Accelerating Magnetic Resonance Spectroscopy Imaging by Compressed Sensing

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INTRODUCTION
Conventional phase encoded MRSI is relatively inefficient and time-consuming because it involves a large number of phase encodings [1]. As a result, small number of phase encodings are commonly employed in practice, leading to low spatial resolution and intra-voxel contamination [2]. In this study, we aimed to examine the applicability of compressed sensing (CS) [3] to MRSI in order to accelerate data acquisition. We proposed that by undersampling the 2D phase encodings in a random and non-uniform manner, the proton MRSI acquisition could be significantly reduced without sacrificing the spectral quality while maintaining high spatial resolution.

METHODS
Random Phase Encoding: The random undersampling procedure was first demonstrated by Lustig et al. [4]. In our study, undersampling along two phase encoding directions was achieved by randomly selecting the phase encoding lines (readout) measured. The acquisition order was phase encoding in Kx-direction, phase encoding in Ky-direction and readout in Kf-direction (Fig. 1a). The sampling density function was quadratic with the highest value in the center of Kx-Ky plane.

Image Reconstruction: As shown in Fig. 1, reconstruction was performed with the following procedures: (1) Undersampled data (Fig. 1a) was reordered into the form Kx-Ky-Kf (S in Fig. 1b) and the unmeasured phase encoding lines were filled with zeros; (2) 2D Fourier transform was performed along Kf direction while 2D Inverse Fourier transform were performed on Kx-Ky plane, producing X; (3) The sparse coefficients [4] of X were calculated and used to generate X*, (4) unmeasured phase encoding lines in Fig. 1a were filled with the values from X* (after transform to k-space, Fig. 1c) to form a new S; (5) steps 2 to 4 were iterated until the data converged. The convergence criterion is \( |X - X^*|/|X| < 10^{-4} \).

Phantom MR Experiments: All experiments were performed on a 7T Bruker scanner. Fig. 2a shows a T1-weighted (T1W) image of a cylindrical phantom containing dimethyl sulfoxide (DMSO) and distilled water separated by a plastic layer. The phantom was then scanned using a PRESS sequence with TR/TE = 1000/20ms, NEX = 1, no water suppression and scan time of 17 mins. The FOV 30×30mm × 20×20mm was used, with a total of 1024 phase encodings were acquired. Random undersampling was done retrospectively by randomly selecting 551 phase encodings from the full dataset (undersampling ratio = 53%).

In vivo Rat Brain MR Experiment: A semi-LASER sequence (TR/TE = 800/40ms) with 4 steps phase cycling was applied to a 1D Fourier transform was performed along Kf direction while 2D Inverse Fourier transform were performed on Kx-Ky plane, producing X; (3) The sparse coefficients [4] of X were calculated and used to generate X*, (4) unmeasured phase encoding lines in Fig. 1a were filled with the values from X* (after transform to k-space, Fig. 1c) to form a new S; (5) steps 2 to 4 were iterated until the data converged. The convergence criterion is \( |X - X^*|/|X| < 10^{-4} \).

RESULTS: With 53% undersampling ratio, the random phase encoding method preserved the boundaries between water and DMSO (Fig. 2b), with minimal ringing artifact (Fig. 2c). With 50% undersampling ratio, Fig. 3c and 3d showed that the proposed undersampling led to a 2 fold increase in resolution for delineation of the lesion boundary with only 31% increase in scan time. With the same number of averages and ¼ the voxel volume, the proposed undersampling approach still yielded spectral SNR comparable with the control.

CONCLUSION: These experimental results demonstrated that the proposed CS method can be applied to 2D MRSI, yielding a large undersampling and acceleration factor. Such approach can improve spatial resolution without compromising spectral quality.

Fig. 2: (a) T1W image of cylindrical phantom. (b) Water peak maps and DMSO peak maps of fully phase encoded and 53% random phase encoded. (c) Intensity multiplied by 5.