Gradient-Modulated PETRA MRI

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INTRODUCTION

Image blurring that results from off-resonance and fast signal decay of extremely short \( T_2^* \) spins is a common issue in radial ultrashort echo time magnetic resonance imaging (MRI) sequences such as zero echo time (ZTE) \((1, 2)\) and sweep imaging with Fourier transformation \((3)\) techniques. Although higher static magnetic field strengths have a signal-to-noise ratio advantage, such image blurring is even more severe because of increasing chemical shifts and accelerated \( T_2^* \) signal decay. One efficient and simple solution for this image-blurring issue is to use a higher readout bandwidth.

In the ZTE technique, gradients are turned on before applying the radiofrequency (RF) excitation pulse (a constant-amplitude “hard” pulse), and the gradient amplitude remains constant during the ensuing data acquisition. Therefore, excitation and readout bandwidths need to be the same; a higher readout bandwidth results in a higher excitation bandwidth. However, a high excitation bandwidth (i.e., short pulse length) leads to an increase in the specific absorption rate (SAR) and RF pulse peak power, which are the major limitations of ZTE in human applications \((4)\). Another severe limitation of using higher bandwidths in ZTE is an increasing demand for fast switching between transmit and receive (T/R) modes. The slow T/R switching of most standard clinical MRI scanners (>20 \( \mu \)s) results in missing critical data points in a large region around the center of \( k \)-space \((5)\).

Pointwise-encoding time reduction with radial acquisition (PETRA) has been recently introduced \((6, 7)\) and overcomes the T/R switching limitation of clinical scanners by combining ZTE and single-point imaging (SPI) \((8)\). PETRA acquires the missing \( k \)-space center region in ZTE with additional SPI acquisitions. However, although PETRA alleviates the requirement on fast T/R switching, the use of a higher bandwidth increases the number of missing \( k \)-space center points. Accordingly, higher bandwidths along with slow T/R switching in clinical scanners results in a large missing region around the \( k \)-space center, namely a long additional SPI acquisition time.

We introduce herein a novel PETRA technique with gradient modulation (GM) that enables high readout bandwidths while keeping a relatively low excitation bandwidth. GM significantly reduces SAR and RF peak power and keeps the missing center \( k \)-space region small. Tolerance to image blurring as a result of off-resonance and fast \( T_2^* \) signal decay was evaluated in simulation and experimentally by comparing PETRA with and without GM. Finally, we conducted inner ear imaging of a normal subject at 7 T.
**Figure 1.** (A) Sequence diagram of PETRA with and without GM and magnification of the region around excitation shown by a red box (B). GM-PETRA is identical to PETRA until gradients start ramping up. (C) k-space sampling in PETRA and GM-PETRA is a hybrid of 2 sampling strategies. The center region is sampled with SPI (red points) and the peripheral region by radial readout (green lines). The boundary of the 2 sampling regions ($k_{SP}$) is limited by $t_k = pw/2 + t_0$ ($k_{SP} = \gamma G_{ex} \cdot t_k$; blue circle). (D) GM-PETRA samples k-space more quickly by increasing the gradient amplitude after excitation. Although PETRA sampling with a high bandwidth (120 kHz) achieves faster sampling around the k-space center, GM-PETRA with lower excitation bandwidth ($G_{ex} = 60$ kHz) catches up on it quickly as the gradient modulation increases ($G_{max} = 125$ and 200 kHz).

**Equation 1:**

$$B_{1max} \propto \alpha \cdot BW_{ex}$$

**Equation 2:**

$$SAR \propto \alpha^2 \cdot BW_{ex}$$

Therefore, a lower $BW_{ex}$ reduces SAR and RF peak power. Another advantage of using a lower excitation bandwidth is the reduction of missing points around the k-space center because the number of missing points in each ZTE acquisition is given by $t_k \cdot BW_{ex}$.

**Simulation**

To evaluate the ability of GM-PETRA to address image blurring as a result of off-resonance and short $T_2^*$ signals, numerical simulations were conducted for a 3D Shepp–Logan phantom. The phantom contained 4 compartments: off-resonance ($\Delta \omega = 1050$ Hz; 3.5 ppm at 7 T), on-resonance, and 2 short $T_2^*$ values (500 and 200 $\mu$s). $T_2^*$ values for on- and off-resonance spins were set to 40 ms, and the frequency of the short $T_2^*$ spins was on-resonance ($\Delta \omega = 0$ Hz). The k-space signal $S(k)$ was calculated by

$$S(k) = s(k) \cdot e^{j \pi \Delta \omega t} \cdot e^{-\gamma T_2^*}$$

**Equation 3:**

$$k(t) = \int_0^t \gamma G(t')dt'$$

**Equation 4:**

where $s(k)$ is the k-space without off-resonance and $T_2^*$ signal decay, $G(t)$ is a gradient vector in 3D space, $t$ is the sampling time after the center of the hard pulse excitation (k-space center), and $k(t)$ is the 3D k-space coordinates. A plot of
Simulations were performed with excitation bandwidths of 60 and 120 kHz in PETRA and a 60-kHz excitation followed by bandwidth increases to 125 and 200 kHz in GM-PETRA.

**Experiments**

All experiments were performed with a 7 T MRI scanner (Siemens, Erlangen, Germany) using a head coil with 8 transceiver and 24 receiver channels (Nova Medical, Wilmington, MA). For $B_1^+$ shimming, complex $B_1^+$ maps were estimated for each of the 8 transmit channels using a fast, low-FA, multichannel $B_1^+$ calibration scan (9). A $B_1^+$ shim solution was optimized to maximize the efficiency over a selected region of interest (10).

To test image blurring caused by off-resonance and/or short $T_2^*$ signals, a breast phantom and apple were imaged with PETRA and GM-PETRA. The breast phantom had fat and water compartments inside ($\Delta \omega = 1050$ and 0 Hz, respectively). The excitation bandwidth ($\gamma G_{\text{r}}, \text{field of view [FOV]}$) was set to 60 kHz, which provided $\text{pw} = 20 \mu s$ and $\text{TE} = 50 \mu s$. The following parameters were used in PETRA and GM-PETRA: $\text{FA} = 5^\circ$; $\text{retention time [TR]} = 5 \text{ ms}$; total acquisitions $= 65 536$, including 123 SPI acquisitions; and scan time $= 5 \text{ min 29 s}$. For PETRA, a higher excitation bandwidth of 120 kHz was also tested for comparison, where TE $= 45 \mu s$ and 925 SPI acquisitions were needed. In GM-PETRA, the gradient modulation was set to $\gamma G_{\text{max}}, \text{FOV} = 125$ and 200 kHz as in the simulations. The slew rate of the ramping gradient was fixed to 100 mT/m/ms (50% of the maximum slew rate) to avoid deleterious effects from eddy currents, resulting in a ramping time of 60-80 and 120-170 $\mu s$ for 125 and 200 kHz, respectively. For imaging the apple, a conventional 3D gradient echo (GRE) acquisition was performed to demonstrate its short $T_2^*$ signal decay, which resulted from the large volume of intercellular air spaces (11); the TE was set to its minimum of 2.8 ms. The other parameters were $\text{TR} = 5.8 \text{ ms}$ and scan time $= 5 \text{ min 23 s}$.

Inner ear imaging of a healthy subject was conducted under an approved internal review board protocol. Sequence parameters were set to the same as those in the breast phantom and apple imaging, but FA was $4^\circ$ because of a severe SAR limitation at 7 T. PETRA acquisition was carried out only with a 60-kHz excitation because 120 kHz exceeded the SAR limit. 3D GRE imaging was also performed for comparison.

Image reconstruction was performed offline with a home-built program written in C++/CUDA (NVIDIA, Santa Clara, CA). $K$-space sampling density, including the effects from the ramp sampling in GM-PETRA, was compensated by the iterative density correction (12). The density-corrected radial $k$-space data were reconstructed to a 3D image with non-uniform fast Fourier transform (13).

**RESULTS**

**Simulation**

Simulations showed that off-resonance blurring (“duplicated edges” around the off-resonance compartment) improved as the bandwidth increased (Figure 2); blurry edges got thinner for higher bandwidths. By increasing the GM-PETRA amplitude from 60 to 125 kHz (GM-PETRA 60–125 kHz), off-resonance artifacts became comparable to PETRA with a 120-kHz excitation (PETRA

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120 kHz). GM-PETRA 60-200 kHz showed further improvement as a result of reduced off-resonance blurring. For short $T_2^*$ spins, increasing the bandwidth improved edge sharpness (Figure 1C). PETRA 120 kHz showed an increased ability to preserve short $T_2^*$ signals (i.e., higher signal intensities), especially for spins with $T_2^* = 200 \mu s$, because of its shorter TE and faster sampling around the $k$-space center (Figure 1D). To preserve extremely short $T_2^*$ signals, higher excitation bandwidths are desirable.

Experiments
Breast phantom images with PETRA and GM-PETRA were consistent with simulation results (Figure 3A); the off-resonance blurring that resulted from the chemical shift of fat and magnetic susceptibility differences were improved as the bandwidth increased. In apple imaging, GM-PETRA and PETRA showed higher sensitivity to signals from the mesocarp (flesh) that was almost invisible with 3D GRE because of its extremely fast $T_2^*$

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**Figure 3.** (A) Improvement of the off-resonance artifacts as a result of chemical shift (fat-water) and susceptibility differences in a phantom experiment with GM-PETRA. Duplicated edges around fat regions became thinner as the GM increased; small gaps between fat compartments were visualized better in images from higher bandwidth (yellow arrowheads). Image blurring and distortion around the edges of water compartments also improved along with an increase of the bandwidth (light blue arrowheads). PETRA 120 kHz showed similar image blurring artifacts to GM-PETRA 60–125 kHz (data not shown), which is consistent with the simulation results shown in Figure 2. (B) Image blurring as a result of short $T_2^*$ decay and/or off-resonance demonstrated in an apple imaging. GM-PETRA allowed finer structures in the mesocarp (flesh) to be appreciated compared to PETRA; these structures were completely invisible in 3D GRE because of their very short $T_2^*$ values. Although the calyx had a relatively long $T_2^*$ (which makes it visible with 3D GRE), it was completely blurred out with PETRA because of strong susceptibility effects (red arrowheads). (C) An MIP image from GM-PETRA clearly allows fine structures in the mesocarp to be appreciated, including fibers compared to the other 2 acquisition methods.
signal decay (Figure 3B). The PETRA image was visually blurry compared to GM-PETRA because of off-resonance associated with susceptibility differences and fast T₂* signal decay. In maximum intensity projection (MIP) images, GM-PETRA visualized fine structures in the apple better than the other 2 methods (Figure 3C). 3D GRE also visualized structures such as vascular bundles, endocarp (core), and exocarp (skin), but many of fine structures in the mesocarp were missing.

Visualization of small structures in the inner ear was improved by GM-PETRA (Figure 4A). Although PETRA and GM-PETRA had comparable sensitivity to short T₂* signals, PETRA suffered from severe image blurring mainly because of strong susceptibility differences at the air-tissue interfaces. The strong susceptibility differences accelerated the T₂* signal decay, making it difficult to visualize some structures of the inner ear (eg, small nerves) with 3D GRE (yellow arrowheads). MIP images calculated for GM-PETRA by masking out the surrounding tissues around the inner ear. The MIP images allowed complex structures in the inner ear to be visualized. MIP images projected to different image planes are available in the online Supplemental Materials (Figure 4B).

**DISCUSSION**
Grade modulation has been introduced in the PETRA technique to increase readout bandwidth, which, in turn, can reduce image-blurring artifacts resulting from off-resonance and T₂* signal decay. This work demonstrates how gradient modulation enables
flexible settings of excitation and readout bandwidths. The possibility of using a decreased excitation bandwidth while keeping the readout bandwidth high can alleviate the severe limitations in clinical applications imposed by SAR and RF peak power, which are proportional to excitation bandwidth (Equations 1 and 2).

Gradient modulation was previously implemented to increase achievable flip angles and to reduce off-resonance blurring in radial ultrashort TE techniques (14–16). One benefit of the higher flip-angle availability in GM-PETRA is to achieve stronger T1-weighted image contrasts, whereas many ultrashort TE techniques provide mostly proton density-weighted contrasts because of their relatively low flip angles. In ZTE, as the bandwidth increases, achievable flip angles were severely limited as a result of short pulse widths. When a longer pulse is used, the sinc-shaped excitation profile of the hard pulse causes blurring artifacts around the edges of the FOV. This issue was addressed in previous studies by correcting the sinc-shaped profile in image reconstruction (17) or by using frequency- (or phase-) modulated pulses to flatten the excitation profile (18, 19). By combining gradient modulation with frequency-modulated pulse excitation, the limitation can be further alleviated. However, because frequency-modulated pulse excitation generates a nonlinear phase distribution in space (20), one needs to handle the nonlinear phase by using methods such as algebraic (18, 21) and iterative (19) reconstruction methods.

One disadvantage of using gradient modulation is an increase of acoustic noise, as compared with the original PETRA sequence, which is extremely quiet (6, 7). Therefore, there is a tradeoff between GM-PETRA’s increased acoustic noise and its ability to decrease blurring artifacts and mitigated SAR and/or RF peak power limitations. Moreover, as the bandwidth increases, the sequence becomes more sensitive to gradient performance, including group delays and eddy currents. Gradient miscalibration may thereby induce image quality degradation.

Another tradeoff of using higher bandwidths in GM-PETRA while keeping other sequence parameters fixed (as in this study) is between reduced image blurring and SNR. Although a conspicuous increase of noise was not observed with the current settings of up to 200 kHz at 7 T, randomly distributed noise can be efficiently removed by applying image reconstructions with regularization such as with compressed sensing (22–24). Such image reconstruction methods can improve image quality as well as accelerate data acquisition by extracting information from coil sensitivity variations between multiple receiver coils and sparsity in an appropriate transform domain (eg, wavelet and total variation).

In conclusion, GM-PETRA showed significantly less image blurring as a result of off-resonance and fast T2* signal decay compared to the conventional PETRA technique. Gradient modulation can improve PETRA image quality and mitigate the limitation on SAR and RF peak power without requiring specific hardware modifications on clinical scanners.

Supplemental Materials:
Video: http://tiny.cc/tom-01-02-s003

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