Reduction of In-flow Enhancement in 3D Balanced SSFP cine sequences

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Introduction: Three dimensional (3D) b-SSFP cine acquisition of the whole heart and great vessels has generated considerable interest over the last few years. This approach allows simplification of cardiac MRI akin to cardiac CT, but its clinical application has been hampered because of a reduction in contrast between blood and myocardium. In 2D b-SSFP sequences there is very good contrast between the blood and the myocardium, however, in 3D, the blood signal intensity drops drastically. The purpose of this study is to describe the decrease in signal from blood in 3D b-SSFP sequences and to provide potential practical solutions. Experiments and simulations show that blood signal in 3D b-SSFP sequences decays because of a lack of in-flow enhancement.

Methods. ***Experiments:*** In all the experiments a b-SSFP sequence with $\alpha = 60$, TR/TE ≈ 3/1.5 was used. The scans were performed on a 1.5T system (Achieva, Philips Medical System), equipped with a 32 receive channels [4]. To quantify the difference in contrast we acquired in 5 volunteers, firstly a non-angulated 3D whole heart cine sequence (22 s breath-hold, 13 cardiac phases, resolution 2.7 mm^3, 55 transversal slices) and secondly a set of 2D short-axis slices (multiple breath-holds (15 s each), 20 cardiac phases, resolution 2.2x2.2x10 mm^3) covering both ventricles. The 3D data set was reformatted into the same view as the 2D scan and then the contrast was measured by selecting a ROI in the blood pool and myocardium during a diastole phase. Additionally, to study the inflow effects, we acquired in one volunteer a 2D phase contrast scan and a 3D b-SSFP sequence (12 axial slices, resolution of 2x2x7 mm^3, 20 cardiac phases) both positioned over the Inferior Vein Cava (IVC). Segmentation of the IVC for all cardiac phases and slices were done to analyse the data and compare with simulations. ***Simulations***: Magnetization evolution of a b-SSFP sequence was calculated using matrix formalism [1,2,3]. Modelling of the in-flow was performed by subdividing the 12 imaging slices (N) into 1680 thin 0.05 mm sub-slices (Ns). The pulsatile measured flow (phase contrast scan) in the IVC, was used as input for the simulations. Matrix operations were performed for each of the sub-slices separately. Finally, the total b-SSFP signal for each imaging slice in each TR was calculated by the complex sum over Ns/N. The following parameters were used for the simulations: T1=1000, T2=100, $\alpha = 60$, TR/TE = 3/1.5. On-resonance condition was considered.

Results and Discussion: In all volunteers a loss of blood signal intensity was found in the 3D examination compared to the 2D scans (see Fig a,b). Fig (c,d) displays, for each volunteer, the intensity level of blood (black bars) and myocardium (grey bars) from the 2D (c) and 3D (d) scans. The mean drop in blood signal intensity from 2D to 3D was 34%. Figure e) shows that at two different cardiac phases (5,14) the blood signal intensity changes because of changes in the flow in the IVC. There is also a 50% drop in the maximum signal between the slices nearest and furthest away from the inflow of blood. Results of the simulations are shown in Fig f), the maximum intensity value for each slice is shown for both simulated data and real data. The maximum intensity for a given thickness of the 3D volume can be calculated by integrating signal values trough the slices. Fig g) shows signal intensity drops 25% by increasing the thickness of the 3D volume from 14mm to 77 mm. The lost of blood signal intensity in 3D whole heart imaging is due to the fall in the spins of blood flowing into the heart as these have already been excited by a number of RF pulses, (i.e. only small fraction of fresh spins are entering the imaging volume). Possible solutions to this include changing the angulation and size of the 3D volume to increase the inflow of fresh spins or dividing the 3D data set into smaller slabs. For instance the acquisition of a 3D volume in short axis geometry (fig 1h) excluding as much of the atriums, IVC and Superior Vein Cava (SVC) as possible improves the contrast by a factor 2.8 (compared to the 3D non-angulated data) during diastole. The image quality obtained is comparable to the 2D multi-slice technique, but this approach is not ideal because it requires special planning and provides an incomplete data set that is difficult to reformat into other views. Another possibility is to acquire a non-angulated 3D scan containing different slabs (See fig I), the disadvantage of this approach is that slabs which contains the atriums have lower signal (slab 1,2 in fig i), however the data can still be reformatted into any view.

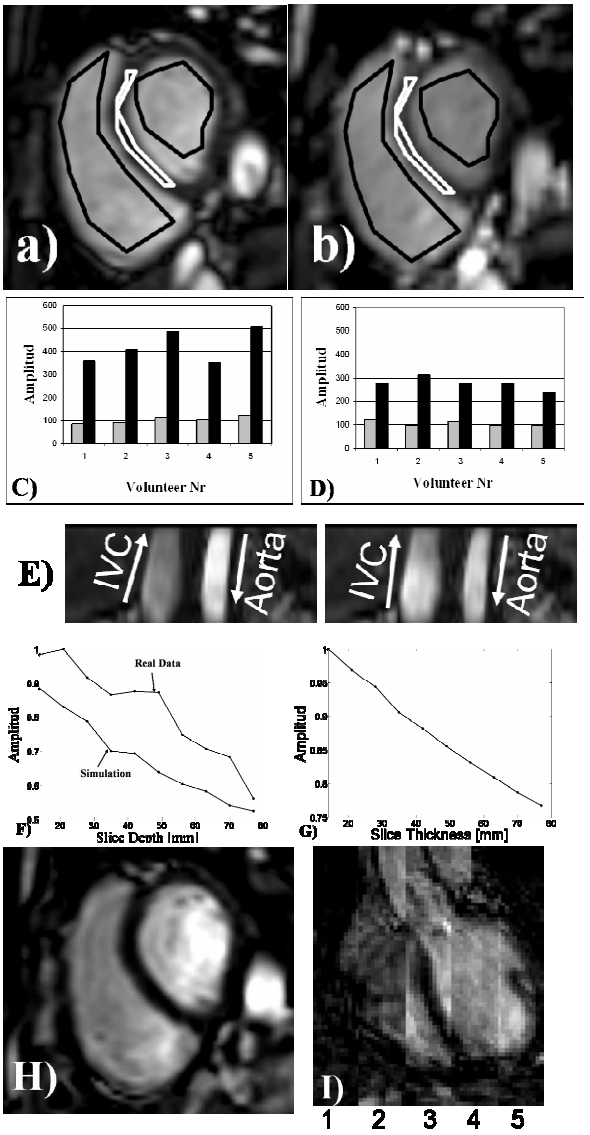

Fig. a) Multice-slice SA view. b) reformatted 3D data. c), d) Signal intensity of the blood and myocardium from 2D and 3D scans respectively. e) reformatted 3D data in coronal view showing the aorta and IVC at different time points f),g) Simulation result of blood signal decays, and maximum intensity expected as a function of slice thickness. h)Slice of a 3D volume of the short axis view. I) Slice of a multi-slab acquisition.

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Conclusions: Simulations and experiments have shown that 3D b-SSFP sequences are sensitive to in-flow enhancement. Some potential solution has been discussed. Further investigation is required to enable the acquisition of 3D b-SSFP volumes in the clinical setting.