Compressed Sensing Multi-Spectral Imaging of the Post-Operative Spine

Pauline Wong Worters[1], Kyunghyun Sung[1], Kathryn J Stevens[1], Kevin M Koch[2], and Brian A Hargreaves[1]
[1]Stanford University, Stanford, CA, United States, [2]ASL, GE Healthcare, Waukesha, WI, United States

INTRODUCTION: Compressed sensing (CS) has seen many recent developments, enabling its use in clinical MRI [1]. An area that could lend itself particularly well to CS is MRI near metallic implants, which has been developed in recent years to resolve and correct for metal-induced spatial distortion [2-4]. These methods, generally known as multi-spectral imaging (MSI), have a relatively long acquisition time (8-16× standard) due to the extra encoding needed to excite and resolve the range of off-resonant frequencies induced by the presence of metal. However, *the total amount of signal is unchanged* so there is an inherent sparsity in MSI that could be exploited by CS to reduce acquisition time. Due to coil geometry, parallel imaging is difficult in spine imaging. The acquisition time reduction from CS could enable imaging of patients in severe pain and unable to withstand long acquisition times.

The goal here was to demonstrate and evaluate CS-MSI in human subjects with spinal hardware. First, we demonstrated the feasibility of CS-MSI by retrospective undersampling, reconstruction and comparison to a fully-sampled T2-weighted dataset. Second, we acquired prospectively undersampled CS-MSI to verify the consistency of T2-weighted contrast in MSI of the spine.

METHODS: MSI acquisitions of the post-operative spine were retrospectively undersampled and reconstructed using approximate message passing (AMP [5], see details under Reconstruction). The reduction factor was determined semi-quantitatively using a structural similarity index (SSI) [6]. A mean SSI >0.95 (with 1.0 being best and -1.0 being worst) in a region-of-interest was imposed to maintain a reasonable reduction factor.

Thirteen T2-weighted sagittal spine datasets acquired at 1.5T were undersampled and reconstructed. The original and CS images were anonymized and compared side-by-side (randomized to left and right locations) by an experienced MSK radiologist using a 5-point scale (much worse; somewhat worse; no difference; somewhat better; much better) in three categories: (1) visualization of nerves and nerve roots; (2) image artifact (metal-induced or not); and (3) overall image quality. A one-sided, paired Wilcoxon test was performed against the null hypothesis that CS images are somewhat worse than the original.

Next, T2-weighted sagittal images were acquired in volunteers with no metal to investigate the effect of prospectively undersampled CS-MSI on signal and contrast.

<u>*Reconstruction*</u>: All reconstructions used an Intel Core2Duo machine (2.4 GHz, RAM 2 GB, bus speed 1.07 GHz). Gaussian-windowed variable-density sampling was used (Fig. 1a). For the L1-minimization, approximate message passing (AMP, Fig. 1b) [5], a variation of iterative soft thresholding, was used to enable faster convergence. Also, a surface coil intensity correction was implemented to reduce the signal variation due to the linear spine array coil.

<u>*CS-MSI Acquisition*</u>: The parameters were: 3T; sagittal; TE 110-120 ms; TR 8 s; FOV 26×26 cm^2; matrix 256×104-124; slice thickness 4 mm; 20-40 sections; ETL 26; 22 excited spectral locations/slices; # ZPE matched to # sections; half Fourier; cut k-space corners. The echo-train ordering of the acquired k-space views used a non-separable ky-kz scheme (Fig. 1a) of Busse [7]. The acquisition times were 13:58 min for standard MSI and 8:06 min for CS-MSI – **a 42% reduction**.

RESULTS: Fig. 2 shows images and SSI maps from three reduction factors. From the analysis, an outer reduction factor of 2 was chosen for the retrospective study. Fig. 3 shows examples from the retrospective study, including the reconstruction times and percentage of k-space that was sampled. Results from the comparison indicate that **retrospective CS-MSI images are the same as or better than the original MSI images**, within the tolerance of "somewhat worse" to "same as", for all categories: nerve visualization: p = 0.00018; image artifact: p = 0.00031; image quality: p = 0.0030. Fig. 4 shows standard and prospectively acquired CS-MSI (2.6× outer reduction) images with comparable image contrast and quality.

DISCUSSION: Generally, successful application of CS requires data with high SNR – MSI is inherently more SNR-limited than other MRI methods due to the high readout bandwidth used to reduce signal loss and distortion. Limited SNR in MSI restricts the reduction factors achievable while maintaining image quality. Imaging at 3T may enable higher reduction to compensate for increased encoding necessary.

CONCLUSION: We have demonstrated the use of CS to enable a 1.7-2× further reduction of acquisition time in spinal MSI without compromising image quality.

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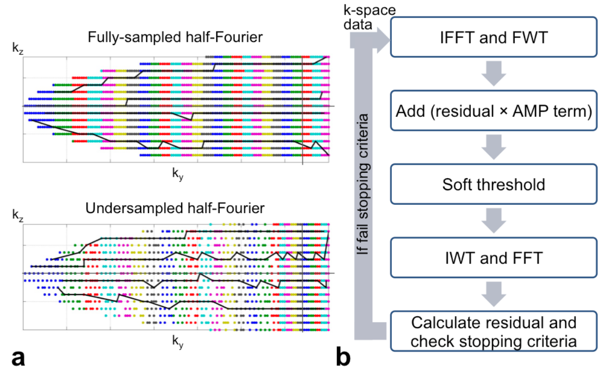

Fig. 1: (a) Fully-sampled and Gaussian-windowed prospectively undersampled half-Fourier FSE k-space. The colors refer to the echo number; four echo-train paths are shown. (b) The AMP soft thresholding algorithm used in the CS reconstruction. (F/IWT = forward/inverse wavelet transform.)

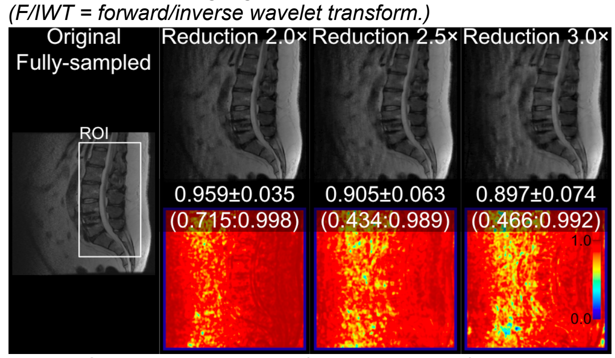

Fig. 2: Sagittal spine images, fully-sampled half-Fourier and at 2-3× outer reduction factors, and SSI maps (bottom) obtained with comparison to the fully-sampled image. The values refer to the SSI average ± s.d. (minimum:maximum) in the ROI.

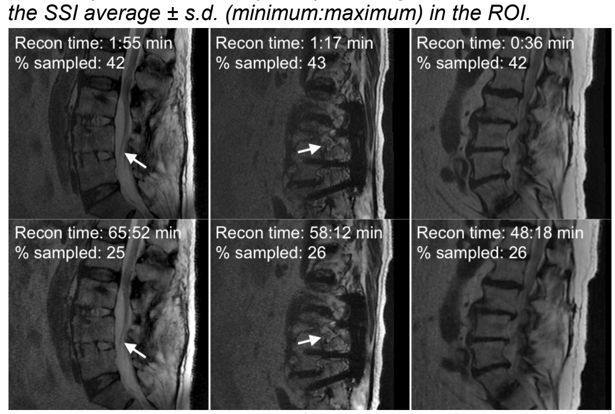

Fig. 3: Three T2-weighted MSI cases. Top: original; bottom: retrospective CS. The percent sampled in the fully-sampled half-Fourier is under 50% due to cut k-space corners. The arrows point to nerve roots that are well depicted in all images.

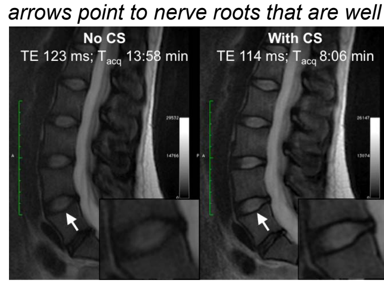

Fig. 4: Standard MSI & prospective CS MSI showing little effect of undersampling on T2 contrast in the spine. The arrows and inset point to blurring in the no-CS case probably due to motion during the longer acquisition.