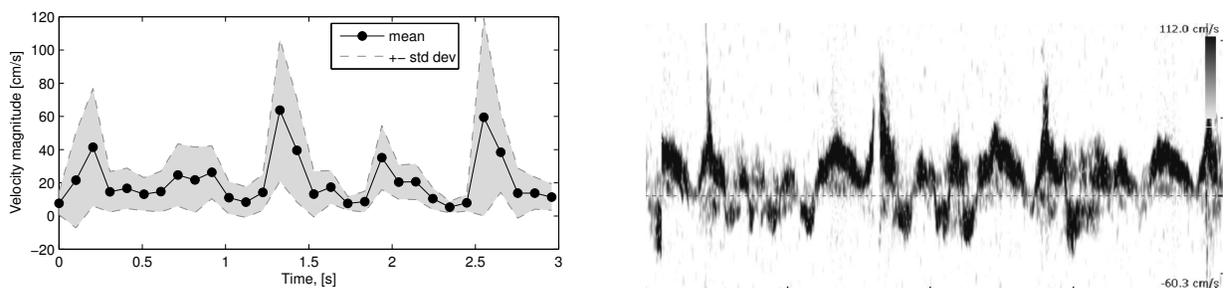


Figure 6: (Left) mean and standard deviation of 2-D VFI-derived velocity magnitude across a 3-sec acquisition and (Right) the corresponding 1-D angle-corrected velocity estimates from conventional spectral Doppler imaging, where velocities range from -60.3 cm/s to +112.0 cm/s and the acquisitions cover nearly 4 seconds.

One image frame from this tricuspid acquisition and another frame from the same volunteer's aortic valve are shown in Fig. 7. The direction and magnitude velocity estimates can be seen in the ROI. This view provides new information into complexity of the flow patterns, which may be useful for judging valve health and turbulence in the sinuses.

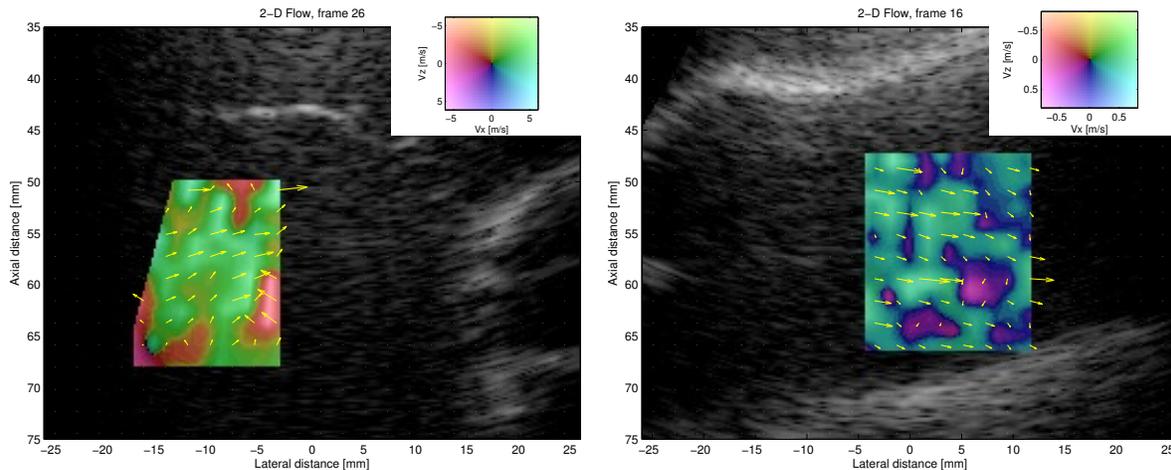


Figure 7: B-Mode image frames with VFI colors and velocity arrows overlaid inside the ROI. (Left) contains a frame from the tricuspid valve acquisition and (Right) shows flow in the aorta, with the aortic valve at the left edge of the image.

In Fig. 8, a conventional color flow mode (CFM) image is shown. Since the main flow is nearly perpendicular to the ultrasound propagation direction, this can be problematic for the precise measurement of flow velocities and interpretation of the CFM image from this acoustic window. The conventional color map switches abruptly from red to blue as the flow direction changes with respect to the steering angle of the transmitted pulses. Trained cardiologists and sonographers are familiar with reading the conventional CFM image color map, but the 2-D color wheel map of VFI provides more information about the true magnitude and direction of complex cardiac flow without relying on operator intervention for angle correction.

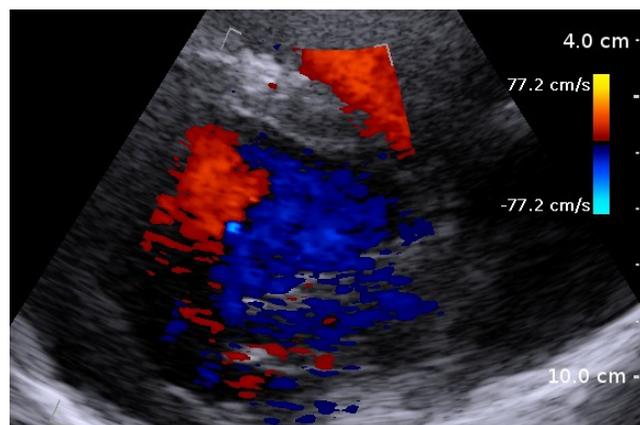


Figure 8: Conventional CFM image with red to blue color mapping scale, obtained from a ProFocus scanner.

5. CONCLUSION AND PERSPECTIVES

This work demonstrated the development of VFI on a phased array. First, optimization simulations were completed and flow measurements made at depths up to 9 cm in a programmable flow rig. Transthoracic *in vivo* cardiac scans were performed for the tricuspid, mitral and aortic valves. Results were compared to conventional spectral Doppler velocity measurements and CFM images.

This initial study had some limitations, including sample size, and a larger pre-clinical study has been completed which will address repeatability and variability. Processing and display were performed off-line in this work, but the TO method and estimation algorithms are suitable for real-time use, as is presently done in the BK Medical linear array VFI mode or by using a graphics processing unit (GPU) in an attached research workstation as shown at this meeting last year.¹³ The frame rate achieved in this work was insufficient to sample the temporal dynamics of the heart around peak systole, so in future work, synthetic aperture flow should be pursued to enable continuous sampling and estimation at a higher frame rate.

As a diagnostic tool, VFI will improve guidance and feedback to cardiologists and researchers through improved visualization of turbulence and the removal of the angle-correction of conventional spectral Doppler estimation.

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