Detailed flow visualization in fetal and neonatal hearts using 2-D speckle tracking

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***Abstract*—Two-dimensional blood speckle tracking has shown promise for measuring the complex flow patterns in neonatal hearts when based on linear array and high-frame-rate plane wave imaging. For phased array pediatric imaging, additional challenges emerge due to the reduced lateral bandwidth and increased imaging depth and field-of-view. In this work, a clinically approved setup with pediatric phased array probes and unfocused pulses was used to investigate the potential of blood speckle tracking to acquire 2-D vector velocity maps for neonates, infants and children with congenital heart disease.**

Promising results were observed for depths < 10 cm, where complex cardiac flow patterns could be visualized. However, due to the small aperture available, diffraction effects could be observed. Further, as the depth dependent lateral resolution and loss in signal-to-noise ratio degrades tracking results for increasing depths, a larger feasibility study is needed to establish clinical viability.

Vector velocity maps were also obtained from fetal examinations with the phased array setup as well as with a diverging beam setup on a research scanner, where detailed secondary flows such as the vortex formations in the ventricles of the fetal heart could be observed.

I. Introduction

Ultrasound Doppler imaging is essential in the evaluation of fetal and postnatal circulation. Early detection of flow disturbances is the key to discover abnormalities, but also in the follow up to assess growth, to help the clinicians make a suitable plan for delivery and treatment of the newborn child, and to follow up children with congenital heart disease.

While color flow imaging (CFI) is a highly useful tool to detect and classify congenital heart defects, the images only portray the radial blood velocity component. This angle-dependency and the limited measurable velocity range may result in ambiguous images which require interpretation. To increase the physiological understanding of blood flow in both healthy and diseased hearts we aim to develop and investigate the use of 2-D vector velocity imaging in fetal and pediatric cardiology.

One approach to obtain the 2-D vector velocity field is to utilize blood speckle tracking, where the movement of the blood speckle is tracked from frame to frame, providing an estimate of the blood vector velocities [1]. When combined with high-frame-rate imaging techniques such as plane wave imaging, angle-independent cardiac blood velocity information could be achieved using a linear array probe [2]. For pediatric applications a linear array probe is, however, only feasible when imaging neonates, and for fetal imaging a curvilinear array is preferred. Thus, our aim here is to develop and investigate the use of blood speckle tracking based on phased array and curvilinear imaging.

II. Methods

A. Data acquisition

A real-time imaging setup based on unfocused transmit pulses and 16 parallel receive lines was set up on a modified GE Vivid E9 scanner (GE Vingmed, Horten, Norway). Because of the limitation in the number of real-time parallel receive lines, 3-9 transmit events were needed to cover the region-of-interest (ROI). A duplex setup with separate focused B-mode imaging resulted in an overall frame rate of 40-80 frames-per-second (fps), depending on the packet-size (8-22), image depth (<10cm) and ROI width. IQ-data was beamformed in real-time in the ultrasound scanner and saved for further off-line processing. The acquisition was clinically approved for the GE 9L linear array probe and two phased array pediatric probes (GE 6S and 12S).

In addition, a Verasonics Vantage system (Verasonics, Inc., USA) with a curvilinear array (C5-2v) was set up with a duplex diverging wave scheme. For the B-mode sequence, the image was formed using coherent compounding of 7 diverging waves, whereas for the blood flow sequence, 22 diverging waves formed the color flow ensemble.

B. In vivo study

An on-going feasibility study using unfocused high-frame-rate imaging and blood speckle tracking includes N=45 patients in Trondheim, Norway. The acquisition was first implemented on the GE 9L linear array probe, and was recently expanded with imaging using the pediatric phased arrays (6S or 12S).

The initial recordings using the 6S phased array and the Verasonics curvilinear array for imaging the fetal heart has also been conducted on two pregnant volunteers where the fetus was healthy and developing normally.

Patient safety measurements were conducted for all the sequences and were within the guidelines from the US Food and Drug Administration (FDA). Written informed consent

was obtained from the pregnant women and the parents of the neonates before examination.

C. Blood velocity estimation

Before blood velocity estimation is possible, the IQ-data must be clutter filtered to remove the strong echo from the surrounding tissue. In this work a 4th order FIR filter with cut-off velocity $v_{co} \approx 1/3 \cdot v_{Ny}$ was used. A GPU-optimized 2-D blood speckle tracking algorithm was developed for velocity estimation, running close to real-time. The speckle tracking algorithm determines the displacement of the blood speckle from frame to frame using an initial sum-of-squared-differences (SSD) pattern matching and subsequent parabolic subsample interpolation. To improve the velocity resolution, IQ-data was linearly interpolated prior to tracking.

The tracking algorithm has previously been validated in simulations and a flow phantom setup [2], [3]. To validate the ST velocity estimates *in vivo*, we have compared the results with angle-corrected pulsed wave (PW) and continuous wave (CW) Doppler.

D. Focused vs unfocused imaging

Acquisitions using unfocused imaging benefit from a much higher frame rate than conventional focused imaging, but the loss in signal-to-noise ratio (SNR) and penetration depth can be a limiting factor. With the current limitation on the number of parallel receive lines (16) for the clinical approved setup on the GE E9 scanner, several (3-9) plane wave transmissions were used to cover the desired ROI. While sending only one diverging beam could potentially cover the whole ROI and dramatically increase the frame rate, a significant decrease in SNR is also expected. This trade-off was investigated using a Field II simulation setup [4], [5] to generate relevant beam profiles and inspect the difference in generated pressure, and beam width and uniformity.

III. Results

A. Beam profile comparison

Beam profiles for a phased array pediatric probe for plane and diverging beams, and a beam focused at 10 cm depth are shown in Fig. 1. The full aperture was used on transmission and the same voltage level was given for all cases. The diverging beam had an opening angle of 40 degrees. A non-scanconverted image of the simulated plane wave transmission is shown in the left panel and the lateral beam profiles for the simulated plane, diverging and focused beams at four depths are shown to the right. The depths are indicated with horizontal lines in the left image. The diverging beam covers a larger angle span in depth, but at the cost of lower pressure compared to the plane and focused beams. At 10 cm depth the plane wave pressure is at approximately 6 dB below the focused beam, covering an angle span of about 4 degrees. At this depth, the diverging beam is covering an angle span of 15 degrees, but with 6 to 8 dB below the pressure generated by the plane wave.

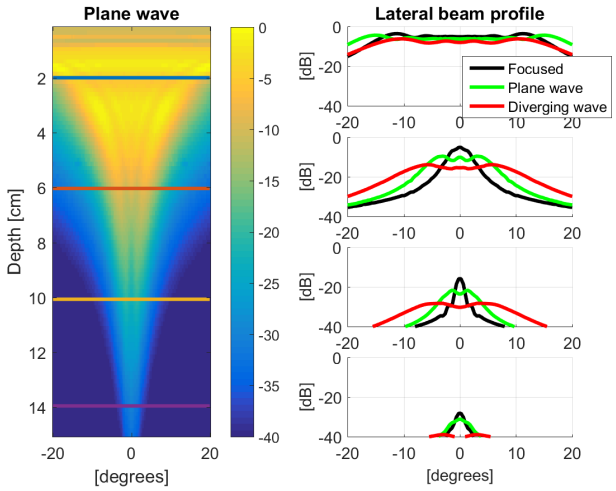

Fig. 1: To the left is an image of the simulated plane wave transmission (non-scanconverted). The four depths for the lateral beam profiles shown to the right are indicated with horizontal lines.

B. Validation of velocity estimates

In Fig. 2 the speckle tracking velocity estimates in a small region-of-interest (ROI) are validated towards CW Doppler from a line going through the given ROI. The ST velocity estimates are averaged in the ROI (yellow square in the image in the left panel) and plotted on top of the angle-corrected CW Doppler trace in the right panel. The velocities in the ROI exceed the Nyquist velocity for a large part of the cardiac cycle and are close to 1 m/s at the highest. As can be observed, the ST velocity estimates have a good correspondence with the CW Doppler trace throughout the cardiac cycle.

C. Newborn results

Fig. 3 shows images acquired with the 9L linear array and the 6S phased array probe of the same 10-days-old newborn with a pulmonary valve stenosis. As can be observed, the flow field in the images are similar and shows the vortex forming in the dilated pulmonary artery distal to the valve.

D. Fetal results

In Fig. 4 a healthy fetus at 24 weeks gestation is imaged with the 6S phased array probe. In the left panel, a B-mode image is annotated with the four chambers of the heart. To the right, the speckle tracking velocity estimates are shown as arrows overlaid a color flow image. The 2-D speckle tracking approach is able to capture the vortex forming in both ventricles.

Fig. 5 shows the first results of the diverging beam acquisition with the Verasonics curvilinear array. The inflow to the ventricles are imaged and a vortex formation in the right ventricle is visualized.

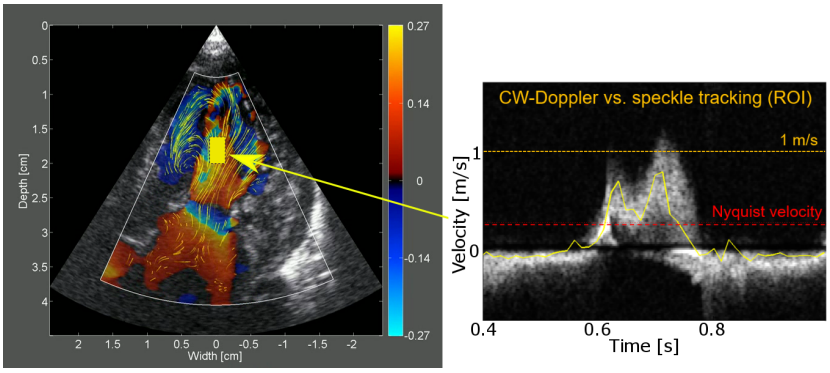

Fig. 2: Validation of the velocity estimates from speckle tracking towards continuous wave (CW) Doppler from a small region-of-interest (ROI).

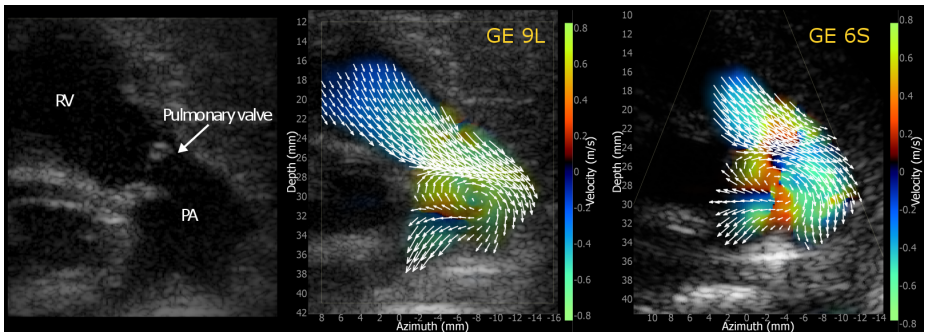

Fig. 3: A 10-days-old newborn with a pulmonary valve stenosis imaged. Left: B-mode image of the right ventricle (RV), pulmonary valve and pulmonary artery (PA). Middle and right: Images acquired with the GE 9L linear array and the GE 6S phased array probe, respectively. The speckle tracking velocity estimates are shown as arrows overlaid the color flow image. A large vortex formation in the pulmonary artery is seen in both recordings.

IV. Discussion

Phased array probes have been tested for the purpose of achieving detailed flow images of neonatal and fetal hearts with unfocused imaging and 2-D blood speckle tracking. We have previously reported that linear arrays can be used to image neonatal hearts as we can image straight through the ribs [6], however, the general pediatric probes are phased array probes and the aim here was to further develop our methods for these probes which can also image older children.

Using plane waves in conjunction with phased arrays have pros and cons. In our clinical approved real-time setup, we are limited to 16 parallel receive lines and thus need several transmissions to cover the desired ROI. Due to diffraction effects, the width of the plane wavefront will be reduced in depth and thus several plane waves would be needed to cover the ROI regardless of the limited number of receive lines. For diverging wave transmissions, a larger ROI could be covered with only one or few transmissions. However, for phased-arrays the lower pressure versus depth as seen in Fig. 1 will result in a reduced SNR and penetration. And unfortunately, it is not straight forward to utilize coherent compounding to retain this SNR due to the high Doppler PRF needed in this clinical context.

The resolution is lower with the general pediatric phased array compared with a linear array due to the smaller aperture and lower frequency range. The aperture results in a depth-dependent lateral resolution, which degrades the lateral tracking quality for larger depths. Despite these drawbacks, our experience and preliminary validation towards spectral Doppler and comparisons with the linear array imaging show promising results. In the patients we have imaged (image depths < 10 cm), the vortex formation in the cardiac chambers could be mapped, in addition to a qualitative impression of the primary and secondary blood flow patterns. However, small shunts were not as easily mapped as with color flow imaging, partly due to the spatial smoothing required to lower the variance of the lateral velocity estimate.

In the fetal example in Fig. 4, the image depth was 6-7 cm. Imaging small chambers approximately 1 cm wide at this depth with a phased-array is challenging, but the large features of inflow and vortex formation in the ventricles were well visualized in this example.

A larger field-of-view and improved lateral resolution are obtained with the curvilinear array, which is the routine probe for fetal imaging. In the example shown in Fig. 5, the fetus was easy to image, with the heart positioned at around 4 cm depth and limited fetal movement. All four chambers were clearly visible and blood SNR was sufficient for velocity estimation

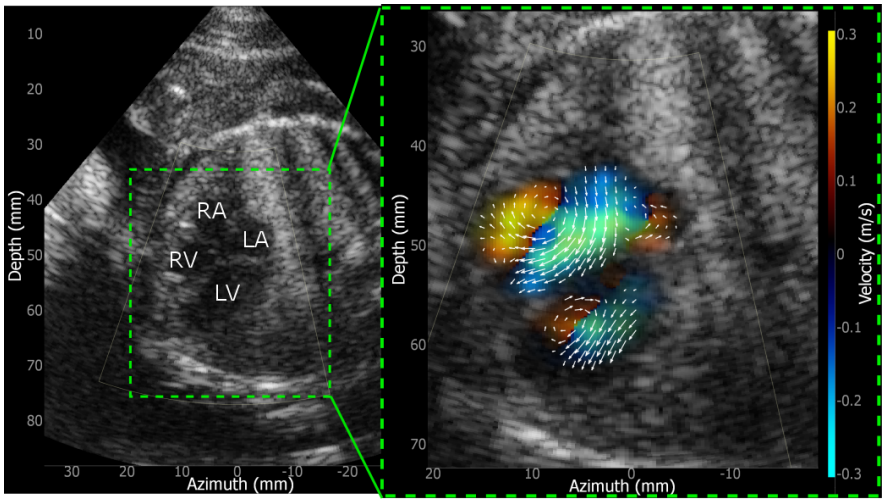

Fig. 4: Fetus at 24 weeks gestation imaged with the GE 6S phased array. A vortex is forming in both the right ventricle (RV) and the left ventricle (LV).

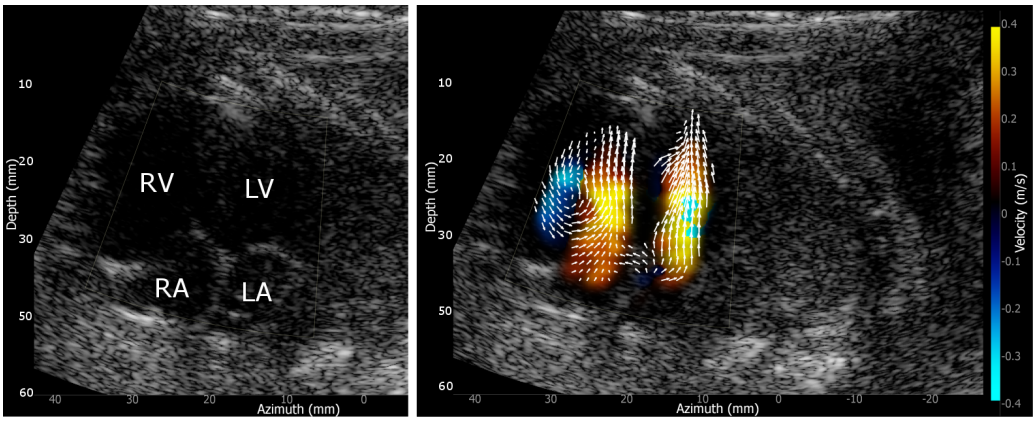

Fig. 5: Fetal imaging at 26 weeks gestation with the Verasonics curvilinear array. ST estimates represented with arrows are overlaid the color flow images. Inflow to the left and right ventricle (LV and RV) and vortex formation on the right side.

using speckle tracking. For fetal imaging the required imaging depth can vary substantially according to fetal age and movement, as well as due to variations in the body mass index (BMI) of the pregnant. A feasibility study using diverging waves and the Verasonics curvilinear array for fetal imaging will be started later this year.

V. Conclusion

Pediatric phased array probes utilizing unfocused pulses and blood speckle tracking was used to image complex flow patterns in neonates and children with congenital heart disease. While promising results have been demonstrated for image depths < 10 cm, limitations in penetration is expected compared to conventional color flow imaging, and a feasibility study will be needed to map the true clinical potential of this approach.

Vector velocity maps were also obtain for two fetal examinations using high-frame-rate speckle tracking, and a planned feasibility study will further investigate the method's potential for fetal imaging.

References

[1] L. N. Bohs, B. J. Geiman, M. E. Anderson, S. C. Gebhart, and G. E. Trahey, "Speckle tracking for multi-dimensional flow estimation," *Ultrasonics*, vol. 38, no. 1, pp. 369–375, 2000.

[2] S. Fadnes, S. Bjærum, H. Torp, and L. Lovstakken, "Clutter filtering influence on blood velocity estimation using speckle tracking," *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*, vol. 62, no. 12, 2015.

[3] J. Van Cauwenberge, L. Lovstakken, S. Fadnes, A. Rodriguez-Molares, J. Vierendeels, P. Segers, and A. Swillens, "Assessing the performance of ultrafast vector flow imaging in the neonatal heart via multiphysics modeling and in-vitro experiments," *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*, vol. 3010, no. c, pp. 1–1, 2016.

[4] J. A. Jensen and N. B. Svendsen, "Calculation of Pressure Fields from Arbitrarily Shaped, Apodized, and Excited Ultrasound Transducers," *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*, vol. 39, no. 2, pp. 262–267, 1992.

[5] J. A. Jensen, "FIELD: A Program for Simulating Ultrasound Systems," *Medical and Biological Engineering and Computing*, vol. 34, no. SUPPL. 1, pp. 351–352, 1996.

[6] S. Fadnes, S. Nyrnes, H. Torp, and L. Lovstakken, "Shunt flow evaluation in congenital heart disease based on two-dimensional speckle tracking," *Ultrasound in Medicine and Biology*, vol. 40, no. 10, 2014.