T_2 Shuffling with Partial Fourier Acquisition and Reconstruction

Jonathan I Tamir[1], Martin Uecker[2], Shreyas Vasanawala[3], and Michael Lustig[1]
[1]*Electrical Engineering and Computer Sciences, University of California, Berkeley, Berkeley, CA, United States,*
[2]*Diagnostic and Interventional Radiology, University Medical Center, Goettingen, Goettingen, Germany*
[2]*German Centre for Cardiovascular Research (DZHK), Goettingen, Germany*
[3]*Department of Radiology, Stanford University, Stanford, United States*

Target Audience: MRI researchers interested in pulse sequence design and image reconstruction.

Introduction: Volumetric fast spin-echo (3D FSE) imaging is clinically desirable because of its robustness to off-resonance and its utility for obtaining many types of image contrasts at isotropic resolution. Nonetheless, the use of 3D FSE for musculoskeletal imaging remains a challenge due to the loss of apparent resolution (i.e. blurring) from T_2 decay[1-2]. The blurring is a result of signal relaxation during long echo trains that is not accounted for in the reconstruction. Recently, we introduced T_2 Shuffling[3], a 3D FSE-based acquisition and reconstruction method that mitigates signal blur and resolves images at multiple contrasts along the signal relaxation curve. T_2 Shuffling uses a randomized view ordering and re-acquires phase encodes throughout the echo train. The 4D data are processed with a compressed sensing reconstruction that constrains the temporal signal evolution to a low-dimensional subspace. In this work we extend the data sampling and reconstruction pipeline to support a partial Fourier acquisition. The partial Fourier constraint reduces the apparent acceleration and enables the use of additional prior information in the reconstruction.

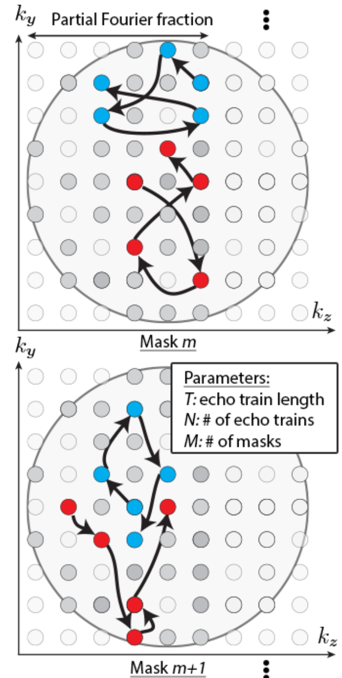

Fig 1. A series of sampling patterns are used to generate randomly shuffled echo trains. The blue (resp. red) phase encodes are chained to form echo trains.

Pulse Sequence Design: The CUBE 3D FSE pulse sequence (GE Healthcare) was modified to re-sample phase encodes at different echo times[3]. Fig. 1 depicts the echo train view-ordering scheme. The echo times are segmented into M batches and a variable-density Poisson disc sampling pattern is generated for each of the M batches. For the m^{th} sampling pattern, a sliding window is moved until T/M points are selected, where T is the echo train length. These points are randomly ordered to form the m^{th} segment of one echo train. This is repeated until all points are covered, and the procedure continues with the next sampling pattern. The parameter M controls the re-sampling redundancy in *k-t* space. To support fractional k-space coverage, the acceleration factor of each sampling pattern is reduced by the partial Fourier fraction and masked in the portion of k-space not to be acquired.

Reconstruction Algorithm: The forward model is $y = E\Theta x$, where x is the time series of real-valued, positive images, Θ is the smooth image phase (estimated from a central calibration region in the first echo), and E is the ESPIRiT[4] encoding operator. The model is relaxed, and the signal is approximated by $x = \Phi\alpha$, where Φ is a pre-determined temporal basis[3,5] of dimension K and α are the temporal image coefficients. Fig. 2 shows the reconstruction formulation. The time series of images are constrained to be real and positive, and locally low rank regularization[6] is used to exploit spatial correlations in the temporal coefficients.

$$\underset{\alpha}{\text{minimize}} \quad \frac{1}{2}||y - E\Theta\Phi\alpha||_2^2 + \lambda \sum_r ||R_r(\alpha)||_*$$
$$\text{subject to} \quad \Phi\alpha \in \mathbb{R}_+^n$$

α: Temporal image coefficients
y: Measured data $\quad E$: Encoding operator
Θ: Image phase $\quad \Phi$: Temporal basis
$||R_r(\cdot)||_*$: Block-wise nuclear norm

Fig 2. Reconstruction Formulation.

Experiments: We extended the numerical brain phantom developed by Guerquin-Kern et al.[7] to incorporate T_1 recovery and T_2 relaxation. The phantom was used to compare varying degrees of sampling redundancy and fraction of acquired k-space. Under IRB approval and informed assent, a pediatric patient's knee was scanned using the T_2 Shuffling pulse sequence (TR/TE = 1400/6 ms, ETL=80, 8 coils, 7 min. 30 sec. scan time) with 0.6 fractional k-space coverage. The sampling patterns used a redundancy batch size of $M = 8$. The data were reconstructed ($K = 4$) into 78 virtual echo time images as outlined by Fig. 2, with the phase/positivity constraint only for partial Fourier. The reconstruction was implemented using BART[8].

Results and Discussion: Fig. 3 shows the reconstruction of the numerical phantom at the 5^{th} virtual echo time without and with partial Fourier acquisition. Fig. 4 shows the in vivo reconstruction at the 5^{th} (21 ms) and 30^{th} (90 ms) virtual echo times. The use of partial Fourier acquisition and reconstruction further constrains the signal evolutions and reduces residual incoherent aliasing artifacts. The method can be used to trade off scan time and SNR.

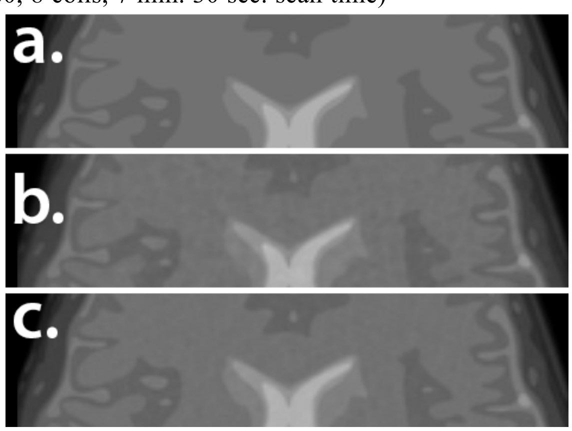

Fig 3. (a) Zoom-in of numerical brain phantom. (b) T_2 Shuffling reconstructions without and (c) with partial Fourier (0.6 fraction) at the 5^{th} virtual echo time.

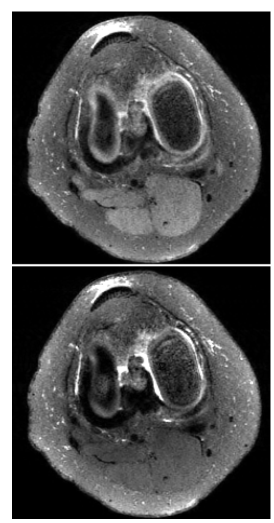

Fig 4. In vivo T_2 Shuffling reconstruction with partial Fourier fraction of 0.6 at 5^{th} and 30^{th} virtual TEs.

Conclusion: A partial Fourier acquisition reduces incoherent artifacts for T_2 Shuffling and brings multi-contrast 3D FSE within reach of clinical utility.

References: **[1]** Busse, MRM, 2008; 60:640-49. **[2]** Tariq, ISMRM 21,1664, 2013. **[3]** Tamir, ISMRM 23, 3399, 2015. **[4]** Uecker, MRM, 2014; 71:990-1001. **[5]** Huang, MRM, 2013; 70:1026-37. **[6]** Zhang, JMRI, 2013, doi:10.1002/jmri.24551. **[7]** Guerquin-Kern, MRM, 2012; 32:626-636. **[8]** BART: 10.5281/zenodo.31907.