Accurate peak velocity detection in the ascending aorta: Comparison between *k-t* BLAST accelerated FVE and echocardiography

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INTRODUCTION Peak blood velocities, as detected by echocardiography, play an important role in the evaluation of aortic valve stenosis. However, it is not always possible to achieve reliable results due to poor acoustic window conditions. In previous studies, MR phase contrast (PC) measurements have been compared to echocardiography as the clinical standard procedure [1]. Although a good correlation between MR and echocardiography was found, PC tended to systematically underestimate peak velocities. To address the shortcomings of PC measurements, Fourier velocity encoding (FVE) may be employed to resolve velocity distributions within a voxel. FVE measurements are, however, not readily applicable in patients due to their long acquisition times. Recently, the *k-t* BLAST reconstruction framework [3] has been incorporated to accelerate FVE data collection without sacrificing resolution in the temporal, spatial or velocity dimensions. This work aimed at investigating the accuracy of this new FVE technique in terms of peak velocity detection, even at reduced spatial resolution. In phantom experiments, peak velocities were detected for different pulsatile flow rates using non-accelerated and accelerated FVE acquisitions. Furthermore, *in vivo* FVE data were collected in 4 healthy volunteers and two patients with aortic valve stenosis using breath-held, accelerated FVE sequences. All measurements were compared to the Doppler ultrasound standard.

METHODS FVE data collection was accelerated according to the *k-t* BLAST framework (Fig. 1) [3,4]. In this approach, aliasing in the high-resolution undersampled data is resolved utilizing low spatial- and velocity-resolution training data (prior knowledge). This results in artifact-free high resolution images.

Phantom experiments were carried out to investigate the accuracy of accelerated FVE scans in terms of peak velocity detection. For this purpose, pulsatile flows were generated in tubes with inner diameters of 8 mm and 20 mm and varying degrees of area stenosis (0%, 25%, 50%, 75%, 90%) resulting in velocities in the range from 30 to 550 cm/s. FVE data were acquired with no, 8- and 16-fold acceleration on a 1.5T MR system (Philips, Best, The Netherlands). Subsequently, peak velocities were calculated according to the approach proposed by Galea et al. [5] and compared to Doppler ultrasound. Scan parameters used for the FVE measurements were: FOV: 58x230 mm^2, resolution: 0.9x0.9x5 mm^3, TE/TR: 4.0-7.1/12.0 ms, FVE steps: 16. Non-angulated, pulsed wave Doppler data were acquired using a standard General Electric, Vivid 7 system. To analyze the impact of large image voxel on the peak velocity detection, the in-plane spatial resolution of the accelerated FVE scans was reduced, mimicking breath-held *in vivo* acquisitions in the ascending aorta. The experiments were repeated with the following settings: FOV: 227x330 mm^2, resolution: 2.6x2.6x5 mm^3, TE/TR: 3.0-6.0/5.2-7.3 ms, FVE steps: 16. For comparison, peak velocities were additionally determined using conventional PC velocity mapping (FOV: 58x230 mm^2, resolution: 0.4x0.4x5 mm^3, TE/TR: 4.3-5.3/12.0 ms, 2 signal averages).

In vivo, peak velocities were measured in 4 healthy volunteers and 2 patients with aortic valve stenosis using 8- and 16-fold accelerated FVE measurements (scan parameters: resolution: $(1.3\ mm)^2$-$(2.8\ mm)^2$ TE/TR: 2.3-3.1/4.8-6.6 ms, FVE steps: 16, venc: 150-700 cm/s, 24-32 cardiac phases, breath-hold duration: 15-20 sec). For comparison, Doppler echocardiography was performed in the same session using pulsed wave mode (volunteers) and continuous wave mode (patients).

RESULTS The phantom experiments showed a good agreement between peak velocities detected by Doppler ultrasound and by accelerated and non-accelerated FVE (Fig. 2a). At higher velocities (>350cm/s), accelerated FVE slightly underestimated peak velocities compared to conventional FVE. For large voxel volumes, peak velocities derived from accelerated FVE data stayed within 10% of the values from the high-resolution, non-accelerated FVE scan, while standard PC exceeded the 10% limit (Fig. 2b).

Figure 3 shows the *in vivo* results acquired in the ascending aorta of both healthy volunteers and patients with aortic stenosis. The time curves of the peak velocities acquired in volunteer 1 are depicted in Figure 3a. A good agreement between accelerated FVE and echocardiography was found (< 15%) (Fig. 3b). Only for one patient, FVE significantly overestimated the echocardiography result. Figure 4 shows a scout image revealing the flow jet (arrow) and the slice position of the FVE measurements performed in patient 2. On the right, the maximum-intensity-projection (MIP) and its derivative used for peak velocity detection according to Galea et al. [5] are depicted. The detected peak velocity was 439 cm/s.

DISCUSSION In this work, the accuracy of FVE measurements accelerated using the *k-t* BLAST approach was investigated in terms of peak velocity detection. Phantom experiments showed that the accuracy of peak velocity detection was preserved for accelerated FVE scans. Slight underestimation of high velocities might be caused by slight temporal low-pass filtering from the *k-t* BLAST reconstruction at high acceleration factors. *In vivo* experiments showed a good agreement between echocardiography and accelerated FVE data acquired in healthy volunteers. In one patient, results from FVE and echocardiography were in full agreement, while in the other patient peak velocities from FVE were considerably higher than the ultrasound measurements. A review of this patient's medical records revealed a history of unsuccessful ultrasound examinations due to pour acoustic window conditions, while accelerated MR FVE was able to detect peak velocities. This fact confirms the value of having an MR alternative. In conclusion, *k-t* BLAST accelerated MR FVE holds promise to allow for accurate peak velocity detection even in patients with poor acoustic window.

REFERENCES 1. Caruthers S. D. et al. Circulation 2003. 108:2236-2243 2. Moran P. R. et al. Magn Reson Med 1984, 1(4):197-203 3. Tsao J et al. Magn Reson Med 2003. 50 (5):1031-1042; 4. Hansen M. S. et al. Proc. ESMRMB 2004, MAGMA. 17(1):1-329, 5. Galea D. et al. Med Phys 2002, 29(8):1719-1728

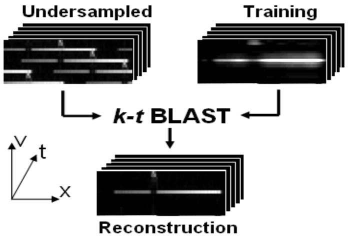

Figure 1. Schematic of the *k-t* reconstruction framework. High-resolution undersampled and low-resolution training data are reconstructed to unaliased high-resolution images.

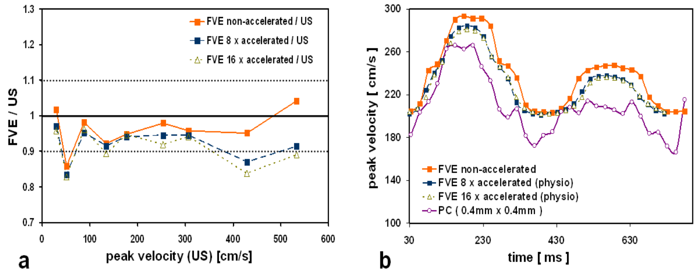

Figure 2. Phantom experiments: **a** Comparison of accelerated and non-accelerated FVE with Doppler ultrasound. **b** Peak velocities over time measured in a 20 mm tube with 75% stenosis using low-resolution FVE and conventional PC velocity mapping.

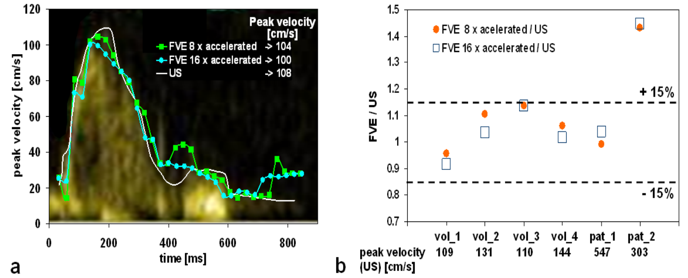

Figure 3. **a** Peak velocities over time acquired in volunteer 1 **b** Comparison of peak velocities acquired in healthy volunteers (vol) and patients with aortic valve stenosis (pat).

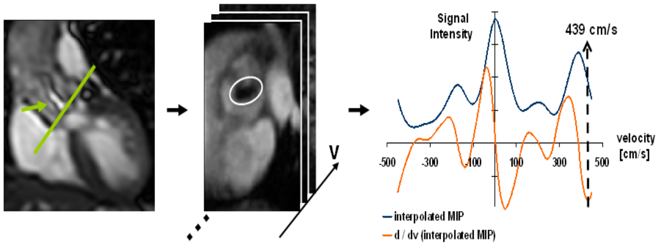

Figure 4. Visualization of flow jet (arrow) and position of FVE imaging slice (line). Slice of FVE data representing static tissue in peak systole, valve only partly opened (ellipse). Corresponding MIP and its derivative used to determine peak velocity.