Introduction  During the past decade, ultra-short echo-time (UTE) MRI has evolved into a promising technique for directly imaging tissues with very short-T2* relaxation time on the order of hundreds of microseconds (1). In order to capture the fast-decay signal, effective echo times (TEs) in UTE sequences need to be on the order of tens of microseconds or less, which can be accomplished custom designed RF pulses, such as half or hard pulses, and sampling strategies, e.g. radial center-out trajectory. However, 2D UTE imaging is inherently time-inefficient: half-pulse excitation requires two scans with opposite slice-selection gradient polarities to achieve spatial selectivity; radial center-out sampling doubles the scan time for full k-space coverage. In this work, to improve UTE efficiency, we developed 3D Compressed Sensing (CS) (2) UTE (COMPUTE) imaging with a hybrid-radial encoding strategy (3), thereby achieving an acceleration factor of 10. Phantom and in vivo results demonstrated the feasibility to apply CS to UTE imaging.

Methods  Pulse Sequence Compared with 2D UTE, 3D UTE provides volume coverage and higher SNR efficiency. Since a hard pulse is employed, 3D UTE is also immune to artifacts associated with half-pulse excitation and obviates the need for two scans with opposite slice-selection gradient polarities. A hybrid stack-of-radial pattern, rather than pure 3D radial acquisition was chosen as the sampling trajectory for the following reasons: 1. It is more straightforward to achieve anisotropic sampling doubles the scan time for full k-space coverage. In order to improve UTE efficiency, we developed 3D Compressed Sensing (CS) (2) UTE (COMPUTE) imaging with a hybrid-radial encoding strategy (3), thereby achieving an acceleration factor of 10. Phantom and in vivo results demonstrated the feasibility to apply CS to UTE imaging.

Simulation  The undersampled data were synthesized by randomly sampling kx and variably undersampling the projection views. A power of 5 of distance from the kx center was chosen as the sampling probability density function to achieve an undersampling factor of 2 in the kx dimension. 250 equiangular views were selected in the central kx portion and 125 views in the edge kx region. With this undersampling strategy, a total acceleration factor of 10 was achieved.

CS Reconstruction  The images were reconstructed by solving the following optimization problem with total variation (TV) as the sparsity constraint (2): \[
\hat{m} = \arg \min_{m} \| F m - d \|_1 + \lambda \| D m \|_1
\] Here \(\hat{m}\) is the reconstructed image, \(y\) denotes the undersampled k-space data and \(\lambda\) is the regularization parameter. \(F\) represents the undersampled Fourier transform operator that maps the image onto the k-space data according to the sampling pattern in COMPUTE, and \(D\) is the finite difference operator. \(\| \cdot \|_p\) denotes the vector’s p-norm. A nonlinear Conjugate Gradient algorithm was used for solving Eq.1 (2). The images were also reconstructed from the undersampled data by zero-filling with density compensation (ZF-w/dc), which consists of zero-filling the missing k-space data, multiplying with k-space density compensation factor (DCF), IFFT along kx and NUFFT (4) on the kx–ky plane. DCF is computed from the probability density function to

Results  Axial phantom images reconstructed from full and undersampled data sets with CS and ZF-w/dc are shown in Fig. 2. The streaking artifacts are apparent with ZF-w/dc reconstruction (Fig. 2c). These artifacts are significantly reduced with CS reconstruction (Fig. 2b). To further compare the reconstruction accuracy, the signal profiles along the dashed lines indicated in Fig. 2a-c are plotted in Fig. 2d, demonstrating that the CS reconstruction recovered the signal from undersampled k-space data with high accuracy. Fig. 3 shows images of the mid-tibia of a 25-year-old male volunteer in axial and coronal planes with different reconstructions. The CS reconstructed images are still comparable to the fully sampled images but correspond to one sixth of the original scan time. Some smoothing effects are observed in the images from CS reconstruction. Optimization of the regularization parameter \(\lambda\) would further improve the CS reconstructed images.

Conclusion  COMPUTE, consisting of a custom-designed hybrid 3D UTE sequence, combined with CS reconstruction, has achieved an acceleration factor of 10 with no perceptible image quality degradation. In work in progress, long-T2* suppression modules are being incorporated into COMPUTE, allowing spatial sparsity of the soft-tissue suppressed images to be exploited to achieve even higher acceleration factors. We anticipate COMPUTE to be particularly beneficial to in vivo applications of 3D IR-UTE imaging, which provides the highest and most uniform short-T2* contrast, but currently suffers from impractically long scan time (5).