

3D Compressed Sensing CPMG with Group-Sparse Reconstruction for Myelin Water Imaging

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Introduction

Myelin content is an important marker for central nervous system pathology. One way to generate myelin water maps is to utilize a multi-echo CPMG sequence. However, CPMG is inherently slow, which is especially a concern in *in vivo* small mammal studies with conflicting requirements of very high SNR and spatial resolution and reasonably short scan time. Compressed sensing (CS) has been shown to accelerate 2D CPMG acquisition for myelin water imaging with good result using a randomized Cartesian sampling scheme¹. However, further improvements, such as higher geometric efficiency and shorter echo time, can be achieved by 3D based CPMG acquisition. This study investigates the use of CS with 3D CPMG for myelin water imaging.

Methods

CS 3D CPMG experiments were carried out on a 7 T animal scanner (Bruker, Germany) using a 13 mm i.d. solenoid coil (128 × 128 × 16 matrix, TE/TR = 1500/6.738 ms, 32 echoes, 1.28 × 1.28 × 1.6 cm FOV). Two sets of data were acquired from excised rat cervical spinal cord samples that underwent dorsal column transection injury at the C5 level: 3× accelerated dataset at 1 and 2 averages (phase cycled), and 2× accelerated dataset at 1 average. Both sets included matching unaccelerated data for reference. The sampling scheme used is illustrated in Figure 1. Each echo was sampled differently to prevent constructive aliasing interference and further enforce sparsity. Readout lines are fully sampled. Daubechies 8 wavelet at 3 levels of decomposition was used as the sparsifying transform. Group-sparse reconstruction was performed using SPGL1². Non-negative least square analysis was used to calculate the T2 distribution³. Myelin water fraction maps were generated by dividing the integral from 7.75–20 ms range by the total integral of the T2 distribution.

Results and Discussion

Selected results are shown in Figure 2 and 3. The fully sampled dataset produced excellent MWF map in all slices (with exception of the central slice being corrupted with zipper artifact, shown in Figure 4). The CS reconstructed echo images are also highly accurate for both datasets. MWF maps of the 2× accelerated scan are significantly better in quality than 3× accelerated scans at 1 or 2 averages, despite taking less time than the latter. There is minimal improvement in data quality in the 3× accelerated dataset at two averages beyond removal of the zipper artifact from the central slice (Figure 4), which can be avoided by careful slab positioning. Furthermore, the 3× accelerated dataset over estimates the MWF by approximately 10% while the 2× accelerated dataset matches the MWF produced by the unaccelerated reference scan.

In modifying the 3D CPMG sequence for undersampling, changes were made to the phase encoding scheme. While single phase encode gradient pulse was used to encode all echoes, phase encode of opposite polarity around the echo signal were used for the undersampled data, which is less than optimal⁵, but is necessary for the chosen Cartesian undersampling scheme. The undersampling scheme used is easy to implement most multi-echo techniques on any scanner for. A non-Cartesian sampling scheme coupled with better probability density function will likely improve image quality further.

In *in vivo* acquisition where aliasing requires an increase in FOV, 2× accelerated scan have the potential to reduce a physiologically challenging acquisition time to a reasonable level.

Conclusions

3D CS multi-echo CPMG with group-sparse reconstruction is a promising approach for increasing acquisition efficiency in myelin water mapping for time constraint experiments.

Acknowledgements

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References: 1. Chen HSM, *et al.* 19th ISMRM, No. 2849, 2011; 2. van den Berg E *et al.*, SIAM J on Sci Comp, 2008, **31**:890; 3. Whittall KP, *et al.* Magn Reson Med, 1997, **37**, 34; 4. Jones CK, *et al.* Magn Reson Med, 2003, **50**, 206; 5. Does MD, *et al.* J Magn Reson, 1998, **131**,25.



Figure 1. Gaussian variable density sampling scheme. The 1st echo phase encoding scheme for 2x acceleration is shown. Each echo was sampled differently. *k*-space is fully sampled in the readout direction. One third of sampling points are assigned to the central fully sampled ellipse. The ellipse dimensions are proportional to the phase encode dimensions.

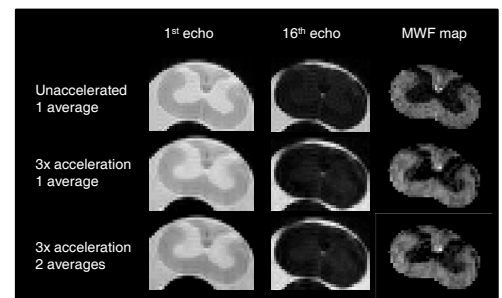


Figure 2. Comparison of echo images and MWF maps of 3x accelerated acquisition to unaccelerated acquisition. MWF maps are scaled from 0 to 1, and the echo images are normalized. Slice 6 of 16 is shown here.

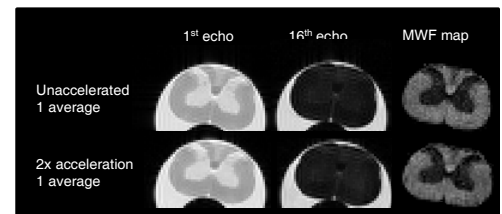


Figure 3. Comparison of echo images and MWF maps at various acceleration factors and signal averaging. MWF maps are scaled from 0 to 1, and the echo images are normalized. Slice 14 of 16 is shown here.

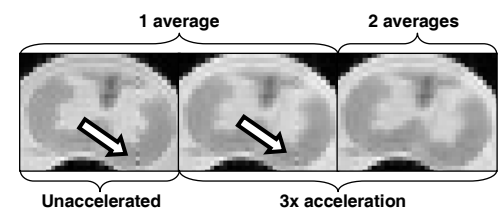


Figure 4. First echo images of central slice. Data acquired at single average without phase cycling suffers from spurious FID signal (zipper artifact, arrow). Slice is immediately adjacent to injury site.