

Cardiac flow mapping using high frame rate diverging wave contrast enhanced ultrasound and image tracking

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Abstract—Contrast echocardiography (CE) ultrasound with microbubble contrast agents have significantly advanced our capability in assessing cardiac function. However in conventional CE techniques with line by line scanning, the frame rate is limited to tens of frames per second, making it difficult to track the fast flow within cardiac chamber. Recent research in high frame-rate (HFR) ultrasound have shown significant improvement of the frame rate in non-contrast cardiac imaging. In this work we show the feasibility of microbubbles flow tracking in HFR CE acquisition in vivo with a high temporal resolution and low MI as well as the detection of vortex near the valves during filling phases agreeing with previous study.

Keywords—High frame rate, ultrafast contrast enhanced ultrasound, Cardiac flow quantification, Image/speckle tracking, echocardiography, Ultrasound Imaging Velocimetry (UIV), Particle image velocimetry (PIV)

I. INTRODUCTION

Ultrasound contrast agents (UCA), or microbubbles, for contrast enhanced ultrasound (CEUS) ultrasound imaging is revolutionising the role of medical ultrasound in clinical practice [1]. These bubbles are highly sensitive to ultrasound, and once introduced into the blood stream, they can generate significant signal enhancement. Various signal processing techniques have been developed to achieve highly sensitive, specific, and quantitative imaging of the bubbles for flow and perfusion imaging [2], [3].

Another significant advance in biomedical ultrasound is the development of high frame-rate (HFR) ultrasound imaging techniques for various clinical applications [4]. Different approaches have been proposed to improve the frame rate for cardiac imaging. These include multi-line acquisition, multi-line transmission and diverging wave transmission [5-11]. The benefit of imaging with diverging waves has been shown for

both 3D cardiac Doppler [8] and cardiac elastography [9]. The first combination of HFR cardiac imaging using pulse inversion (PI) and diverging waves for contrast echocardiography (CE) ultrasound, named HFR CE, for in-vivo myocardium perfusion experiments was shown recently [11]. The contrast between the heart chamber full of ultrasound contrast agents and the myocardium was improved by a factor of 2 compared to standard focused transmission, even with a peak negative pressure for HFR CE that was 4 times lower than conventional focused CE transmission. Moreover, comparing to ~30Hz in standard CE, HFR CE can reach a frame rate of up to 5000Hz, allowing accurate tracking of fast flow structure and dynamics in the cardiac chambers.

A recent study shows the benefit of HFR cardiac imaging for flow vortex detection by using a Duplex mode (B-mode + Doppler) but without microbubble contrast agents, the signals from blood cells are weak [12]. Another clinical research shows the potential of visualising and tracking vortex with conventional CE at a frame rate of 204 ± 39 frames / s but the field of view is limited and the frame rate is still low for tracking the very fast cardiac flow [13]. We have recently demonstrated the use of HFR CE and image tracking for ultrasound imaging velocimetry (UIV, also known as echo-Particle Imaging Velocimetry or e-PIV) to map arterial flow both in vitro and in vivo [14]. A unique flow profile visualisation and quantification using controlled microbubbles destruction has also been demonstrated using HFR CE [15].

The aim of this work is to demonstrate the feasibility of cardiac flow mapping in vivo using HFR CE.

II. MATERIALS AND METHODS

A HFR CE system based on a 128-Verasonics platform (Verasonics Inc., Redmond, WA) mounted with a 96 element

TABLE I
TRANSMISSIONS PARAMETERS

	HFR-CE	CE
Frequency (Cycles)	1.50 MHz (3)	
Angle range	30	90
Number of angles/lines	11 (x2 PI)	82 (x2 PI)
Focus	---	80 mm
Frames	1000	120
PRF	5500 Hz	5500 Hz
Frame rate	250 Hz	30 Hz
MI at 80 mm	0.05	0.2

P4-1 phased-array transducer was used during in-vivo sheep experiments.

A. Transmission

Two transmissions have been used during the experiments: a conventional CE transmission and a HFR CE transmission based on the diverging transmission [10].

The CE using a line by line scanning approach is similar to those used in most of the commercial ultrasound scanner. In order to obtain one CEUS frame, successive focused wave were transmitted along an angular width. Moreover for each angle step, two successive signals in opposite phase are transmitted, recorded and summed. It consists of the pulse inversion image formation.

In order to obtain a diverging wave, a virtual point source was created behind the probe creating a diverging beam which enlarge the region illuminated. Similar to the plane wave imaging, a single diverging wave has a low contrast and resolution, therefore coherent diverging compounded image is obtained by varying the position (steering) of the virtual point source and by coherently averaging the echoes of the diverging transmissions. For each steering angle, two successive pulses of opposite phase were transmitted and combined in post-processing to form the PI image. The transmission parameters are show in Table I.

B. In-vivo sheep experiment setup

The in-vivo CEUS experiment was conducted under licence from the UK Home Office at the University of Edinburgh. The CEUS HFR was evaluated in-vivo on an adult female Scottish Greyface sheep under terminal general anaesthesia maintained using isoflurane [16] The probe was held using a metallic arm in order to capture the same image of the left ventricle. The sheep were positioned slightly on the left side that ensured optimal heart imaging and avoided reflections from the ribs. During the acquisition, ventilation was transiently paused by extubation to avoid chest movement.

C. echo-PIV tracking

In this study, an incoherent ensemble correlation approach is used to track the heart chamber flow [15]. A joint correlation function is obtained by summing the cross-correlation maps acquired from the multiple low resolution images (LRI) obtained from the beamformed images for each angle. Such a modification is performed as the decorrelation of the speckle

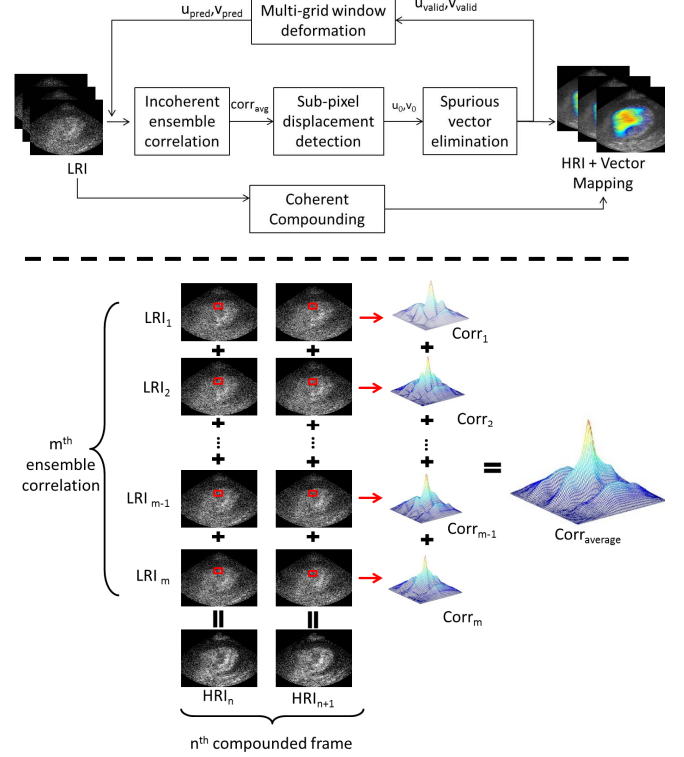


Fig. 1. (Top) Basic workflow of the modified echo-UIV algorithm to generate velocity mapping overlaid on the high resolution images (HRI) (Bottom) Principle of incoherent ensemble-correlation approach where cross-correlation is applied to each a pair of low resolution images (each beamformed angle) to generate an averaged correlation with higher signal to noise ratio.

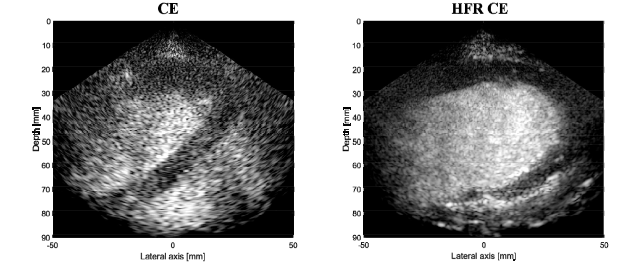


Fig. 2. In-vivo sheep left ventricle log-compress (Left) conventional CE and (Right) HFR CE images displayed with a 50 dB dynamic range.

pattern depends not only on the velocity gradient, but also on the motion artefacts that generated from the flow motion between pulse emissions. This is different from the coherent ensemble correlation approach which is done on the final coherent compounded images. By averaging the correlation plane computed from the low resolutions images, the motion artefacts can be circumvented and robust velocity estimation can be acquired. The schematic workflow of modified-UIV analysis and the principle of incoherent ensemble-correlation are illustrated in Figure 1.

III. RESULTS

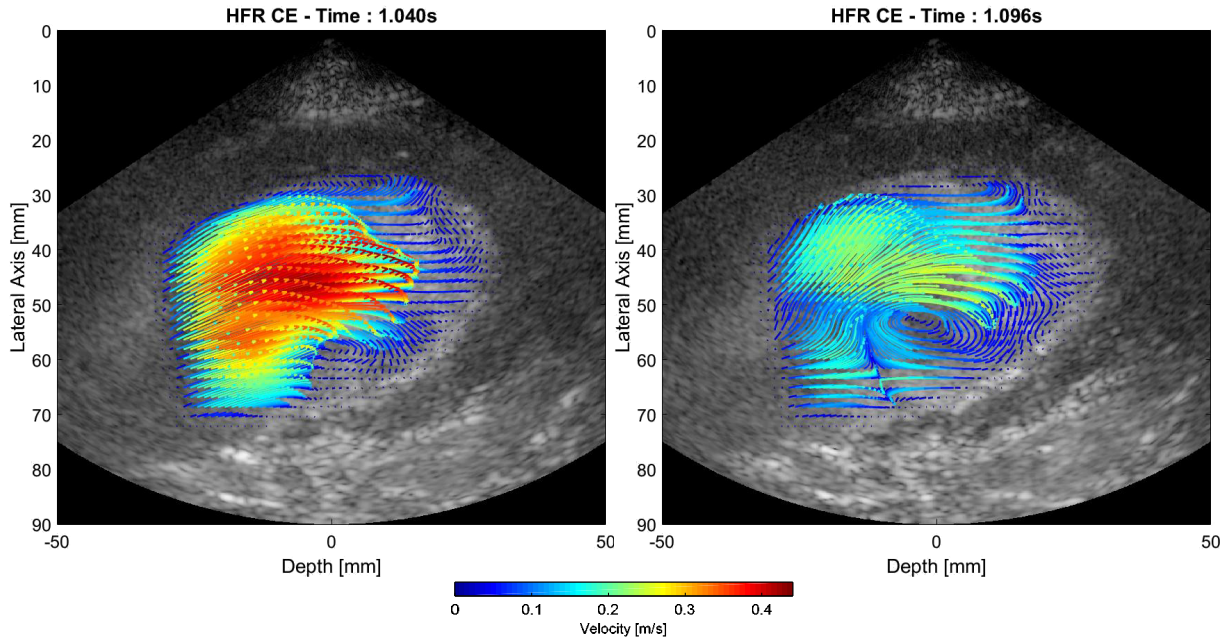


Fig. 3. Flow velocity mapping frames of a sheep left ventricle based on microbubbles tracking [15]. The first frame (Left) is few milliseconds after the opening of the valves. The second frame (Right) shows the vortex formation inside the LV chamber.

Figure 2 shows a single frame of the conventional CE and HFR CE acquisitions. The echo-PIV tracking has been processed on the data acquired using both conventional and HFR techniques. However, the tracking of the chamber was failed when using the conventional CE images and therefore not shown in the figure. Figure 3 shows the flow velocity mapping of HFR CE images in one cardiac cycle. The maximum velocity is obtained few milliseconds after the opening of the valves as shown on the left image in Figure 3. A maximum of 0.45 m/s was also estimated. After few milliseconds there is a vortex form inside the left ventricle and was clearly visible in right image. Such flow patterns repeated all along the acquisition.

IV. DISCUSSION AND CONCLUSION

In this work we show the feasibility of microbubbles flow tracking in high frame rate (HFR) contrast echocardiography (CE) ultrasound acquisition with a high temporal resolution and low MI. Previous research study has shown the possibility to use HFR cardiac imaging for flow vortex detection by Duplex mode [12]. Furthermore, clinical paper highlight the possibility to track vortex with a conventional CE acquisition but with a limited field of view and a frame rate still not enough for tracking very fast cardiac flow [13].

The proposed HFR CE method based on a novel echo-PIV tracking shows the potential to detect vortex near the valves during filling phases agreeing with clinical study [13]. The velocity map shown in Figure 3 highlight the robustness of the tracking of fast flow all along the acquisition, however the maximum velocity detected may be lower compared to the maximum flow inside the left ventricular chamber (up to 1 m/s). Several factors may explain this low value such as the non-optimal position and angle of the ultrasound probe relative to the cardiac chamber during the acquisition. Moreover the

different parameters of the acquisition and echo-PIV tracking algorithm are not optimized.

The preliminary results are promising and highlight the possibility of our novel echo-PIV tracking for CE acquisition. Further in-vitro experiments with controlled flow and in-vivo acquisition will be conducted to optimise acquisition and echo-PIV parameters.

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