Shunt quantification in congenital heart disease based on two-dimensional speckle tracking

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***Abstract*—In this work we investigated how high frame rate speckle tracking based on plane wave imaging could be used to improve the quantification of peak velocities in shunt flows due to septal defects. Simulated jet flow was used to optimize acquisition and tracking parameters. *In vivo*, a packet based acquisition scheme was used where focused B-mode scans were interleaved high frame rate flow images (100 fps). Results showed that speckle tracking provides calibrated velocities in the shunt flow throughout the cardiac cycle, and improved estimates of peak velocities used for diagnosing shunt severity were acquired.**

I. Introduction

Atrial and ventricular septal defects (ASD/VSD) are the two most common forms of congenital heart defects. In pediatric cardiology, color flow imaging (CFI), continous wave- (CW) and pulsed wave- (PW) Doppler are used to detect and quantify shunt flows. CFI and PW-Doppler are based on pulsed transmission of ultrasound waves, and the velocity along the beam is estimated from the frequency shift between received ultrasound echos. The sampling of the received echoes limits the highest measurable velocities according to the Nyquist sampling theorem and velocities above the Nyquist limit will give aliasing artifacts in the image. CW-Doppler continually transmit and receive ultrasound signals in parallel and measures all velocities along the beam. Since there is no sampling of the received signal in CW-Doppler, even the highest blood velocities can be measured. There is however no way of specifying the location of the different velocities. The fundamental limitations of these methods can result in ambiguous flow images and signal dropouts which may lead to undetected defects and misdiagnosis [1].

One established measure of shunt flow is the Q_p/Q_s-ratio. Q_p/Q_s is the ratio of flow in the pulmonary and systemic circulation, and interventional closure or corrective surgery of the shunt is recommended for patients with a Q_p/Q_s ratio $\geq$ 1.5. The ratio is calculated using the following equation,

$$\frac{Q_p}{Q_s} = \frac{\mathrm{VTI}_{\mathrm{RVOT}} \times \pi \times (\frac{1}{2} D_{\mathrm{RVOT}})^2}{\mathrm{VTI}_{\mathrm{LVOT}} \times \pi \times (\frac{1}{2} D_{\mathrm{LVOT}})^2}, \tag{1}$$

where D_{RVOT} and D_{LVOT} are the measured diameters and $\mathrm{VTI}_{\mathrm{RVOT}}$ and $\mathrm{VTI}_{\mathrm{LVOT}}$ are the estimated velocity-time-integral of the right and left ventricle outflow tract. With todays clinical methods, the measurements are uncertain and the Q_p/Q_s-ratio for shunt quantification is not widely used. Instead the B-mode and CFI images are studied and the peak velocity and diameter of the shunt, together with the general condition of the newborn typically lead to the diagnosis.

Recent development in ultrasound imaging technology provides a substantially higher acquisition rate which gives new possibilities in angle-independent flow estimation. High frame rate speckle tracking for two-dimensional blood flow estimation overcomes the angle-dependency and can track velocities beyond the Nyquist limit. In this work we investigate the use of speckle tracking to quantify shunt flow velocities.

II. Methods

By utilizing plane wave imaging and parallel receive beamforming, instantaneous flow images at high frame rates *and* high Doppler ensemble sizes can be achieved. This setup has previously been shown beneficial for blood speckle tracking [3], a method that has yet to reach clinical practice due to a lack of robustness. In speckle tracking, a kernel region is identified in a first image acquisition and then the best match of the kernel is searched for in a subsequent acquisition, determining the displacement of the kernel. In this work the best match was found using the sum-of-squared-differences (SSD) algorithm. Instantaneous images and high frame rates minimize speckle decorrelation and improve speckle tracking results. In addition, temporal and spatial averaging is used to increase the robustness of the estimates. As in color flow imaging, clutter filtering was used to remove the strong tissue signal and make the blood speckle visible before speckle tracking. In this work a fourth order FIR filter with cut-off at $0.33 \times V_{\mathrm{Nyquist}}$ has been used.

A. Ultrasound simulations

A computational fluid dynamics (CFD) model of shunt flow was made using Fluent 13.0 (Ansys Inc.). The simulations were done with varying shunt sizes and pressure conditions, resulting in jet flows of varying size and velocities. The resulting simulated flow fields were used as input to ultrasound simulations with Field II [2] (see Figure (1)). The simulation results were used to optimize the ultrasound acquisition and speckle tracking setup for *in vivo* evaluation.

B. In vivo

A SonixMDP ultrasound scanner (Ultrasonix, Richmond, BC, Canada) with a 4-9 MHz linear transducer and a Sonix

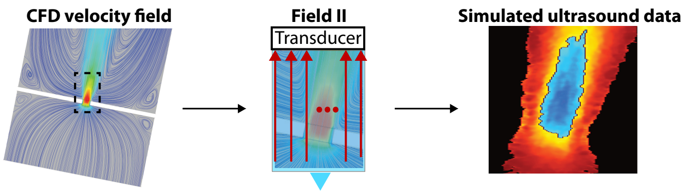


Fig. 1: Simulation setup: the CFD velocity field is used as input to the ultrasound simulation.

TABLE I: Image acquisition parameters

Probe	UX L9-4/38
Probe type	128 element linear array
Pulse center frequency	5 MHz
Transmit aperture	3.89 cm (full aperture)
Receive F-number	1.1
PRF	8 kHz
Ensemble size	32
Frame rate	100 Hz

DAQ for channel data acquisition was used for *in vivo* recordings. A packet based acquisition scheme was implemented on the scanner to get both sufficiently good B-mode images and high frame rate flow images. Unfocused transmit beams with a PRF of 8kHz was used for the flow images with an ensemble size of 32. Full parallel receive beamforming was utilized to achieve frame rates of 100 fps. The B-mode sequence was interleaved the flow images with two interleave group needed to cover the whole image. Focused transmit beams and a PRF of 12kHz was used for the B-mode images. All the channel data was acquired and the beamforming was done offline after the ultrasound recordings.

III. Results and discussion

A. Simulations

The speckle tracking velocity estimates from the simulated ultrasound data were compared with the CFD ground truth velocity field. A color flow image of simulated flow through a shunt of diameter 6 mm is shown in Figure 2 together with the speckle tracking and ground truth velocity fields. The white arrows represent the ground truth velocity field from CFD and the black arrows are the estimated velocities from speckle tracking. Figure 3 shows the velocity profile from speckle tracking compared with the ground truth velocity profile in the area of the shunt. In this area the speckle tracking results show a good agreement with the ground truth.

However, in the areas with very large spatial velocity gradients, the speckle tracking estimates deviates from the ground truth; the estimated velocities are drawn inwards to the center of the jet. This phenomena was investigated further with simulations of point scatterers, which resulted in a possible explanation to the observed deviation. The full aperture plane wave transmissions produce larger edge waves effects than focused transmit beams (see Figure 4a and 4b). When the simulated point scatteres had large velocity differences, the edge waves interfered when one point scatter passed another,

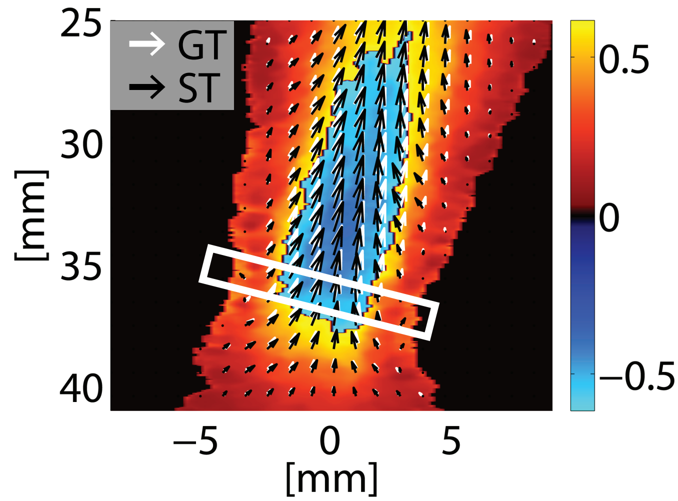


Fig. 2: The speckle tracking and ground truth velocity fields are represented with black and white arrows, respectively, on top of a color flow image.

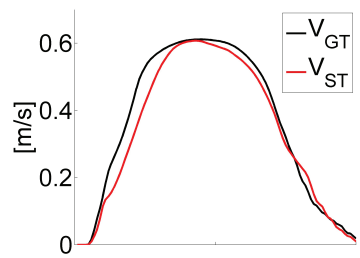
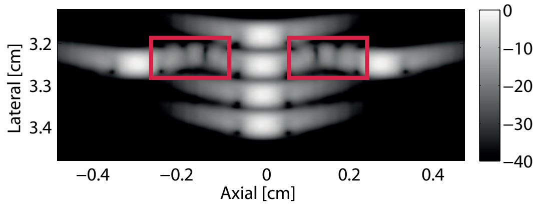


Fig. 3: The speckle tracking and ground truth velocity profiles in the shunt area are compared.

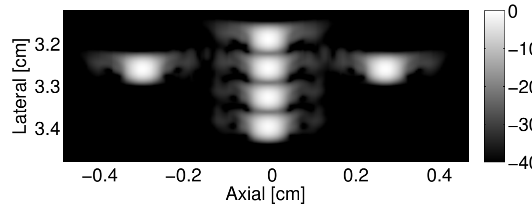
creating an interference pattern with apparent lateral movement in the edge waves. Receive apodization reduces the edge waves, but it reduces the lateral resolution as well. To best solve this trade-off with the ultrasound system available, hamming apodization and a low F-number of 1.1 was used in the *in vivo* beamforming.

B. In vivo

Flow speckle tracking was done for newborns with atrial and ventricular spetal defects. Figure 5 and Figure 6 show the color flow images of patients with a ventricular septal defect and an atrial septal defect, respectively. The speckle tracking estimates are represented by white arrows which indicate the direction and magnitude of the flow. Both speckle tracking and autocorrelation velocity estimates were validated with retrospective PW-Doppler, i.e. the PW spectrums and the speckle tracking and autocorrelation estimates were all based on the same recording. The shunt region was manually tracked in both pasients, so that the sample volume for PW-Doppler could follow the center of the shunt throughout the cardiac cycle. In Figure 7, the PW spectrum of the VSD and the ASD is shown together with the velocity estimates. The

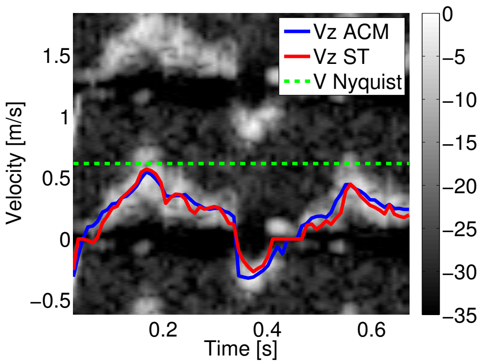


(a) Plane wave transmit beams

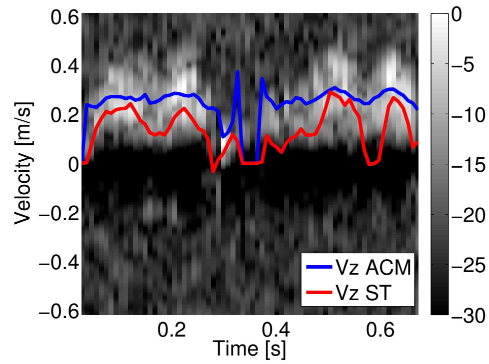


(b) Focused transmit beams.

Fig. 4: Simulated point scatteres with interfering edge waves. The edge waves are much less apparent with focused transmit beams.



(a) Retrospective PW spectrum from the sample volume indicated in Figure 5 shown together with the speckle tracking and autocorrelation estimates. Used for validation of the ST estimates.



(b) Retrospective PW spectrum from the sample volume indicated in Figure 6 shown together with the speckle tracking and autocorrelation estimates.

Fig. 7: Simulation results

velocity estimates from speckle tracking agreed well with the PW spectrum both for the VSD and the ASD. The velocities in the ASD is lower than the velocities in the VSD, and closer to cut-off velocity of the clutter filter. With a PRF of 8kHz and center frequency of 5Mhz, the Nyquist limit is approximately 0.6 m/s, giving a filter cut-off of $0.33 * \text{vNyquist} \approx 0.2$ m/s. With this clutter filter, overestimation is a problem for the autocorrelation estimates of the ASD velocities, as seen in Figure 7b. Figure 8 shows the shunt area of the VSD with the speckle tracking estimates. The velocity profile in the shunt for the frame with highest velocities is shown in Figure 9 and compared with the auto-correlation estimates. In this case, the autocorrelation estimates are angle-corrected based on the speckle tracking estimates.

The results show that speckle tracking may provide calibrated velocity profiles in the shunt throughout the cardiac cycle and therefore improved estimates of peak velocities as currently used for diagnosing shunt severity may be acquired. The speckle tracking estimates can be used to angle-correct Doppler-based methods, and if combined they may further improve the robustness of the velocity estimates.

IV. Conclusion and further work

High frame rate speckle tracking based on plane wave imaging tracking provides angle-independent two-dimensional velocity estimates which can provide more accurate peak velocity measurements for shunt flow evaluation. Further work will investigate speckle tracking possibilities for improved Q_p/Q_s measurements and direct quantification of shunt flow.

References

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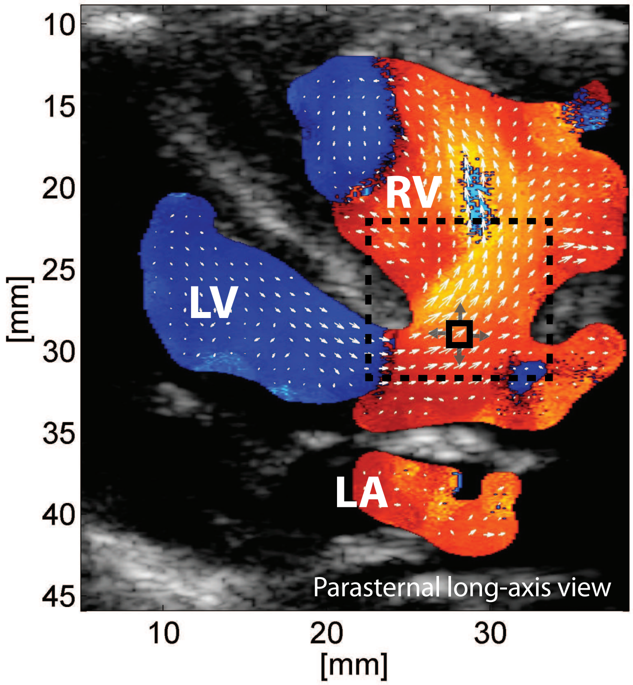

Fig. 5: Patient 1 (8 days, 3340 g): Color flow image and speckle tracking estimates of a ventricular septal defect. The sample volume for the retrospective PW spectrum in Figure 7a is indicated in the shunt region. The dashed rectangle is shown again in Figure 8.

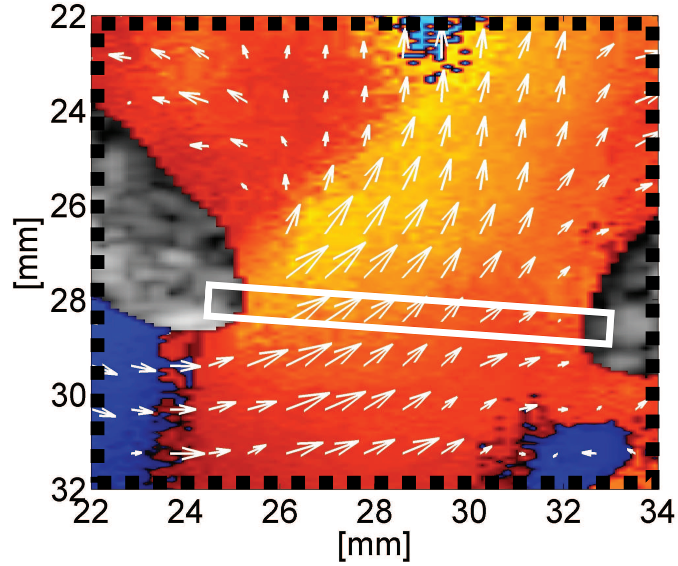

Fig. 8: The shunt region indication with a dashed rectangle in Figure 5. The white arrows are the speckle tracking estimates.

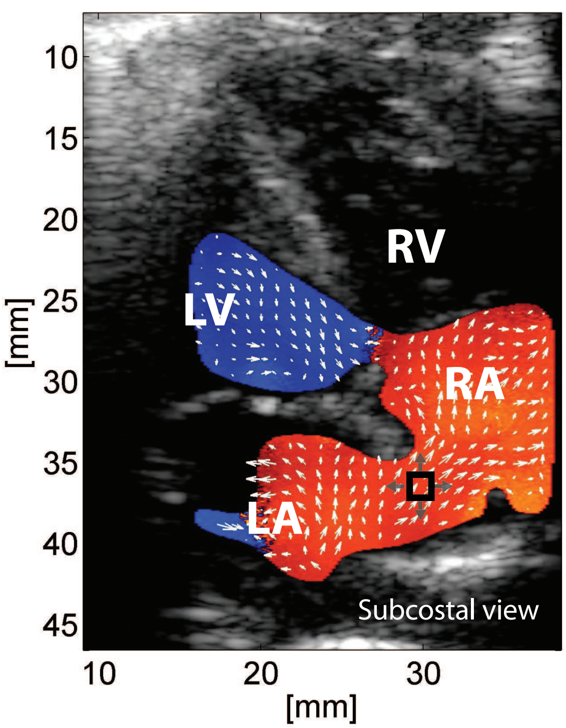

Fig. 6: Patient 2 (25 days, 2165 g): Color flow image and speckle tracking estimates of an atrial septal defect. The sample volume for the retrospective PW spectrum in Figure 7b is indicated in the shunt region.

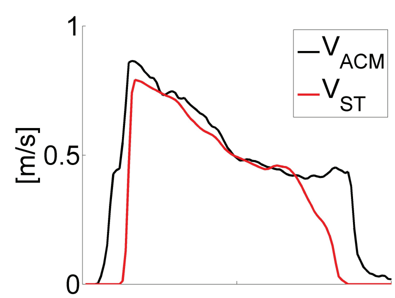

Fig. 9: The estimated velocity profile in the shunt region of the VSD indicated in Figure 8, for both the speckle tracking and the autocorrelation approach. In this case the autocorrelation estimates were angle-corrected based on the speckle tracking results.