Three-Dimensional Myocardial Perfusion MRI with an Undersampled 3D Hybrid Radial Sequence

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Introduction: Current myocardial perfusion imaging methods with MRI typically provide three to four 2D slices each heartbeat. Greater spatial coverage of the heart with high temporal and spatial resolution and good SNR could improve the utility of cardiac MRI perfusion. 3D perfusion MRI might be advantageous compared to 2D in terms of a consistent volume acquisition and more robustness to inter-frame motion and undersampling. Here an undersampled hybrid 3D turbo-FLASH sequence with in-plane radial k-space sampling and Cartesian encoding along kz and saturation recovery preparation was developed. Reconstruction was done using spatially-temporally constrained reconstruction (STCR) (1) to remove the undersampling artifacts.

Methods: For 3D hybrid radial acquisition, inconsistent projections can cause severe streaking artifact in-plane and there can be aliasing artifacts in the slice direction. In order to minimize these inconsistencies, we first perform simulation studies and determine optimal acquisition parameters. *Simulations:* The signal of the n-th readout of saturation recovery turboFLASH can be presented as

$$M_{xy}(n) = M(1-e^{-\frac{TD}{T_1}})a^{n-1} + M(1-e^{-\frac{TR}{T_1}})\frac{1-a^{n-1}}{1-a} \qquad [1],$$

where $M = M_0 \sin(\alpha) e^{-\frac{TE}{T_2^*}}, a = \cos(\alpha) e^{-\frac{TR}{T_1}}$,

TD is the time between the saturation pulse and the first readout radio-frequency (RF) pulse. From equation [1], it can be derived that when

$$\alpha = \cos^{-1}\left(e^{\frac{TR}{T_1}}\left(1-\frac{1-e^{-\frac{TR}{T_1}}}{1-e^{-\frac{TD}{T_1}}}\right)\right) \qquad [2],$$

$M_{xy}(n)$ is independent of n. In this case, the consistency is perfect. This helps to select TD and α values to get a steady-state signal. Fortunately, for a given TD, the flip angle depends only weakly on T1, as seen in Figure 1.

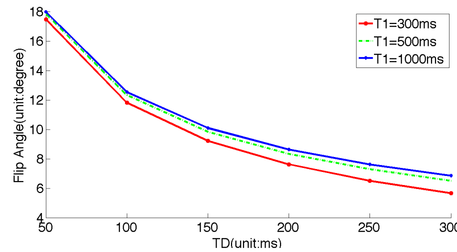

Figure 1: The flip angle-TD plot calculated using equation [2] with TR=2.5ms. The flip angle is insensitive to T1 changes.

With TR=2.5 ms, for each set of T_1 (ranging from 100 ms to 2000 ms), TD (from 50 ms to 300 ms), and α (from 2º to 30º), a signal intensity-readout index curve can be determined by equation [1]. Here the coefficient of variation (CV), the standard deviation divided by the mean value, was used to evaluate the stability of the curve (consistency of signal intensity changing with readout index). For T_1=700 ms, the CV image with different TD and α using 72 readouts is shown in Figure 2. The sets composed of TD and flip angle, such as (150 ms, 12º), (150 ms, 10º), and (200 ms, 8º), provide the smallest CV values. Similar results were found for other T_1 values; and it can be concluded that the signal consistency was insensitive to T_1. These results were consistent with equation [2], as also shown in Figure 1.

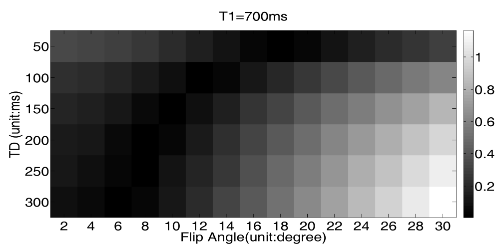

Figure 2: The coefficients of variation with different TD and different flip angle when T_1=700 ms. The sets composed of TD and flip angle, such as (150 ms, 12 degree), (150 ms, 10 degree), and (200 ms, 8 degree), provide smallest CV value.

A volunteer was imaged on a 3T scanner, the DC term (the sum of signal intensity over the excited volume) of 48mm short-axis slab covering the center portion of left ventricle were recorded for the first 144 readouts after saturation with TD=190ms and TR=2.5 ms, and the signal intensity–readout curves are shown in Figure 3.

Invivo acquisitions: 5 subjects were scanned using a 3.0 T Siemens scanner with an ECG-gated, 3D saturated recovered turbo-FLASH sequence. Imaging was done during free breathing. The radial sampling was rotated in both the slice encoding and time dimensions. Either Gd-BOPTA or gadofoveset trisodium was injected. The imaging parameters were: TR=2.1-2.9ms, TE=1.1-1.4ms, flip-angle=12-14º, SRT=140-170ms, FOV=360x360mm^2, number of rays per slice=20-24, 8-10 slices with partial Fourier factor in slice direction=6/8, spatial resolution=(1.8-2.8)x(1.8-2.8)x(6-10)mm^3, total readout time≈300 ms for one time frame. Reconstruction was done with STCR (1).

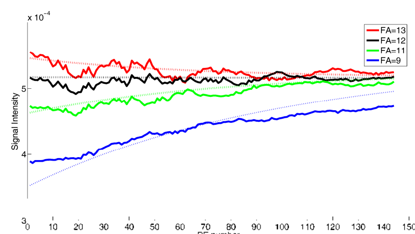

Figure 3: The signal intensity changes with readout index acquired with TR=2.5,TD=190ms and the specified α of 9º, 11º, 12º, 13º, are shown by the solid line; the dotted lines show the signal intensity changes with T_1=700ms, TR=2.5ms, TD=190ms and α of 6.4º, 8º, 8.7º, 9º, using equation [2]. Due to B1+ inhomogeneity, the specified flip angle is 1.4 times larger (2), which agrees with the experimental measurements.

Results: Three set of images are shown in Figure 4-5, which reveal reasonable image quality with high coverage. Similar results have recently been reported with a Cartesian sampling pattern (3).

Conclusion: 3D stack-of-star sampling and STCR provides a promising alternative for 3D myocardial perfusion imaging. Further work is needed to optimize the approach and to compare with 2D acquisitions.

References: (1) G. Adluru et al., J Magn Reson Imaging 29 (2009) 466-73.
(2) K. Sung et al, Magn Reson Med 59 (2008) 441-6. (3) V. Vitanis et al, Magn Reson Med 64 (2010) December.

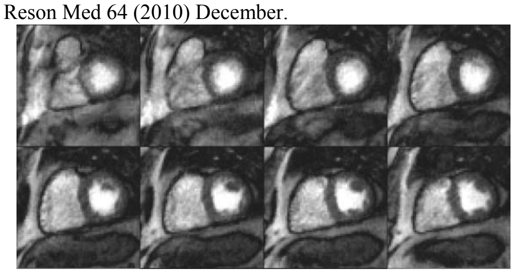

Figure 4: One time frame, acquired free-breathing, from base towards apex of 3D stack of stars in a large heart, acquired at 3T with 20 rays per kz, 8 kz encodes, 6mm slices, STCR reconstruction. The 2 edge slices were not shown here.

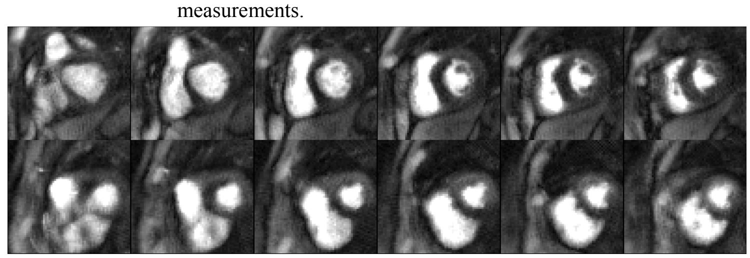

Figure 5: One time frame, acquired free-breathing, from base towards apex of 3D stack of stars in two hearts, acquired at 3T with 24 rays per kz, 6 kz encodes, 8mm slices, STCR reconstruction. The 2 edge slices were not shown here.