3D SPOILED GRADIENT-RECALLED ECHO SEQUENCE WITH COMPRESSED SENSING FOR DCE-MRI: IMPROVED TEMPORAL RESOLUTION AND IMAGE CONTRAST

Bin Chen[1], Kai Zhao[2], Bo Li[3], Wenchao Cai[2], Xiaoying Wang[1,2], Jue Zhang[1,3], and Jing Fang[1,3]

[1]Academy for Advanced Interdisciplinary Studies, Peking University, Beijing, China, [2]Dept. of Radiology, Peking University First Hospital, Beijing, China, [3]College of Engineering, Peking University, Beijing, China

Purpose:

High temporal resolution contributes the measurement of functional parameters in Dynamic Contrast-Enhanced MR imaging (DCE-MRI) [1]. However, simultaneous improvement of temporal and spatial resolutions by conventional methods is still challenging. To balance the tradeoff between spatial and temporal resolution in dynamic MR imaging, compressed sensing (CS) technique is introduced in our study. We aim to demonstrate the actual improvement of temporal resolution without degrading the spatial resolution by using the undersampling of phase encoding, and contrast-to-noise ratio (CNR) of the corresponding DCE images are verified.

Materials and Methods :

MRI: Three healthy male New Zealand white rabbits (3.3 to 3.5 kg) underwent a longitudinal DCE-MRI study on 3.0T MR scanner (Signa Excite^{TM}; GE Medical Systems, Milwaukee, WI, USA). 3D spoiled gradient-recalled echo sequence modified with CS scheme was scanned before and after the administration of 0.05mmol/kg of Gd-DTPA with the following parameters: TR=3.3ms, TE =1.3ms, FA=15°, slice thickness=3mm, matrix=128 × 128, FOV=180mm and 16 slices were acquired. Four accelerations (2x, 3x, 4x, 8x) were scanned as well as the fully sampling every other day for each animal in DCE MR imaging.

3-D CS reconstruction: Compressed sensing recovers underlying image from undersampled data by solving the L_1-norm minimization subject to certain constraints [2]. In sparsifying transforms of the optimization, a Total-Variation (TV) penalty was performed as well [3]. Thus, the optimization problem was described as follows:

$$\text{minimize} \quad \lambda_W \left\| \psi f \right\|_1 + \lambda_{TV} TV(f) + \left\| F_u f - d \right\|_2^2,$$

$$\text{subject to} \quad \left\| F_u f - d \right\|_2 \le \varepsilon,$$

where the underlying image of interest is f, and ψ denotes the sparsifying transform operator. λ_{TV} and λ_W trade specific sparsifying transform with finite-differences sparsity, F_u is undersampled inverse Fourier transformation operator, **d** is the available undersampled k-space data, ε controls the fidelity of the reconstruction which is set below the expected noise level [2].

Post-processing: Regions of interest (ROIs) of cortex and medulla were manually placed in the corresponding renal parenchyma of each accelerated DCE images to generate the signal intensity curves. In CNR analysis, cortex and medulla regions were delineated manually, and CNR were calculated for comparison with the fully sampled images after 10s, 30s and 1min enhancement:

CNRc-m = (Sc-Sm) / SDn;

where Sc was the signal intensity of the cortex and Sm was the signal intensity of medulla, respectively. SDn was the standard deviation of the background noise.

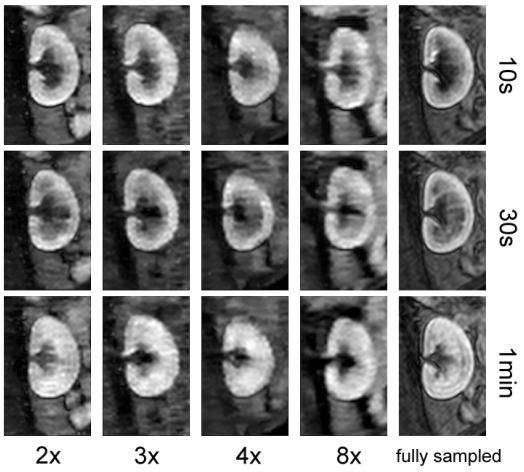

Fig. 1 Typical DCE MR image series at different CS accelerations: strong cortical and medullary contrast is found after contrast enhancement, and are significantly higher than fully sampled at 10s and 30s, p<0.05; 2x, 3x and 4x accelerated images show acceptable quality comparing with fully sampling, but not for the 8x.

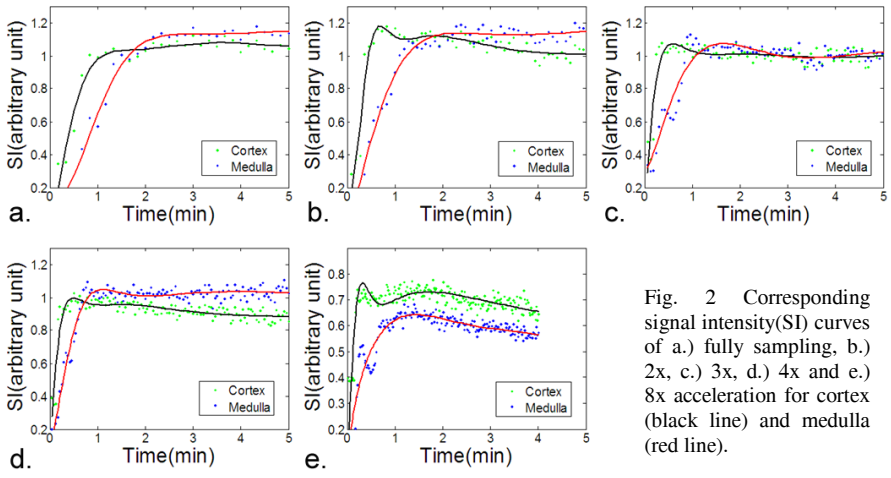

Fig. 2 Corresponding signal intensity(SI) curves of a.) fully sampling, b.) 2x, c.) 3x, d.) 4x and e.) 8x acceleration for cortex (black line) and medulla (red line).

Results:

Typical DCE images of different accelerations and the fully sampling at 10s, 30s, 1min after the injection of contrast are represented in Fig. 1, respectively. Each accelerated image (2x, 3x, 4x) has comparable image quality to the fully sampled image, but the highest accelerated images (8x) exhibits much blur due to over undersampling of k-space data in periphery. Rarely artifact is found in CS reconstructed images (first three columns in Fig. 1). Signal intensity curves of each corresponding region at different acceleration rates: a.) fully sampling, b.) 2x, c.) 3x, d.) 4x, and e.) 8x acceleration are shown in Fig.2, respectively. An increase of time points for DCE-MRI acquisition is clearly observed with the increase of acceleration factor, demonstrating the improvement of temporal resolution. Compared to the fully sampled signal intensity curves, shown as Fig.2 a.), CS accelerated signal intensity curves show obvious peak characteristics. CNRc-m of CS reconstructed images are significantly higher than fully sampled images: 12.9 ± 4.0 for 2x (p=0.0213), 13.0 ± 4.9 for 3x (p=0.0362), 14.2 ± 8.0 for 4x (p=0.0194), and 20.3 ± 6.1 for 8x (p=0.0025) vs. 8.1 ± 2.5 (CNR of fully sampling) at 10s enhancement; 19.0 ± 5.1 for 3x (p=0.0058), 19.8 ± 5.9 for 4x (p=0.0056), and 12.7 ± 2.8 for 8x (p=0.0154) vs. 7.5 ± 1.1 (fully sampling) at 30s enhancement; no significant differences are found at 1min enhancement for all acceleration cases (p>0.05).

Conclusions:

DCE-MRI combined with CS acceleration technique is adopted as a useful imaging scheme, resulting in actual high temporal resolution, which could contribute more creditable quantitative renal perfusion measurements. As a non-uniform k-space undersampling method, CS reconstruction could improve image quality and reduce artifacts for 3D DCE-MRI [2,4], thus leading to the improvement of CNR.

References:

[1] Smith DS, et al. Phys Med Biol. 2011;56(15):4933-46.
[2] Lustig M, et al. Magn Reson Med. 2007;58(6):1182-95.
[3] Tsaig Y, et al. Signal Process. 2006;86:533–548.
[4] Otazo R, et al. Magn Reson Med. 2010;64:767-76.