Published in final edited form as:

Magn Reson Med. 2016 September ; 76(3): 814–825. doi:10.1002/mrm.25977.

Ramped Hybrid Encoding for Improved Ultrashort Echo Time Imaging

Hyungseok Jang^{1,2}, Curtis N Wiens¹, and Alan B McMillan¹

¹Department of Radiology, Wisconsin Institute for Medical Research, University of Wisconsin, Madison, WI 53705, USA

²Department of Electrical and Computer Engineering, University of Wisconsin, Madison, WI 53706, USA

Abstract

Purpose—We propose a new acquisition to minimize the per-excitation encoding duration and improve the imaging capability for short T_2^* species.

Method—In the proposed ramped hybrid encoding (RHE) technique, gradients are applied before the RF pulse as in pointwise encoding time reduction with radial acquisition (PETRA) and Zero TE (ZTE) imaging. However, in RHE, gradients are rapidly ramped after RF excitation to the maximum amplitude to minimize encoding duration. To acquire central k-space data not measured during RF deadtime, RHE utilizes a hybrid encoding scheme similar to PETRA. A new gradient calibration method based on single-point imaging was developed to estimate the k-space trajectory and enable robust and high quality reconstruction.

Result—RHE enables a shorter per-excitation encoding time and provides the highest spatial resolution among ultrashort T_2^* imaging methods. In phantom and in vivo experiments, RHE exhibited robust imaging with negligible chemical shift or blurriness caused by T_2^* decay and unwanted slice selection.

Conclusion—RHE allows the shortest per-excitation encoding time for ultrashort T_2^* imaging, which alleviates the impact of fast T_2^* decay occurring during encoding, and enables improved spatial resolution.

Keywords

Hybrid Encoding; UTE; PETRA; ZTE; Gradient Calibration; Single Point Imaging

Introduction

MR imaging of objects with extremely short transverse relaxation (T_2^*) times such as bone(1–4), brain(5,6), lung(7–9), or teeth(10,11) is challenging due to the rapid signal decay of these tissues and the physical limitations of MR hardware. Particularly, the performance

Address for correspondence: Alan B McMillan, Associate Scientist, Department of Radiology, University of Wisconsin – Madison, Rm 1111, Wisconsin Institutes for Medical Research, 1111 Highland Avenue, Madison, WI 53705-2275, abmcmillan@wisc.edu, Phone: (608) 262-2622, Fax: (608) 265-9840.

of the gradient system is limited in slew rate and amplitude and thus is a critical factor in the design of Ultra-short TE (UTE) acquisitions. Conventional nonselective frequency-encoded 3D UTE (FE-UTE) methods utilize a radial trajectory to rapidly frequency encode k-space, by encoding the free induction decay as rapidly as possible in a “center-out” acquisition(12). In these methods, data encoding must wait for the signal to recover from the transmitter/receiver switching time (deadtime) to obtain non-corrupted central k-space data. However, overall encoding time is not optimal because the gradient must be ramped from zero to the maximum amplitude after deadtime.

Other techniques use non-zero gradients during RF excitation to reduce the achievable echo time. These methods include single point imaging (SPI)(13) and related techniques such as Single-Point Ramped Imaging with T₁ Enhancement (SPRITE)(14–17), Back-projection Low Angle ShoT (BLAST)(18), Rotating Ultra-Fast Imaging Sequence (RUFIS)(19), and Water And fat Suppressed Projection MR Imaging (WASPI)(20). Although these imaging schemes are simple and allow reduced echo time imaging, data cannot be collected during the receiver deadtime, thus complicating acquisition of the central regions of k-space and limiting maximum gradient amplitude. To address this issue, methods such as Zero TE (ZTE)(3,4,9–11,21), Pointwise Encoding Time Reduction With Radial Acquisition (PETRA)(22), and Sweep Imaging with Fourier Transform (SWIFT)(23–25) have been proposed. SWIFT utilizes gapped frequency-swept pulses across the gradient bandwidth. ZTE and PETRA are highly similar, ZTE utilizes an algebraic reconstruction to estimate missing central regions of k-space and PETRA utilizes SPI to encode the central regions of k-space. Unfortunately, the maximum gradient amplitude in ZTE and PETRA is limited by unwanted slice selectivity due to bandwidth constraints of the RF pulse(21,26,27).

In PETRA/ZTE, gradients are set to the maximum encoding amplitude, $G_{\max}$, before the application of a short, high-bandwidth RF pulse, as depicted in the pulse sequence diagram shown in Figure 1-a, to save the time required to ramp gradients and thereby shorten total encoding time. However, the non-zero gradient during RF excitation results in an unwanted slice selection effect, where the magnetization is not uniformly flipped, but subject to a non-uniform (e.g., sinc-shaped) excitation profile as shown in Figure 1-b. The effective orientation and width of the slice selection changes according to the orientation and amplitude of the encoding gradients. Therefore, encoded k-space data experiences a different slice selection, resulting in blurring and artifacts in the reconstructed image, manifested radially from the gradient isocenter. This artifact gets stronger as $G_{\max}$ increases, and a larger region of the image is affected by the blurriness.

This blurriness can be avoided by simply using an encoding gradient with low amplitude. However, in that case longer encoding time is required to achieve the desired spatial resolution, resulting in two significant limitations. First, the long encoding time reduces the spatial resolution of short T₂* species(28), resulting in blurring and loss of detail in the very components that are being imaged. Second, this results in chemical shift artifacts of the second kind (intravoxel fat-water interference), which results in an out-of-phase appearance, particularly at 3T and above. Shorter RF pulses with higher bandwidths can be used to alleviate the slice selection artifact; however, this limits the maximum attainable flip angle and thus reduces SNR and the capability to achieve T₁-weighted contrast. Several methods

have been proposed to address the slice selection problem by performing post processing or utilizing hyperbolic secant RF pulses(21,26,27).

In this study, we have developed a new encoding scheme, termed ramped hybrid encoding (RHE), which allows reduced per-excitation encoding time and minimized slice selectivity effects to improve the sharpness of high resolution UTE imaging. In RHE, gradients are held at low amplitude (e.g., below 7mTm^{-1} with a $24\mu\text{s}$ hard pulse for field of view (FOV)=200mm) during RF excitation to minimize slice selectivity, and ramped to their maximum amplitude immediately following RF pulse. A 1D SPI-based gradient calibration method was developed to estimate the k-space trajectory of the encoding gradients. The efficacy of RHE was evaluated by comparing it to other UTE imaging schemes in computer simulation, phantom, and in vivo experiments.

Methods

Ramped Hybrid Encoding

We propose RHE as a technique to allow the greatest flexibility compared to currently available methods in controlling unwanted slice selectivity while optimizing overall encoding time for UTE imaging. Figure 2-a shows the pulse sequence diagram for RHE. In RHE, an initial gradient during RF excitation, G_{RF} , is chosen to be small enough to minimize slice selectivity (by considering the limitations of the frequency profile of the RF pulse). After application of the RF pulse, the gradient is ramped to the maximum encoding amplitude, G_{max} , at the highest slew rate possible to minimize overall sampling duration. Data are acquired after RF deadtime until the desired spatial resolution is achieved.

As in PETRA, RHE utilizes SPI to measure data in central k-space regions that frequency encoding omits during RF deadtime. Central k-space is encoded by Cartesian SPI, and the outer k-space is acquired by frequency encoding as shown in Figure 2-b. Note that in Figure 2-a the solid line in the pulse sequence diagram (PSD) shows the gradient amplitude along the readout direction used to scan half radial spokes (blue arrows in Figure 2-b). The readout gradient is then rotated at each TR to frequency-encode k-space. In Cartesian SPI sampling (red dots in Figure 2-b), the maximum gradient amplitude is linearly scaled as dotted lines in Figure 2-a shows to encode different k-space point at the constant encoding time over TRs. Note that the same maximum gradient is applied to both SPI and frequency encoding to prevent discontinuity in encoding times at the interface between the two different encoding schemes. This acquisition can also be extended to multi-echo acquisitions, where Figure 2-c shows the pulse sequence used to obtain multi-echo RHE images with 5 half-echoes obtained within a single acquisition. Note that SPI encoding is used to fill the central region of k-space in each half-echo.

In RHE, the diameter of the SPI-encoded region in k-space, N_{SPI} , is determined by the following equation.

$$N_{\text{SPI}} = \lceil 2\bar{\gamma} \text{fov}_D (G_{\text{RF}} t_D + 0.5 g_s t_D^2) \rceil \text{ if } t_D \leq (G_{\text{max}} - G_{\text{RF}}) / g_s, \lceil 2\bar{\gamma} \text{fov}_D (G_{\text{max}} t_D - 0.5 (G_{\text{max}} - G_{\text{RF}})^2 / g_s) \rceil \text{ otherwise,}$$

(1)

where $\bar{\gamma}$ is the gyromagnetic ratio in unit of HzT^{-1} , g_s denotes gradient slew rate in units of $\text{Tm}^{-1}\text{s}^{-1}$, fov_D denotes the desired FOV, and t_D is the desired echo time chosen after deadtime. Due to eddy currents that effectively derate the gradients in ramping, SPI data are prone to be slightly oversampled and hence result in a larger FOV than the desired FOV (fov_D) at the desired TE (t_D). The FOV can be corrected in the reconstruction stage using conventional convolution gridding methods. In practice, larger N_{SPI} can be intentionally used to obtain oversampled SPI data allowing some flexibility in selecting TE when RF deadtime is not known a priori.

The maximum gradient amplitude during RF excitation, G_{RF} , can be selected by considering both slice selectivity and N_{SPI} . An upper bound for G_{RF} can be analytically determined using the expected RF pulse shape and its frequency profile. However, large G_{RF} amplitudes may result in impractical scan times due to the large required N_{SPI} . In that case, G_{RF} needs to be reduced to allow reasonable scan times. The maximum readout gradient, G_{max} , can be as large as possible within the constraints of the readout bandwidth and safety factors such as gradient heating and peripheral nerve stimulation.

Gradient calibration

In RHE, data is acquired during ramping gradients. Therefore, timing errors and eddy current effects may distort the k-space sampling trajectory, and hence naïve reconstruction based on the prescribed gradient parameters is generally not suitable. In this study, we developed a rapid new calibration method that benefits from the well-known zoom-in effect (decreasing FOV with increasing phase encoding time delay) in SPI(15,16,29), based on a similar concept to previous work by Balcom et. al(30). For calibration, three sets of 1D projection images are acquired using an SPI scheme in each gradient axis. Note that 1D SPI imaging can be easily implemented in any pulse sequence by scaling the gradients to enable pure phase encoding. Typically this calibration data can be acquired very rapidly, within several seconds for all gradient axes.

1D single point images can be reconstructed without calibration. The three sets of 1D projection images across a range of encoding time are reconstructed at native FOVs (exploiting the zoom-in effect), as depicted in Figure 3. The image matrix shown in Figure 3-a contains 1D projection images (y-axis) versus phase encoding time delay (x-axis) for a gradient direction encoded by 1D single point imaging. The size of the object (bright region in center of FOV) increases with encoding time (zoom-in effect). The speed of the FOV change in Figure 3-a is directly proportional to the gradient strength shown in Figure 3-b, exhibiting acceleration in ramping up, constant change in plateau, and deceleration in ramping down. Therefore, the gradient waveform can be calibrated by directly computing the scaling factors between neighboring phase encoding time delays within the 1D images.

The FOV scaling factors between images are found automatically using unconstrained nonlinear optimization (Nelder-Mead Simplex). A reference image is first selected as the latest time delay, t_{ref} , and the relative scaling factors between t_{ref} and other time delays are found by minimizing the L2-norm of the error function as shown in the following equation.

$$\text{FOV scale}(t) = \text{FOV}(t_{\text{ref}}) / \text{FOV}(t) = \underset{s}{\text{argmin}} \sqrt{\sum_{x=1}^N |I(t_{\text{ref}}, x) - I(t, s(x - N_c) + N_c)|^2} \quad (2)$$

, where $I(t, x)$ denotes magnitude of 1D image at encoding time t and spatial position x , N is 1D matrix size, s is a scaling factor between images, and N_c is index for the center of image (e.g., for matrix size= N , $N_c = \lfloor N/2 \rfloor$). Images are transformed based on the scaling factor, s , to find the best scaling factor. This scaling transformation can be performed in the either image or k-space domain by using an affine transform or convolution gridding, respectively. In this study, the transform was performed in the image domain using bilinear interpolation because it provided reliable results that could be computed much faster than using a gridding approach.

Once proper FOV scaling factors are found across all phase encoding time delays, relative k-space position can be recovered. Note that the FOV scaling factors only describe the relative scaling difference between encoding times. To obtain the absolute FOV, we examine the RHE data acquired during constant gradient. First, the slope of FOV scaling factors is calculated at a time (t_{ref}) when the gradient is constant (and known) G_{max} . Then, the slope can be used to calculate the true FOV at the reference encoding time, t_{ref} , using following equation.

$$\text{FOV}(t_{\text{ref}}) = c(N - 1) / (2\gamma G_{\text{max}}) \quad (3)$$

, where c is slope of FOV scaling factor found at constant gradient. Now, the FOV for the entire encoding time, $\text{FOV}(t)$, can be recovered by simply using equation 2 with the given $\text{FOVscale}(t)$ and $\text{FOV}(t_{\text{ref}})$.

Image reconstruction

After the k-space trajectories are estimated via the above gradient calibration method, the acquired SPI and radial data are combined together. 3D convolution gridding is applied to obtain the k-space with desired FOV(31–33). To control variable density sampling within k-space, iterative density compensation(34) is applied. Note that the sampling density along a half radial spoke in frequency encoding sampling density is dependent upon the readout bandwidth and the shape of the encoding gradient, while within the SPI region it is determined by TE and the shape of the encoding gradient. In Figure 4, a block diagram shows how raw data are processed to obtain a final RHE image.

Computer simulation

To compare encoding times and the resultant image quality between RHE and other UTE imaging schemes, a 1D computer simulation was performed. Note that a conventional point spread function (PSF) simulation is not possible because the PSF is spatially-variant at each encoding position in k-space, as described above. Therefore, each point in k-space was independently simulated with different slice selectivity resultant from the differing encoding gradient amplitude, as described above. To generate the 1D digital phantom, 11 tubes were generated with different proton densities, 0.7, 1.0, 0.7, 0.3, 0.7, 1.0, 0.7, 0.3, 0.7, 1.0 and 0.7 in arbitrary units from left to right. The diameter of each tube was 40mm. A mono-exponential T_2^* decay model ($T_2^* = 100\mu\text{s}$ or $500\mu\text{s}$) was simulated for all tubes.

System parameters included a TE of $80\mu\text{s}$, a slewrate of $118\text{ mTm}^{-1}\text{ms}^{-1}$, and a maximum gradient of 35 mTm^{-1} . For PETRA $G_{\text{max}} = 7$ or 20 mTm^{-1} was used. For RHE, $G_{\text{RF}} = 3.5$ or 7 mTm^{-1} and $G_{\text{max}} = 35\text{ mTm}^{-1}$ was used. For RHE and FE-UTE, the gradients were ramped immediately after the RF pulse or after deadtime, respectively. 1D sampling was simulated using frequency encoding or PETRA/RHE encoding to acquire a 500×1 k-space with FOV=500mm, which achieves approximately 1mm resolution. For PETRA and RHE, the slice selectivity effect was simulated using the spatial profiles of the $24\mu\text{s}$ hard pulse shown in Figure 1-b. N_{SPI} was set to the minimum value according to the prescribed gradient shape ($N_{\text{SPI}}=24, 69, 29$, and 40 respectively for PETRA with $G_{\text{max}}=7\text{ mTm}^{-1}$, PETRA with $G_{\text{max}}=20\text{ mTm}^{-1}$, RHE with $G_{\text{RF}}=3.5\text{ mTm}^{-1}$, and RHE with $G_{\text{RF}}=7\text{ mTm}^{-1}$). No eddy current effects were applied in the computer simulation.

Experimental setup

To evaluate the proposed encoding scheme, MR experiments were performed on a 3.0T MR scanner (MR750, GE Healthcare, Waukesha, WI). A phantom experiment was performed to compare UTE imaging schemes (PETRA, FE-UTE, and RHE) with an object that only has short T_2^* components. Human brain imaging was performed with 2 different RF pulses ($8\mu\text{s}$, $24\mu\text{s}$) with flip angle 2° and 6° respectively) and gradient settings. A multi-echo RHE experiment to generate a short T_2^* image was performed in the human knee.

For phantom experiments, a phantom made of Acrylonitrile Butadiene Styrene (ABS) plastic (Big Ben, item # 21013, a cowboy minifigure from Palace Cinema, item # 10232, and a white horse made by LEGO, Billund, Denmark) with T_2^* approximately $400\text{--}500\mu\text{s}$. An 8-ch receive-only head coil (High Resolution Brain Coil, GE Healthcare) was used for the phantom experiment. For in vivo experiments, a human subject was imaged in accordance with local IRB protocols. The 8-ch receive-only head coil was used for in vivo brain imaging, and an 8-ch transmit-receive knee coil (GE Precision Eight Knee Array Coil, Invivo, Gainesville, Florida) was used for in vivo knee imaging.

All parameters used for the phantom, knee, and brain imaging are shown in Table 1. A single echo acquisition (as shown in Figure 2-a) was performed in phantom and brain imaging, while multi-echo imaging (as shown in Figure 2-c) was performed in the knee. For all datasets, the TE was $90\mu\text{s}$, which is defined as the first encoding time after which the receiver is fully recovered from RF deadtime (as shown in Figure 3-a). Deadtime was

determined empirically by observing the signal magnitude at the center of k-space. A sampling period of $2\mu\text{s}$ was used. In the phantom and knee imaging experiment comparing PETRA, FE-UTE, and RHE, N_{SPI} and TR were set identically to allow reasonable comparisons between imaging schemes. N_{SPI} was set to 33, the largest N_{SPI} required by PETRA with largest $G_{\text{max}} (=20\text{mTm}^{-1})$, while TR was set to 3.3ms, is the minimum TR of PETRA with lowest encoding gradient ($G_{\text{max}}=7\text{mTm}^{-1}$). Due to the reduced encoding duration, the minimum possible TR for RHE can be significantly shorter (approximately 2ms).

To perform gradient calibration, three 1D 401×1 SPI images were acquired along each physical gradient axis by linearly scaling the encoding gradient over TRs (401 equispaced steps between $-1.0\times$ and $1.0\times$ of gradient shape to calibrate). The additional scan time required for the calibration was $401(\text{encodings/axis}) \times 3(\text{axis}) \times \text{TR}$, which was 4 sec for the single echo acquisition and 6.7 sec for the multi-echo acquisition. For a more reliable calibration, SPI-based calibration was first performed using a spherical phantom (in a separate imaging session on a separate day and only once for all experiments), which was then used as the initial guess during calibration. The proposed SPI-based calibration was applied to both FE-UTE and RHE imaging.

During image reconstruction, convolution gridding was performed using a Kaiser-Bessel kernel with grid width=5 (for phantom and head imaging) or 7 (for knee imaging) and oversampling ratio=2. Phantom data were gridded to achieve FOV=200mm and matrix size of $201\times 201\times 201$, and brain data were gridded to achieve FOV=240mm and matrix size of $241\times 241\times 241$, which is equivalent to 1mm resolution. In the knee experiment five 3D knee images were reconstructed at TE=90 μs , 1502 μs , 1550 μs , 2900 μs , and 2950 μs with FOV=200mm and a matrix size of $401\times 401\times 401$, which is equivalent to 0.5 mm resolution. Separate fat and water images were computed using Iterative Decomposition of water and fat with Echo Asymmetry and Least-squares estimation (IDEAL)(35). Because we do not expect the short T_2^* species to have different image phase, all five images were used for the IDEAL reconstruction using a non- R_2^* corrected model. The image representative of short T_2^* species was obtained by subtracting the computed water and fat images from the RHE image at TE=90 μs .

Gradient calibration, gridding, and IDEAL reconstruction were performed using MATLAB (The Mathworks, Natick, MA) on a Linux computer with an Opteron 6134 processor (64 bit, 16 core).

Results

Simulation results

Figure 5 shows the simulated curves for the per-excitation encoding time in three different UTE encoding schemes, conventional FE-UTE, PETRA, and RHE, (Figure 5-a) and the corresponding reconstructed images (Figure 5-b,c). As seen in Figure 5-a, RHE with $G_{\text{RF}}=7\text{mTm}^{-1}$ allows the shortest per-excitation encoding time ($=429\mu\text{s}$) between the three methods (1669 μs for PETRA with $G_{\text{max}}=7\text{mTm}^{-1}$, 584 μs for PETRA with $G_{\text{max}}=20\text{mTm}^{-1}$, 562 μs for FE-UTE, and 454 μs for RHE with $G_{\text{RF}}=3.5\text{mTm}^{-1}$) while controlling for blurring

caused by T_2^* or the finite RF pulse duration. The reconstructed images with normalized scales are shown in Figure 5-b,c. Root Mean Squared Error (RMSE) was calculated to compare the simulated images (with normalized scales) to the 1D digital phantom.

When T_2^* is extremely short ($100\mu\text{s}$), RHE with $G_{\text{RF}}=3.5\text{mTm}^{-1}$ provides the most accurate reconstruction ($\text{RMSE}=0.061$) owing to its optimized encoding time and controlled slice selectivity. PETRA images show good fidelity at the center of the FOV, but exhibit loss of detail toward the edges due to the unwanted slice selectivity imposed by high encoding gradients applied during RF excitation. Note that PETRA with $G_{\text{max}} = 20\text{mTm}^{-1}$ provides good reconstruction at the center of the FOV owing to the large $N_{\text{SPI}} (=69)$ where encoding time is constant ($=\text{TE}$), resulting in less intra-readout T_2^* decay. However, a larger N_{SPI} significantly increases the total image acquisition time and is not clinically feasible. FE-UTE shows uniformly reasonable results over the entire FOV as expected. When T_2^* is moderately short ($=500\mu\text{s}$), FE-UTE shows the overall best reconstruction ($\text{RMSE}=0.025$), while RHE with $G_{\text{RF}}=3.5\text{mTm}^{-1}$ shows a comparably accurate reconstruction ($\text{RMSE}=0.046$), particularly in the central region of the FOV.

Gradient calibration

Figure 6-a shows 1D projections from the x-axis of a calibration dataset reconstructed at the native FOVs and exhibiting the zoom-in effect (decreasing FOV with increasing phase encoding time delay). Figure 6-b shows the FOV scaling factors found between images in x-direction where 'x' shows five FOV scaling factors corresponding to the five images in Figure 6-a. Figure 6-c shows the calibrated trajectory along 3 gradient orientations. Note that the estimated trajectories are different between gradient axes. Figure 6-d shows the measured trajectory, the prescribed trajectory, and delay-corrected trajectory obtained in the physical z-gradient direction. The delay-corrected trajectory was obtained by matching the linear part of the prescribed k-space trajectory with the measured trajectory. The two superimposed curves for measured trajectory and the delay-corrected trajectory show little difference in the ramping portion of the encoding gradient. As seen in the images reconstructed with the three different k-space trajectories in Figure 6-e, small errors result in significant and obvious reconstruction error as shown in the region yellow arrow indicates, which shows misalignment between low and high frequency component in image due to the erroneous gradient calibration.

Estimation of the k-space trajectory could be computed rapidly. The proposed method required approximately 2 sec to process one image, which equates to $2 \text{ (sec/image)} \times 230 \text{ (images)} / 12 \text{ (# of parallel computation)} \approx 38 \text{ sec}$ for single echo imaging and $2 \text{ (sec/image)} \times 1,630 \text{ (image)} / 12 \text{ (# of parallel computation)} \approx 272 \text{ sec}$ for multi-echo imaging.

Phantom experiment

Figure 7 shows the results of the phantom experiment. Note that in the reconstructed images, RHE (Figure 7-e) preserves the high frequency details of the phantom much better than PETRA (Figure 7-a,b,c) and FE-UTE (Figure 7-d).

PETRA with $G_{\text{max}} = 7 \text{ mTm}^{-1}$ (Figure 7-a) exhibits severe blurriness across the image, due to the long encoding time. PETRA with $G_{\text{max}} = 14 \text{ mTm}^{-1}$ (Figure 7-b) shows an improved

depiction of the object, but exhibits blurriness along the radial direction at the edge of the FOV, which is due to unwanted slice selectivity. PETRA with $G_{\max} = 20 \text{ mTm}^{-1}$ (Figure 7-c) shows the best spatial resolution in the center of the FOV and the worst slice selectivity artifact at the corners of the FOV due to the large gradient applied during RF excitation. FE-UTE (Figure 7-d) shows a good depiction of the object with no slice selectivity artifact. RHE (Figure 7-e) shows higher detail owing to the shorter per-excitation encoding time and the central k-space encoded by SPI (with TE equal to $90\mu\text{s}$). The SNR of RHE (measured SNR=10.3) is higher than FE-UTE (measured SNR=8.7).

In vivo - knee imaging

Figure 8 shows coronal or sagittal slices of knee images at five different TEs obtained using RHE with multi echo imaging capability (Figure 8-a,b,c,d,e,i,j,k,l,m), water images (Figure 8-f,n) and fat images (Figure 8-g,o) obtained using IDEAL, and the resultant short T_2^* images (Figure 8-h,p). Note that short T_2^* images, tissues such as bone, tendon, and ligament are visible with positive contrast. In the coronal plane short T_2^* image (Figure 8-h), the medial collateral ligament and lateral collateral ligament (white arrow) and the medial meniscus (yellow arrow) are visible. In the sagittal plane short T_2^* image (Figure 8-p), the quadriceps femoris tendon (blue arrow), patellar ligament (green arrow), and anterior cruciate ligament (red arrow) are seen clearly.

In vivo - Brain imaging

Figure 9 shows brain images obtained by PETRA, FE-UTE, and RHE with two different RF pulse lengths and for PETRA, three different readout gradient amplitudes. The left 4×5 image matrix shows 2D slices selected from the reconstructed 3D images, and the right 4×5 image matrix shows corresponding zoomed-in images. Figure 9-a to Figure 9-j show images at a mid-sagittal plane, while Figure 9-k to Figure 9-t show images at an axial plane. Figure 9-a,b,c,k,l,m and Figure 9-f,g,h,p,q,r show PETRA images obtained with $8\mu\text{s}$ and $24\mu\text{s}$ respectively. In PETRA with a short RF pulse ($8\mu\text{s}$), slice selectivity is suppressed owing to the relatively broad excitation bandwidth, but SNR is reduced due to the smaller attainable flip angle. Note that with an $8\mu\text{s}$ RF pulse, images are more proton density weighted, while with a longer RF pulse ($24\mu\text{s}$) increased T_1 weighting can be achieved. With the $24\mu\text{s}$ RF pulse, the slice selectivity artifact is more noticeable in PETRA, substantially deteriorating with higher $G_{\max}$ (Figure 9-g,h,q,r), while the chemical shift artifact is aggravated as encoding time decreases with lower $G_{\max}$ (Figure 9-a,f,k,p).

Compared with PETRA, both FE-UTE (Figure 9-d,i,n,s) and RHE (Figure 9-e,j,o,t) show much better image quality with higher spatial resolution and no or minimal slice selectivity artifact respectively. However, as shown in the zoomed-in sagittal images of FE-UTE in Figure 9-d,i and RHE in Figure 9-e,j, RHE exhibits higher signal intensity than FE-UTE in occipital bone indicated by the yellow arrow. In addition, RHE shows higher signal in a tooth as indicated by the green arrow in Figure 9-i,j. In axial images, both RHE and FE-UTE show detailed views of the tissues in the sinuses as shown in zoomed-in images of Figure 9-s,t. Note that the red arrow indicates in Figure 9-s,t RHE shows higher signal intensity than FE-UTE in the region where the meninges are visible.

Overall, among the UTE imaging schemes presented here, RHE shows the highest spatial resolution, best short T_2^* contrast, no apparent chemical shift artifact owing to the shortest per-excitation encoding time, and well-controlled slice selectivity with a $8\mu\text{s}$ or $24\mu\text{s}$ RF pulse.

Discussion

In this study, we proposed a new scheme, termed RHE, for time-optimal per-excitation encoding in UTE imaging. Reductions in encoding duration improve the spatial resolution for short T_2^* species(28) and reduce chemical shift artifacts. Moreover, the ability to control G_{RF} and the resultant slice selectivity allows greater flexibility regardless of the desired FOV for UTE imaging. While a high G_{RF} is desired to shorten encoding time, there exists an upper limit of G_{RF} to avoid objectionable slice selectivity artifacts. Increasing the bandwidth of the RF pulse (e.g., using a shorter RF pulse) increases the attainable G_{RF} for good quality image; however, shorter RF pulses also limit the maximally attainable flip angle and thus can reduce SNR and/or desired T_1 image contrast. Gradient modulation during RF has been shown to be useful in other sequences, such as SWIFT, to reduce specific absorption rate(SAR)(25).

The use of higher amplitude encoding gradients to attain faster encoding inevitably requires a larger SPI encoded region (N_{SPI}) for hybrid encoding techniques such as PETRA or RHE. As seen in Figure 5, increased N_{SPI} improves short T_2^* imaging, with the caveat of substantially increasing total scan time. Reduction of N_{SPI} can harm spatial resolution for species where the T_2^* is short relative to the per-excitation encoding duration (e.g., Figure 5-b, FE-UTE vs. RHE [$N_{\text{SPI}}=29$ or 40] with similar encoding durations). Therefore, it may be beneficial to prescribe RHE with a reasonably large N_{SPI} to balance between the beneficial qualities of a bigger SPI region and total imaging time. Note that the contribution of SPI encoding to spatial resolution may be more significant in the 1D simulation than 2D or 3D since 1D radial acquisition is more susceptible to T_2^* decay than 2D or 3D radial acquisition(28).

Due to the shorter encoding time, sampling density along a radial spoke in the frequency encoded region is reduced in RHE compared to PETRA/ZTE, which penalizes the SNR for long T_2^* components. However, in spite of the SNR advantage that slow encoding allows, longer encoding results in overall degradation in image quality (loss of spatial detail for short T_2^* components and chemical shift artifacts). Thus, the reduced readout duration for RHE is important for improving image quality for UTE imaging. Indeed, there is no other encoding strategy to reach the extent of k-space in a more time-efficient manner than RHE when B_1 limitations prohibit the desired flip angle and field of view. If additional SNR is necessary, traditional techniques such as increased averaging or optimal coil configuration would apply.

Additional improvements in RHE image quality and functionality are possible. For example, post processing strategies proposed to correct slice selectivity in ZTE and PETRA(21,26,27) can be employed to alleviate blurriness artifacts. Moreover, hybrid encoding schemes with an oversampled SPI encoded region may allow reconstruction of images at multiple TEs in

early encoding times, which can be used to estimate short T_2^* parameters with a single experiment (e.g., using k-space extrapolation methods as recently proposed in single point electron paramagnetic resonance imaging(36)). The additional scanning time imposed by oversampling SPI encoded region can be reduced by using variable density sampling pattern and appropriate reconstruction method such as compressed sensing using k-space domain data(37,38) or model-based reconstruction using k-space domain data and free induction decay (FID) data (parameter domain) simultaneously(39,40).

Recently, there has been significant interest in developing sequences with extremely low acoustic noise levels(23,41–43). Using RHE, the acoustic noise will be higher; however, the encoding duration will always be lower compared to B_1 limited PETRA/ZTE encoding strategies. Thus, quiet scanning may be incompatible with high quality imaging of short T_2^* species. For RHE, lower noise scanning could be achieved at the cost of reduced encoding performance.

Suppression of long T_2^* species was demonstrated by incorporating a multi-echo RHE acquisition with a chemical shift encoding technique (IDEAL). In the current implementation, all 5 half-echoes were used to estimate long T_2^* fat and water signals. Although inclusion of the first echo may lead to underestimation of the calculated short T_2^* species, stability of the fat and water separation was improved. Qualitatively this bias appears to be small as short T_2^* species could readily be identified. Future work to optimize the acquisition (e.g., echo spacing and number of echoes) and modeling techniques (e.g., modeling of R_2^*) may reduce this and other biases.(44–46)

Gradient calibration is essential to avoid distortion and reconstruction errors in resultant images. In this study, we implemented a new gradient calibration method based on 1D SPI. Unlike other techniques, the method is not dependent upon 2D slice selection(47) or external hardware(48), and the identical pulse sequence can be used with very minimal modifications (only different encodings need to be obtained). The scanning time required for calibration is also very short (well less than 10 seconds for the datasets herein), allowing robust estimation of k-space trajectory on a per-scan basis. This technique is also likely to be useful to measure the k-space trajectory of other pulse sequences in MRI, and is only limited by the number of encoding single-point steps. This paper describes preliminary use of this new calibration technique, and further development is planned in future studies.

Conclusion

In summary, we have proposed a new encoding technique that allows flexible and time-optimal encoding for short T_2^* species. In addition, we developed a new image-based calibration technique using single-point encoding to measure the k-space trajectory for improved image reconstruction.

Acknowledgments

The research reported in this publication was supported by the National Institute of Biomedical Imaging and Bioengineering of the National Institutes of Health under Award Number 1R21EB013770. The content is solely the responsibility of the authors and does not necessarily represent the official views of the National Institutes of Health. Additional research support from the University of Wisconsin Department of Radiology and GE Global

Research is acknowledged. We thank Diego Hernando, PhD for helpful discussions and supplying the IDEAL code used herein.

References

1. Wurnig MC, Calcagni M, Kenkel D, Vich M, Weiger M, Andreisek G, Wehrli FW, Boss A. Characterization of trabecular bone density with ultra-short echo-time MRI at 1.5, 3.0 and 7.0 T--comparison with micro-computed tomography. *NMR Biomed.* 2014; 27:1159–66. DOI: 10.1002/nbm.3169 [PubMed: 25088271]
2. Robson MD, Bydder GM. Clinical ultrashort echo time imaging of bone and other connective tissues. *NMR Biomed.* 2006; 19:765–80. DOI: 10.1002/nbm.1100 [PubMed: 17075960]
3. Weiger M, Stämpf M, Pruessmann KP. Direct depiction of bone microstructure using MRI with zero echo time. *Bone.* 2013; 54:44–7. DOI: 10.1016/j.bone.2013.01.027 [PubMed: 23356986]
4. Wiesinger F, Sacolick LI, Menini A, Kaushik SS, Ahn S, Veit-haibach P, Delso G, Shanbhag DD. Zero TE MR Bone Imaging in the Head. *Magn Reson Med.* 2015; 00:n/a–n/a. doi: 10.1002/mrm.25545
5. Waldman, a; Rees, JH.; Brock, CS.; Robson, MD.; Gatehouse, PD.; Bydder, GM. MRI of the brain with ultra-short echo-time pulse sequences. *Neuroradiology.* 2003; 45:887–92. DOI: 10.1007/s00234-003-1076-z [PubMed: 14508620]
6. Du J, Ma G, Li S, Carl M, Szeverenyi NM, VandenBerg S, Corey-Bloom J, Bydder GM. Ultrashort echo time (UTE) magnetic resonance imaging of the short T2 components in white matter of the brain using a clinical 3T scanner. *Neuroimage.* 2014; 87:32–41. DOI: 10.1016/j.neuroimage.2013.10.053 [PubMed: 24188809]
7. Kuethe DO, Caprihan A, Fukushima E, Waggoner RA. Imaging lungs using inert fluorinated gases. *Magn Reson Med.* 1998; 39:85–88. DOI: 10.1002/mrm.1910390114 [PubMed: 9438441]
8. Johnson KM, Fain SB, Schiebler ML, Nagle S. Optimized 3D ultrashort echo time pulmonary MRI. *Magn Reson Med.* 2013; 70:1241–1250. DOI: 10.1002/mrm.24570 [PubMed: 23213020]
9. Gibiino F, Sacolick L, Menini A, Landini L, Wiesinger F. Free-breathing, zero-TE MR lung imaging. *Magn Reson Mater Physics, Biol Med.* 2014; doi: 10.1007/s10334-014-0459-y
10. Weiger M, Pruessmann KP, Bracher AK, Köhler S, Lehmann V, Wolfram U, Hennel F, Rasche V. High-resolution ZTE imaging of human teeth. *NMR Biomed.* 2012; 25:1144–51. DOI: 10.1002/nbm.2783 [PubMed: 22290744]
11. Hövener JB, Zwick S, Leupold J, Eisenbeiß AK, Scheifele C, Schellenberger F, Hennig J, Elverfeldt DV, Ludwig U. Dental MRI: imaging of soft and solid components without ionizing radiation. *J Magn Reson Imaging.* 2012; 36:841–6. DOI: 10.1002/jmri.23712 [PubMed: 22707436]
12. Tyler DJ, Robson MD, Henkelman RM, Young IR, Bydder GM. Magnetic resonance imaging with ultrashort TE (UTE) PULSE sequences: technical considerations. *J Magn Reson Imaging.* 2007; 25:279–89. DOI: 10.1002/jmri.20851 [PubMed: 17260388]
13. Emid S, Creyghton JHN. High resolution NMR imaging in solids. *Phys B+C.* 1985; 128:81–83. DOI: 10.1016/0378-4363(85)90087-7
14. Balcom BJ, Macgregor RP, Beyea SD, Green DP, Armstrong RL, Bremner TW. Single-Point Ramped Imaging with T1 Enhancement (SPRITE). *J Magn Reson Ser A.* 1996; 123:131–134. DOI: 10.1006/jmra.1996.0225 [PubMed: 8980075]
15. Halse M, Goodyear DJ, MacMillan B, Szomolanyi P, Matheson D, Balcom BJ. Centric scan SPRITE magnetic resonance imaging. *J Magn Reson.* 2003; 165:219–229. [PubMed: 14643703]
16. Kaffanke JB, Romanzetti S, Dierkes T, Leach MO, Balcom BJ, Jon Shah N. Multi-Frame SPRITE: A method for resolution enhancement of multiple-point SPRITE data. *J Magn Reson.* 2013; 230C: 111–116. DOI: 10.1016/j.jmr.2013.01.008 [PubMed: 23454580]
17. Xiao D, Balcom BJ. Hybrid-SPRITE MRI. *J Magn Reson.* 2013; 235:6–14. DOI: 10.1016/j.jmr.2013.07.003 [PubMed: 23916990]
18. Hafner S. Fast imaging in liquids and solids with the Back-projection Low Angle ShoT (BLAST) technique. *Magn Reson Imaging.* 1994; 12:1047–1051. DOI: 10.1016/0730-725X(94)91236-P [PubMed: 7997092]

19. Madio DP, Lowe IJ. Ultra-fast imaging using low flip angles and fids. *Magn Reson Med*. 1995; 34:525–529. DOI: 10.1002/mrm.1910340407 [PubMed: 8524019]
20. Wu Y, Ackerman JL, Chesler Da, Graham L, Wang Y, Glimcher MJ. Density of organic matrix of native mineralized bone measured by water- and fat-suppressed proton projection MRI. *Magn Reson Med*. 2003; 50:59–68. DOI: 10.1002/mrm.10512 [PubMed: 12815679]
21. Schieban K, Weiger M, Hennel F, Boss A, Pruessmann KP. ZTE imaging with enhanced flip angle using modulated excitation. *Magn Reson Med*. 2014; 00:1–10. DOI: 10.1002/mrm.25464
22. Grodzki DM, Jakob PM, Heismann B. Ultrashort echo time imaging using pointwise encoding time reduction with radial acquisition (PETRA). *Magn Reson Med*. 2012; 67:510–8. DOI: 10.1002/mrm.23017 [PubMed: 21721039]
23. Idiyatullin D, Corum C, Park JY, Garwood M. Fast and quiet MRI using a swept radiofrequency. *J Magn Reson*. 2006; 181:342–9. DOI: 10.1016/j.jmr.2006.05.014 [PubMed: 16782371]
24. Idiyatullin D, Corum Ca, Garwood M. Multi-Band-SWIFT. *J Magn Reson*. 2014; doi: 10.1016/j.jmr.2014.11.014
25. Zhang J, Idiyatullin D, Corum Ca, Kobayashi N, Garwood M. Gradient-modulated SWIFT. *Magn Reson Med*. 2015; 00:n/a–n/a. doi: 10.1002/mrm.25595
26. Grodzki DM, Jakob PM, Heismann B. Correcting slice selectivity in hard pulse sequences. *J Magn Reson*. 2012; 214:61–7. DOI: 10.1016/j.jmr.2011.10.005 [PubMed: 22047992]
27. Li C, Magland JF, Seifert AC, Wehrli FW. Correction of excitation profile in zero echo time (ZTE) imaging using quadratic phase-modulated RF pulse excitation and iterative reconstruction. *IEEE Trans Med Imaging*. 2014; 33:961–969. DOI: 10.1109/TMI.2014.2300500 [PubMed: 24710164]
28. Rahmer J, Börner P, Groen J, Bos C. Three-dimensional radial ultrashort echo-time imaging with T2 adapted sampling. *Magn Reson Med*. 2006; 55:1075–1082. DOI: 10.1002/mrm.20868 [PubMed: 16538604]
29. Subramanian S, Devasahayam N, Murugesan R, Yamada K, Cook J, Taube A, Mitchell JB, Lohman JaB, Krishna MC. Single-point (constant-time) imaging in radiofrequency Fourier transform electron paramagnetic resonance. *Magn Reson Med*. 2002; 48:370–9. DOI: 10.1002/mrm.10199 [PubMed: 12210946]
30. Balcom BJ, Bogdan M, Armstrong RL. Single-Point Imaging of Gradient Rise, Stabilization, and Decay. *J Magn Reson Ser A*. 1996; 118:122–125. DOI: 10.1006/jmra.1996.0018
31. Beatty PJ, Nishimura DG, Pauly JM. Rapid gridding reconstruction with a minimal oversampling ratio. *IEEE Trans Med Imaging*. 2005; 24:799–808. DOI: 10.1109/TMI.2005.848376 [PubMed: 15959939]
32. Pipe JG, Menon P. Sampling density compensation in MRI: rationale and an iterative numerical solution. *Magn Reson Med*. 1999; 41:179–86. [PubMed: 10025627]
33. Pipe JG. Reconstructing MR images from undersampled data: data-weighting considerations. *Magn Reson Med*. 2000; 43:867–75. [PubMed: 10861882]
34. Johnson KO, Pipe JG. Convolution kernel design and efficient algorithm for sampling density correction. *Magn Reson Med*. 2009; 61:439–47. DOI: 10.1002/mrm.21840 [PubMed: 19165893]
35. Reeder SB, Pineda AR, Wen Z, Shimakawa A, Yu H, Brittain JH, Gold GE, Beaulieu CH, Pelc NJ. Iterative decomposition of water and fat with echo asymmetry and least-squares estimation (IDEAL): application with fast spin-echo imaging. *Magn Reson Med*. 2005; 54:636–44. DOI: 10.1002/mrm.20624 [PubMed: 16092103]
36. Jang H, Subramanian S, Devasahayam N, Saito K, Matsumoto S, Krishna MC, McMillan AB. Single acquisition quantitative single-point electron paramagnetic resonance imaging. *Magn Reson Med*. 2013; 70:1173–1181. DOI: 10.1002/mrm.24886 [PubMed: 23913515]
37. Lustig M, Donoho D, Pauly JM. Sparse MRI: The application of compressed sensing for rapid MR imaging. *Magn Reson Med*. 2007; 58:1182–95. DOI: 10.1002/mrm.21391 [PubMed: 17969013]
38. Lustig M, Donoho D. Compressed Sensing MRI. *Signal Process Mag*. 2008:72–82.
39. Jang H, Matsumoto S, Devasahayam N, Subramanian S, Zhuo J, Krishna MC, McMillan AB. Accelerated 4D quantitative single point EPR imaging using model-based reconstruction. *Magn Reson Med*. 2014; 00:1–10. DOI: 10.1002/mrm.25282

40. Huang C, Graff CG, Clarkson EW, Bilgin A, Altbach MI. T2 mapping from highly undersampled data by reconstruction of principal component coefficient maps using compressed sensing. *Magn Reson Med*. 2012; 67:1355–66. DOI: 10.1002/mrm.23128 [PubMed: 22190358]
41. Hennel F. Fast spin echo and fast gradient echo MRI with low acoustic noise. *J Magn Reson Imaging*. 2001; 13:960–6. [PubMed: 11382960]
42. Cho ZH, Chung ST, Chung JY, Park SH, Kim JS, Moon CH, Hong IK. A new silent magnetic resonance imaging using a rotating DC gradient. *Magn Reson Med*. 1998; 39:317–321. DOI: 10.1002/mrm.1910390221 [PubMed: 9469717]
43. Hennel F, Girard F, Loenneker T. “Silent” MRI with soft gradient pulses. *Magn Reson Med*. 1999; 42:6–10. DOI: 10.1002/(SICI)1522-2594(199907)42:1<6::AID-MRM2>3.0.CO;2-D [PubMed: 10398943]
44. Pineda AR, Reeder SB, Wen Z, Pelc NJ. Cramer-Rao bounds for three-point decomposition of water and fat. *Magn Reson Med*. 2005; 54:625–635. DOI: 10.1002/mrm.20623 [PubMed: 16092102]
45. Chebrolu VV, Yu H, Pineda AR, McKenzie Ca, Brittain JH, Reeder SB. Noise analysis for 3-point chemical shift-based water-fat separation with spectral modeling of fat. *J Magn Reson Imaging*. 2010; 32:493–500. DOI: 10.1002/jmri.22220 [PubMed: 20677283]
46. Heba, ER.; Hamilton, G.; Sirlin, CB.; Wolfson, T.; Gamst, A.; Loomba, R.; Middleton, MS. Intl Soc Mag Reson Med. Vol. 22. Milan, Italy: 2014. Agreement of 2-, 3-, 4-, 5- and 6-echo MRI-PDFF with MRS-PDFF in 580 adults with known or suspected non-alcoholic fatty liver disease (NAFLD); p. 0138
47. Duyn JH, Yang Y, Frank Ja, van der Veen JW. Simple correction method for k-space trajectory deviations in MRI. *J Magn Reson*. 1998; 132:150–153. DOI: 10.1006/jmre.1998.1396 [PubMed: 9615415]
48. Barmet C, De Zanche N, Pruessmann KP. Spatiotemporal magnetic field monitoring for MR. *Magn Reson Med*. 2008; 60:187–197. DOI: 10.1002/mrm.21603 [PubMed: 18581361]

Figure 1.

Slice selectivity (a) PSD as used in PETRA/ZTE, and (b) Excitation profile of a 24 μ s hard pulse, Note that the degree of slice selectivity increases with encoding gradient amplitude in (b).

Figure 2.

Ramped Hybrid Encoding (RHE). (a) Pulse sequence diagram, (b) sampling scheme, and (c) example of a multi-echo encoding scheme. RHE allows flexible control of G_{RF} to minimize slice selectivity artifacts, and allows the best possible encoding time by rapidly ramping the gradient amplitude after RF excitation. Like PETRA, single point encoding is employed to acquire the central k-space data.

Figure 3.

The zoom-in effect of SPI and gradient calibration. (a) 1D SPI image matrix, and (b) the prescribed gradient shape. Note that the FOV change in (a) directly represents gradient shape in (b), which can be utilized to estimate k-space trajectory.

Figure 4.

Raw data processing in RHE. Raw data acquired by RHE contains Cartesian SPI data, radially frequency encoded data, and 3 sets of 1D SPI data for calibration. After calibration, combined data with estimated k-space position are reconstructed by convolution gridding.

Figure 5.

Simulation of (a) per-excitation encoding time and simulated 1D imaging with (b) $T_2^*=100\mu\text{s}$ and (c) $T_2^*=500\mu\text{s}$. N_{SPI} was set to 24, 69, 29, and 40 respectively for PETRA with $G_{\text{max}}=7\text{mTm}^{-1}$, PETRA with $G_{\text{max}}=20\text{mTm}^{-1}$, RHE with $G_{\text{RF}}=3.5\text{mTm}^{-1}$, and RHE with $G_{\text{RF}}=7\text{mTm}^{-1}$. Note that RHE provides the shortest per-excitation encoding time and the best image reconstruction for short T_2^* imaging over a wider field-of-view than PETRA and FE-UTE.

Figure 6.

Gradient calibration. (a) Five 1D projection images from calibration data in the x-direction exhibiting a zoom-in effect from SPI, (b) the corresponding scaling factors computed from (a), (c) the measured k-space trajectory in each gradient axis, (d) comparison with the prescribed or delay-corrected trajectory, and (e) the resultant images. Note that reconstruction with the prescribed trajectory (middle) results in substantial error in the image (ringing) and an incorrect FOV. The delay-corrected trajectory (right), while having the correct FOV, has blurring and ringing compared to the image reconstructed with the measured trajectory (left).

Figure 7. Phantom experiment. PETRA with (a) $G_{\text{max}}=7\text{mTm}^{-1}$, (b) $G_{\text{max}}=14\text{mTm}^{-1}$, (c) $G_{\text{max}}=20\text{mTm}^{-1}$, (d) FE-UTE with $G_{\text{max}}=35\text{mTm}^{-1}$, and (e) RHE with $G_{\text{RF}}=7\text{mTm}^{-1}$ and $G_{\text{max}}=35\text{mTm}^{-1}$. RHE allows the shortest per-excitation encoding time, yielding the best image quality.

Figure 8.

In vivo knee experiment. Coronal slice of RHE images at TE of (a) 90 μ s, (b) 1502 μ s, (c) 1550 μ s, (d) 2900 μ s, (e) 2950 μ s, (f) water image, (g) fat image, (h) short T_2^* image, sagittal slice of RHE images at TE of (i) 90 μ s, (j) 1502 μ s, (k) 1550 μ s, (l) 2900 μ s, (m) 2950 μ s, (n) water image, (o) fat image, and (p) short T_2^* image. To separate water and fat image IDEAL was applied using five images at TE= 90 μ s, 1502 μ s, 1550 μ s, 2900 μ s, and 2950 μ s after image reconstruction. The short T_2^* image was obtained by subtracting computed water and fat images from the image at TE=90 μ s. Short T_2^* tissues are clearly visible (white arrow: medial collateral ligament and lateral collateral ligament, yellow arrow: medial meniscus, blue arrow: quadriceps femoris tendon, green arrow: patellar ligament, red arrow: anterior cruciate).

Figure 9.

In vivo brain experiment. Mid-sagittal plane image obtained by PETRA with 8μs RF pulse with (a) $G_{\max}=7\text{mTm}^{-1}$, (b) $G_{\max}=14\text{mTm}^{-1}$, (c) $G_{\max}=20\text{mTm}^{-1}$, (d) FE-UTE with 8μs RF pulse and $G_{\max}=35\text{mTm}^{-1}$, (e) RHE with 8μs RF pulse, $G_{\text{RF}}=7\text{mTm}^{-1}$, and $G_{\max}=35\text{mTm}^{-1}$, PETRA with 24μs RF pulse with (f) $G_{\max}=7\text{mTm}^{-1}$, (g) $G_{\max}=14\text{mTm}^{-1}$, (h) $G_{\max}=20\text{mTm}^{-1}$, (i) FE-UTE with 24μs RF pulse and $G_{\max}=35\text{mTm}^{-1}$, (j) RHE with 24μs RF pulse, $G_{\text{RF}}=7\text{mTm}^{-1}$, and $G_{\max}=35\text{mTm}^{-1}$, axial plane image obtained by PETRA with 8μs RF pulse with (k) $G_{\max}=7\text{mTm}^{-1}$, (l) $G_{\max}=14\text{mTm}^{-1}$, (m) $G_{\max}=20\text{mTm}^{-1}$, (n) FE-UTE with 8μs RF pulse and $G_{\max}=35\text{mTm}^{-1}$, (o) RHE with 8μs RF pulse, $G_{\text{RF}}=7\text{mTm}^{-1}$, and $G_{\max}=35\text{mTm}^{-1}$, PETRA with 24μs RF pulse with (p) $G_{\max}=7\text{mTm}^{-1}$, (q) $G_{\max}=14\text{mTm}^{-1}$, (r) $G_{\max}=20\text{mTm}^{-1}$, (s) FE-UTE with 24μs RF pulse and $G_{\max}=35\text{mTm}^{-1}$, (t) RHE with 24μs RF pulse, $G_{\text{RF}}=7\text{mTm}^{-1}$, and $G_{\max}=35\text{mTm}^{-1}$, and its corresponding zoomed-in images on the right. Note that the slice selectivity of PETRA increases and the chemical shift artifact decreases as the strength of the readout gradient increases. By using a short RF pulse (8μs), the slice selectivity artifact can be alleviated, but SNR and T1 contrast are inevitably reduced due to the smaller attainable flip angle. Both FE-UTE and RHE shows better image quality with no apparent chemical shift artifact and slice selectivity artifact, but RHE shows more signal intensity from compact bone structures than FE-UTE.

Table 1

Parameters for MR experiments.

	Phantom					Knee		Brain				
	PETRA	PETRA	PETRA	UTE	RHE	RHE		PETRA	PETRA	PETRA	UTE	RHE
G_{max} (mTm ⁻¹)	7	14	20	35	35	35		7	14	20	35	35
G_{RF} (mTm ⁻¹)	n/a	n/a	n/a	n/a	7	5		n/a	n/a	n/a	n/a	7
RF pulse width (μs)	24					24		8, 24				
Flip angle (°)	6					6		2, 6				
NSPI	33	33	33	n/a	33	33		33	33	33	n/a	33
# of SPI encoding	17707	17707	17707	n/a	17707	17707		17707	17707	17707	n/a	17707
# of FE encoding	80000	80000	80000	80000	80000	80000		80000	80000	80000	80000	80000
Slew rate (mTm ⁻¹ ms ⁻¹)	118	118	118	118	118	118		118	118	118	118	118
TE (μs)	90	90	90	90	90	90, 1502, 1550, 2900, 2950		90	90	90	90	90
T_{Enc} (μs)	1680	838	588	588	438	788		1680	838	588	588	438
TR (ms)	3.3	3.3	3.3	3.3	3.3	5.6		3.3	3.3	3.3	3.3	3.3
RF coil	8ch receive only head coil					8ch T/R knee coil		8ch receive only head coil				
Scan time	5m 23s	5m 23s	5m 23s	4m 28s	5m 23s	9m 10s		5m 23s	5m 23s	5m 23s	4m 28s	5m 23s