Multi-Slice, First-Pass Myocardial Perfusion MRI with Undistorted Arterial Input Function and Higher Myocardial Enhancement at 3T

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***Introduction*:** The reliability of perfusion measurements derived from T_1-weighted, first-pass myocardial perfusion MR images depends on the accuracy of the arterial input function (AIF) [1], which is the concentration of the contrast agent (typically Gd-DTPA) in the left ventricular cavity as a function of time, and the signal-to-noise ratio (SNR) of the myocardial enhancement. The accuracy of AIF conflicts with the SNR of myocardial enhancement, as the T_1 sensitivity is heavily dependent on the concentration of Gd-DTPA and recovery time of magnetization. A combination of long recovery time of magnetization and high dose of Gd-DTPA is desirable for high myocardial enhancement. However, this condition typically leads to underestimation of the AIF due to near complete recovery of the blood magnetization at peak Gd-DTPA concentration. In contrast, a combination of low dose of Gd-DTPA and short recovery of magnetization is desirable for accurate assessment of AIF, but this condition causes low enhancement in the myocardium. A previous study has shown that relatively accurate AIF and high myocardial enhancement can be imaged separately at their optimal T_1 sensitivity time points and spatial resolution (6 mm spatial resolution is sufficient to resolve the left ventricular blood pool from the endocardium) during a high dose injection of Gd-DTPA, using separate TurboFLASH sequences [2]. The purposes of this study were to extend their work in order to: 1) acquire both AIF and myocardial enhancement images sequentially at their optimal T_1 sensitivity time points (short and long recovery time, respectively) and spatial resolution (low and high, respectively) after a single saturation pulse, 2) perform multi-slice imaging using fast gradient echo/echo-planar imaging (FGRE-EPI)[3], and 3) to achieve higher myocardial enhancement by imaging at 3T.

***Methods*:** An FGRE-EPI sequence was modified to image the AIF and myocardial enhancement images sequentially at their optimal imaging conditions after one saturation pulse (Fig. 1). Imaging parameters for the high-resolution acquisition with long recovery of magnetization included: field of view = 400 x 250 mm^2, acquisition matrix = 192 x 90, in-plane resolution = 2.1 x 2.8 mm^2, slice thickness = 10 mm, TE = 1.4 ms, TR = 6.7 ms, flip angle = 10°, echo train length = 4, repetitions = 40, slices = 3, and bandwidth = 1736 Hz/pixel. The saturation recovery delay (TD), defined as the recovery time before image acquisition, was 46.9 ms. The low-resolution image acquisition with short recovery of magnetization used the same set of parameters except: acquisition matrix = 64 x 24, in-plane resolution = 6.3 x 8.9 mm^2, TE = 1.0 ms, TR = 4.2 ms, and TD = 21.6 ms. A 90° composite B_1-insensitive rotation pulse was used to perform more complete saturation of magnetization (pulse duration = 5.1 ms; gradient spoiling duration = 6.6 ms). The total scan time per slice was 200 ms. Baseline proton-density weighted (PDW) images were acquired with no saturation pulse and flip angle reduced to 3 during the first cardiac cycle. All imaging experiments were performed on a 3T whole-body MR scanner (Trio, Siemens) equipped with a 8-channel phased array RF coil and a gradient system capable of achieving a maximum gradient strength of 40 mT/m and a slew rate of 200 T/m/s. Multi-slice imaging of AIF, though not required, provided the option of averaging over multiple data sets or discarding any image sets with artifacts, due to the amplitude modulation across k-space from the initial fast recovery of magnetization. Four healthy human subjects were imaged at 3 short-axis (apical, mid-ventricular, basal) views of the heart. A relatively high dose of 0.1 mmol/kg of Gd-DTPA (Magnevist, Schering) was injected through an 20-G cannula into an antecubital vein at a flow rate of 6 ml/s by a power injector (Medrad), followed by a 20 ml flush of saline. Since the AIF and myocardial signal images were acquired at different recovery times, the blood signal from the AIF was normalized to match the magnitude of its recirculation peak to the corresponding peak magnitude from the high-resolution image set acquired with long TD. The blood signal amplitude of the first pass bolus was computed for both data sets. The SNR of myocardium enhancement was computed from the high-resolution images. T_1-weighted images were also normalized by the corresponding PDW image.

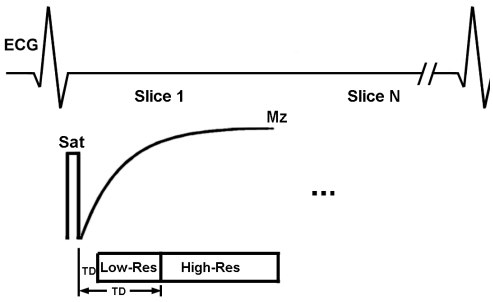

Fig. 1. Schematic diagram of the pulse sequence. The low-resolution AIF and high-resolution myocardial signal images are acquired sequentially at their optimal T1 sensitivity time points after saturation, using FGRE-EPI.

***Results*:** Figure 2 shows representative low-resolution AIF and high-resolution myocardial enhancement images at peak blood enhancement. Figure 3 shows representative blood signal-time curves normalized to match the recirculation peaks. In all cases, the normalized blood signal amplitude of the first pass bolus was consistently higher for the AIF than the myocardial enhancement images (137.8 ± 37.1 vs, 79.5 ± 8.8; $p < 0.001$). A combination of long recovery of magnetization, high dose of GDPA, and 3T imaging yielded relatively high myocardial signal enhancement (SNR = 17.1 ± 3.3) at voxel resolution of 2.1 x 2.8 x 10 mm^3.

***Discussion*:** This study demonstrates that both relatively accurate AIF and higher myocardial enhancement images can be acquired sequentially after a single saturation pulse at their optimal T_1 sensitivity time points and spatial resolution during a high dose injection of Gd-DTPA. Using this multi-slice FGRE-EPI sequence at 3T seems very promising for improving the reliability of perfusion estimation derived from first-pass perfusion MR images.

References

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3. Ding, S et al. MRM 1998; 39:514-519.

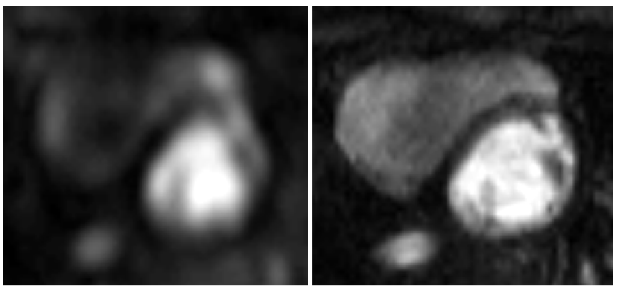

Fig. 2. Representative images of the low-resolution AIF (left column) and high-resolution myocardial enhancement (right column) images at peak blood enhancement.

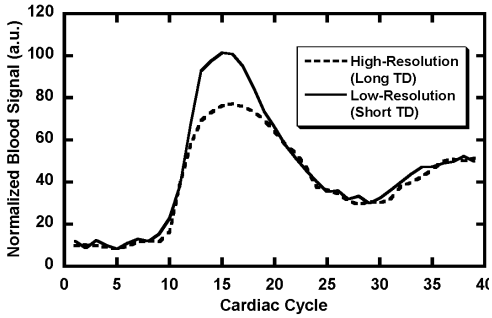

Fig. 3. Representative blood signal-time curves normalized to match the magnitude of the recirculation peaks, showing underestimation of peak signal for the high-resolution (long TD) image due to nonlinear magnetization relaxation.