3D Cine MRI of Mouse Heart with Concurrent Dephasing and Excitation

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Introduction

3D magnetic resonance cine imaging (cine MRI) of the mouse heart is highly challenging due to limited spatial and temporal resolution and cardiac and respiratory motion. Recently, high-resolution 3D cine MRI with reasonable scan times (16-31 min) was achieved using a radial 3D encoding sequence with ramp sampling and contrast agent administration at 7T[1]. Radial sampling sequences can attain short TE and TR and provide decreased motion artifacts due to oversampling of the center of k-space[2] . These advantages make radial sequences well suited for 3D cardiac cine imaging. However, many of the radial ultrashort TE (UTE) sequences have some difficulties for implementation and/or image reconstruction such as ramp sampling[3] and missing k-space center[4]. In contrast, the recently introduced UTE sequence known as COncurrent Dephaseing and Excitation (CODE) is relatively easy to implement, because the acquired signals are not FID but echos[5]. Here we demonstrate CODE 3D cine MRI of mouse heart with prospective electrocardiogram (ECG) gating. We cut down the matrix size while keeping preferable spatial resolution by using a small surface coil (1.5 cm diameter) and oversampled acquisition. In general, without accelerated acquisition (eg, parallel imaging or compressed sensing), temporal resolution in the 3D mouse cardiac MRI is limited to typically ~10 frames/cardiac cycle due to its fast heart rate (>350 bpm). In order to overcome this limitation, we applied sliding window reconstruction in order to increase the nominal temporal frame rate.

Method

The sequence diagram of CODE is shown in Fig.1. In CODE, acquired signal is an asymmetric echo, since readout dephasing is performed during the slab selective excitation which is much shorter than the acquisition window. The asymmetric echo acquisition accomplishes very short echo time and thus enables acquisition before significant T_2^* signal decay. k-space sampling is performed in a radial manner as done in other 3D radial sequences[2,6].

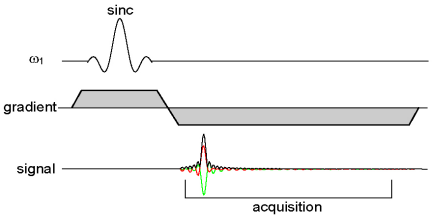

Figure 1. CODE sequence.

Five normal NOD/SCID Gamma mice (Jackson Lab, ~30g) were employed in the study. Briefly, mice were anesthetized with inhaled isoflurane supplemented with oxygen, and then positioned supine with a single loop surface coil placed onto the left side of the chest. ECG signal was monitored and used to gate the MR data acquisition. The mice were free-breathing. Temperature was maintained with a custom built water heated holder.

All the measurements were performed in a 9.4T 31 cm bore magnet. A total of 286,848 views were acquired and the image of each cardiac phase was reconstructed from 23,904 views. The 23,904 views were composed of 48 spirals that were generated by sorting a semi-random Halton sequence. Since motion effects in radial MRI spread in the radial direction[2], using the small number of views per spiral (498 views/spiral) is advantageous for suppressing the motion artifacts. We acquired 48 views for each ECG trigger. The 23,904-view acquisition was repeated 12 times, and at the beginning of every repetition, the view order was shifted by 4 views to allow each cardiac phase to cover all the 23,904 projections. A relatively small FOV of 2.0x2.0x2.0 cm was used to reduce the acquisition time, resulting in TR=2.5 ms. This FOV was slightly smaller than the sensitive region of the coil. Oversampling of 1.25 times was used to avoid signals folding from outside the FOV and making undesirable artifacts. As a result of oversampling, we acquired 120 complex points for each projection and the echo peaks were position around the 11th point (TE = 396 μs). Measurements were performed for 2 nominal flip angles (FA) of 20° and 30°. Acquisition bandwidth was 78,125 Hz and total scan time for one 3D cine imaging was approximately 23 min.

Image reconstruction was performed using a homebuilt C program. First, we picked up views for each temporal frame and performed gridding on a Cartesian 3D k-space[7] following appropriate phase correction. The Keiser-Bessel window with window width=2 and oversampling factor=2 was used in the gridding process. Gridding density correction was performed in an iterative manner[8] (3 times iteration). The gridded k-space was transformed to 3D images with Fourier transform. The matrix size was 160x160x160 and the nominal spatial resolution was 125x125x125 μm^3. The acquired data had 12 temporal frames without view sharing and the temporal resolution was 10 ms. In order to increase nominal temporal resolution, we applied sliding window reconstruction with a sliding step of 1 view, resulting in 45 temporal frames[9] (The "pseudo-temporal" resolution was 2.5 ms).

Results and Discussion

Images in the end-diastolic and end-systolic phase in a cardiac cycle are shown for flip angles of 20° and 30° in Fig.2. While signals from the posterior wall were relatively low due to the B_1 field variation of the surface coil, there was good contrast between myocardium and blood. The contrast between blood and septum (blood/septum) was ~1.6 for FA=20° and ~2.0 for FA=30°. Papillary muscles were clearly detected on the short-axis images in end-diastolic phase, especially for FA=30°.

In general, 3D imaging with a surface coil is challenging, because it suffers spatially inhomogenous transmit and receive, providing intensity variations on images. However, its small spatial coverage would be an advantage in cardiac cine MRI, since short TR and relatively large flip angle are employed so as to enhance contrast between myocardium and blood. From our experience, when we used a coil covering the whole chest, blood signals were saturated during transit through the pulmonary path from right ventricle to left atrium, leading to very poor contrast between blood and left ventricle. In contrast, since excitation with a small coil is limited to small region around the heart, the undesired blood saturation can be avoided.

In the present work, we only utilized ECG gating. Therefore, some amount of background signal and image blurring should be from respiratory motion. And, in some cases, pulsed gradients interfered with the ECG signal and disturbed the correct gating. Improvement in the gating method such as self-gated methods[10] should be desirable in further studies.

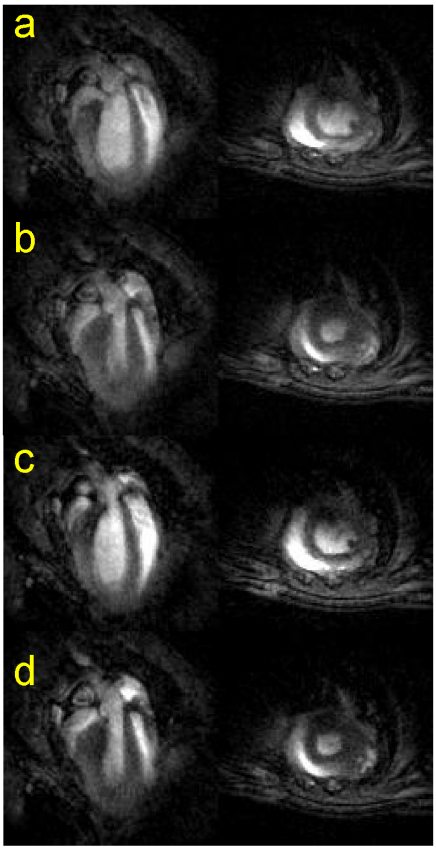

Figure 2. End-diastolic (a,c) and end-systolic (b,d) phase of mouse heart in long-axis (left) and short-axis (right) views. (a,b) and (c,d) were acquired with nominal flip angles of 20° and 30°, respectively, using 3D CODE.

Acknowledgements

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References

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