Title: A Fast Algorithm to Correct Excitation Profile in Zero Echo Time (ZTE) Imaging

Cheng Li[1], Jeremy F. Magland[1], Alan C. Seifert[1], and Felix W. Wehrli[1]

[1]*Laboratory for Structural NMR Imaging, Department of Radiology, University of Pennsylvania, Philadelphia, PA, United States*

Introduction - Zero echo-time (ZTE) imaging methods (e.g. WASPI[1] and PETRA[2]) have shown promise for imaging short T2 species. ZTE offers higher image SNR, less T2* blurring and immunity to distortion artifacts known to occur in ramp-sampled ultra-short echo time (UTE) imaging. However, the presence of an imaging gradient during excitation causes the hard pulse to become spatially selective, leading to blurring and shadow artifacts[3]. Our previous work proposed a correction method by using quadratic phase-modulated RF pulse excitation and iterative reconstruction[4]. However, the running time of the iterative reconstruction takes several hours since each iteration involves a time-consuming full matrix-vector multiplication and around 100 iterations are needed to converge. In this work, we propose a non-iterative reconstruction method to correct the image artifacts caused by the spatial selectivity of hard pulse excitation in ZTE. This non-iterative reconstruction takes less than a minute and the reconstructed image achieves comparable image quality as the one with iterative reconstruction.

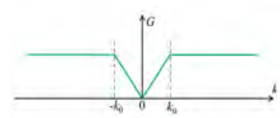

Fig. 1 Plot of the gradient strength G as a function of the distance kr from the k-space center

Methods - *Signal Model of ZTE Imaging:* In standard MRI, the signal in *k*-space, *s*(*k*), is the Fourier transform of the spatial magnetization distribution *m*(*r*). In order to include the non-uniform excitation effect, an excitation profile is superimposed onto *m*(*r*) and *s*(*k*) is expressed as: $s(k) = \iiint m(r)p(f)e^{-i2\pi\langle k,r\rangle}\,dr + \varepsilon$ (1) where *p*(*f*) is the excitation profile as a function of frequency $f = \gamma\langle G, r\rangle$, ε is the complex Gaussian noise. In conventional ZTE, *p*(*f*) is the Fourier transform of a rectangular hard RF pulse, i.e. *sinc* function. Without correction for the non-uniform excitation profile, the resulting image suffers from blurring artifacts and signal attenuation near the boundary of the imaged object.

Image Reconstruction: Since the excitation profile *p*(*f*) is a radial function invariant in the angular dimensions, in principle, the correction can be processed for each radial projection separately, reducing the 3D problem to a series of 1D problems. However, since typical ZTE imaging acquires only fractions of the half-projections, full radial projection data are not readily available to perform the 1D correction. A key observation is that the gradient strength is a continuous function of the distance k_r from the *k*-space center. For example, in PETRA, its hybrid k-space trajectory consists of Cartesian single point imaging in the k-center portion and center-out radial trajectory outside. **Fig. 1** shows the gradient strength as a function of the distance k_r in PETRA, where k_0 is the boundary of the Cartesian *k*-space region Therefore, it allows us to convert the acquired k-space samples to a pure radial trajectory, which can be achieved by first gridding to the Cartesian image space and then inverse gridding back to *k*-space with full radial trajectory. A flowchart of the proposed non-iterative reconstruction algorithm is shown in **Fig. 2**.

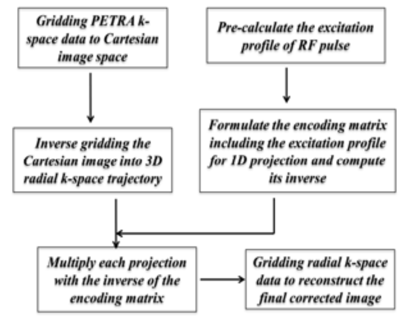

Fig. 2 Flowchart of the proposed non-iterative algorithm

Experiments: A doped-water phantom was imaged at 3T using the PETRA sequence with a four-channel receive-only head coil and body coil RF transmission and the following scan parameters: 1.30 mm isotropic voxel size, matrix size = 192^3, number of half-projections = 30,000, FA = 6º, TE = 100 µs, TR =10 ms. A series of scans were performed with dwell times of 5, 10 and 20 µs while fixing the pulse duration at 20 µs. The head of a 40 year-old healthy male volunteer was also scanned with following scan parameters: 1.17 mm isotropic voxel size, matrix size = 256^3, 50,000 half-projections, FA = 5º, 20 µs pulse duration, TR/TE = 10 ms/85 µs, 5 µs dwell time. In all scans, quadratic phase-modulated RF pulses were used for excitation.

Results- The phantom images with varying dwell times and constant pulse duration are shown in **Fig. 3**. As the dwell time gets shorter, the imaging gradient strength becomes higher and the spectral bandwidth exceeds the RF bandwidth, thereby resulting in more severe image artifacts in the uncorrected images (**Figs. 3a-c**). In the images obtained with both iterative (**Figs. 3d-f**) and non-iterative (**Figs. 3g-i**) reconstruction artifacts are effectively removed. However, the non-iterative method took only 30 seconds, more than 100 times less than the iterative approach while achieving essentially identical results. **Fig. 4** shows *in vivo* ZTE brain images with and without correction in the sagittal and coronal planes corroborating the results of the phantom experiments.

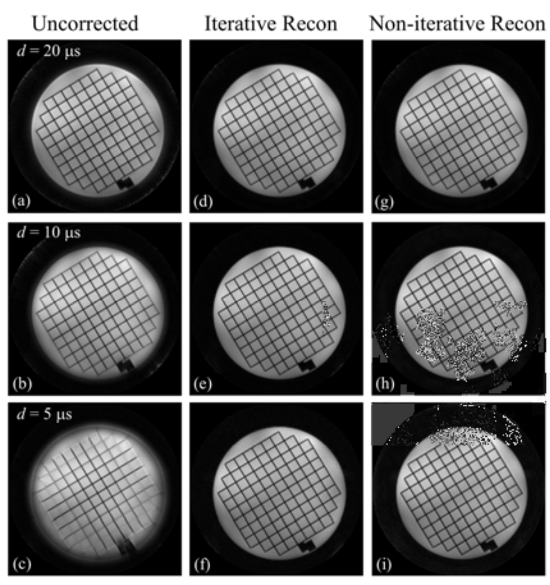

Fig.3 Uncorrected (a-c), corrected with iterative (d-f) and non-iterative (g-i) reconstruction phantom images with varying dwell times and constant pulse duration.

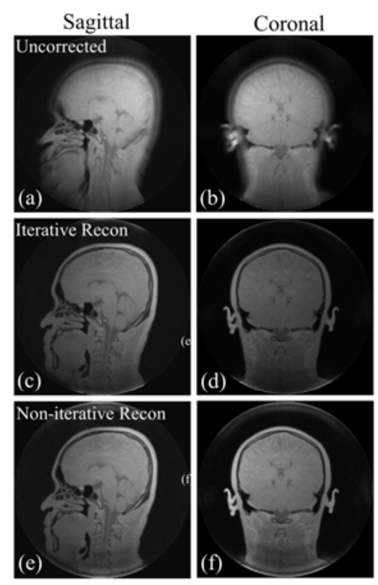

Fig.4 In vivo brain images without correction (a, b), with iterative(c, d) and non-iterative(e, f) reconstruction

Conclusions - In conclusion, a fast non-iterative reconstruction method for correcting artifacts caused by the heterogeneous excitation in ZTE imaging is presented. The new approach allows online ZTE reconstruction with artifact correction and may help establish ZTE as a routine pulse sequence for imaging short-T2 tissue constituents on clinical scanners.

References: 1. Wu Y et al. MRM2007:554-67; 2.Grodzki DM et al. MRM2012:510-518; 3. Grodzki DM et al. JMR2012:61-7. 4. Li C et al. IEEE TMI2014:961-9. **Acknowledgements:** NIH grants R21 NS082953, R01 AR50068, HHMI International Student Research Fellowship.