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Fully Phase-Encoded MRI Near Metallic Implants Using Ultrashort Echo Times and Broadband Excitation

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Abstract

Purpose—To develop a fully phase-encoded MRI method for distortion-free imaging near metallic implants, in clinically feasible acquisition times.

Theory and Methods—An accelerated 3D fully phase-encoded acquisition with broadband excitation and ultrashort echo times is presented, which uses a broadband radiofrequency pulse to excite the entire off-resonance induced by the metallic implant. Furthermore, fully phase-encoded imaging is used to prevent distortions caused by frequency encoding, and to obtain ultrashort echo times for rapidly decaying signal.

Results—Phantom and in vivo acquisitions were used to describe the relationship among excitation bandwidth, signal loss near metallic implants, and T_1 weighting. Shorter radiofrequency pulses captured signal closer to the implant by improving spectral coverage and allowing shorter echo times, whereas longer pulses improved T_1 weighting through larger maximum attainable flip angles. Comparisons of fully phase-encoded acquisition with broadband excitation and ultrashort echo times to T_1 -weighted multi-acquisition with variable resonance image combination selective were performed in phantoms and subjects with metallic knee and hip prostheses. These acquisitions had similar contrast and acquisition efficiency.

Conclusions—Accelerated fully phase-encoded acquisitions with ultrashort echo times and broadband excitation can generate distortion free images near metallic implants in clinically feasible acquisition times. *Magn Reson Med* 000:000–000, 2017.

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Keywords

metallic implant; ultrashort TE; fully phase encoded; single point imaging

INTRODUCTION

The prevalence of joint replacement surgeries to treat degenerative joint diseases have dramatically increased in the past few decades (1,2). Although improvements to implant design and surgical techniques have reduced residual or recurrent symptoms, prosthetic-related complications are common (3–6). This increase in joint replacement surgeries has resulted in a clinical need for noninvasive evaluation of prosthetic-related complications.

Metallic implants pose several challenges for MR imaging. Large magnetic susceptibility differences between metallic implants and tissue generate extremely large perturbations in the main magnetic field (B_0). These spatially varying magnetic fields and rapid changes in the magnetic field can lead to signal voids and frequency-encoding distortions near metallic implants (7–10).

Artifacts arising from metallic implants can be reduced substantially using Carr-Purcell-Meiboom-Gill spin-echo-based 3D multispectral imaging (3D-MSI) techniques such as multi-acquisition with variable resonance image combination (MAVRIC (11) and MAVRIC selective (SL) (12)) and slice encoding for metal artifact correction (13). These techniques use multiple acquisitions spanning different transmit and receive frequencies to cover a wide spectrum of off-resonance frequencies, along with phase-encoding along the slice/ slab-selective dimension. Rapidly dephasing signal near the implant is refocused using spin echoes. Frequency-encoding distortions that manifest as geometric distortions, signal pile-up, and signal void artifacts are limited using a combination of a high readout bandwidth, a limited radiofrequency (RF) excitation bandwidth (e.g., 2.25 kHz), and view angle tilting concepts (12,13).

Fully phase-encoded spin-echo techniques have been proposed to avoid frequency-encoding-related artifacts (14–16). Multiple 3D phase-encoded spin-echo acquisitions at different transmit and receive frequencies are required to excite the broad spectrum near metal implants. Unfortunately, the cost of multiple fully phase-encoded (phase encoding in all three dimensions) is very long acquisition times (hours). Although substantial progress has been made to accelerate these acquisitions, long acquisition times remain as a barrier to the clinical feasibility of these techniques (17–19).

Frequency and fully phase-encoded ultrashort echo time (UTE) gradient-echo techniques have also been proposed for imaging near metallic implants (20,21). These techniques attempt to capture the rapidly dephasing signal before it decays away. Similar to the MAVRIC 3D-MSI concept, a radial frequency-encoded UTE method was proposed by acquiring multiple UTE images at different carrier frequencies spanning a large range of off-resonance (20). Fully phase-encoded UTE acquisitions using the single point ramped imaging with T_1 enhancement (SPRITE) method (22) have also been proposed (21). These

acquisitions have only been acquired in phantom studies as a result of long acquisition times and gradient performance constraints not feasible on clinical scanners.

Therefore, the purpose of this work was to develop a fully phase-encoded UTE method toward distortion-free imaging near metallic implants in clinically feasible acquisition times. To enable this, an undersampled fully phase-encoded 3D UTE acquisition (UTE-FPE) with broadband excitation is proposed. The relationship among excitation bandwidth, signal loss near metallic implants, and T_1 weighting are described using phantoms and a subject with a total knee replacement. Comparisons to MAVRIC SL are performed in phantoms and subjects with total knee and hip replacements.

THEORY

Broadband UTE-FPE

To overcome the challenges of exciting the entire off-resonance spectrum, rapid T_2^* decay, and frequency encoding distortions, we propose a fully phase-encoded UTE acquisition with broadband excitation for imaging near metal (Fig. 1). In this approach, broadband RF excitations using low flip angles and nonselective RF pulses, capable of exciting the entire off-resonance spectrum, are used to acquire images near metal with a single acquisition. Furthermore, the use of phase encoding in all directions prevents frequency encoding–related distortions including large bulk distortions caused by the combination of frequency encoding and broadband RF excitations. Ultrashort echo times are used to capture signal before it experiences rapid T_2^* decay. Clinically feasible acquisition times can be obtained using parallel imaging accelerations that can be performed in all three encoding dimensions. As shown in the proposed UTE-FPE pulse sequence (Fig. 1), excitation occurs with the gradients at an amplitude not exceeding a maximum gradient strength (maximum G_{RF}). The application of gradients during RF excitation introduces an additional source of off-resonance. For nonselective RF pulses to remain nonselective, the bandwidth of the RF pulse must be larger than the off-resonance caused both the metallic implant and the gradients. Therefore, the maximum gradient strength during RF should be large enough to allow fast phase encoding after excitation (i.e., to minimize the echo time (TE)) but small enough to prevent unwanted slice selectivity (23). Immediately after RF excitation, the gradients are ramped to a value equal to or less than the maximum encoding gradient (G_{max}). Multiple temporal samples are acquired throughout the acquisition, whereas the phase encoding gradient remains on, providing images with various field of views (including the prescribed field of view), an effect referred to as the “zoom-in” effect in fully phase-encoded imaging (24,25).

METHODS

Phantoms and in vivo experiments were performed to evaluate the proposed method. For both phantom and in vivo experiments, reference acquisitions were performed. Phantom comparisons to MAVRIC-SL and frequency-encoded UTE (UTE-FE) acquisitions were performed to compare UTE-FPE in terms of signal loss near the implant (MAVRIC) and distortion reduction (UTE-FE). In vivo comparisons to MAVRIC-SL and 2D fast spin-echo (FSE) acquisitions were performed to compare UTE-FPE with clinically used acquisitions in

terms of signal loss near the implant, image contrast, and acquisition times. In vivo experiments were performed after obtaining informed consent and approval from the University of Wisconsin Health Sciences Institutional Review Board.

Total Hip Prosthesis Phantom

Two matching total hip prosthesis phantoms were constructed using a hip prosthesis consisting of a cobalt-chromium-molybdenum alloy head and titanium stem (Alliance X-Series, Integral Porous Primary Hip System, Biomet Orthopedics Inc, Warsaw, IN) and a 3D printed plastic replica of the hip prosthesis.

The UTE-FPE, MAVRIC-SL, and UTE-FE acquisitions of the metallic total hip replacement phantom were performed at both 1.5 T (MR450w, GE Healthcare, Wauke-sha, WI) and 3 T (MR750, GE Healthcare) using a 16-channel wrap coil (NeoCoil, Pewaukee, WI). The UTE-FPE acquisitions using different pulse widths and different maximum G_{RF} values were performed to characterize the effect of changing the maximum G_{RF} on TE, signal loss near the implant, and selectivity artifacts for different RF pulse widths.

In each UTE-FPE acquisition, the pulse width of hard RF pulses (8, 24, and 144) and the maximum gradient amplitudes during RF excitation (maximum $G_{RF} = 0, 10, \text{ and } 20 \text{ mTm}^{-1}$) were varied. For all UTE-FPE acquisitions (1.5/3 T), the following imaging parameters were held constant: parallel imaging acceleration (R) = $2 \times 2 \times 1$, autocalibration signal = $31 \times 31 \times 31$, matrix = $240 \times 132 \times 40$, field of view (FOV) = $240 \times 132 \times 80 \text{ mm}$, repetition time (TR) = 2.8/1.6 ms, flip angle (FA) = 3° , bandwidth = $\pm 250 \text{ kHz}$, $G_{\text{max}} = 31.5/47.5 \text{ mTm}^{-1}$, acquisition time = 10:07/5:47 min. The TEs for the 1.5T acquisitions with G_{RF} of 0, 10, and 20, respectively, were $TE_{8 \mu s} = 512, 432, \text{ and } 386 \mu s$; $TE_{24 \mu s} = 522, 438, \text{ and } 388 \mu s$; and $TE_{144 \mu s} = 622, 528, \text{ and } 452 \mu s$. The TEs for the 3T acquisitions with G_{RF} of 0, 10, and 20 mT/m, respectively, were $TE_{8 \mu s} = 370, 322, \text{ and } 284 \mu s$; $TE_{24 \mu s} = 378, 328, \text{ and } 288 \mu s$; and $TE_{144 \mu s} = 482, 424, \text{ and } 380 \mu s$.

The MAVRIC-SL imaging parameters (1.5/3 T) were as follows: matrix = $256 \times 142 \times 40$, FOV = $240 \times 132 \times 80 \text{ mm}$, $R = 2 \times 2$, interleaves per pass = 12, number of passes = 2, TR = 2604 ms/4601 ms, echo train length (ETL) = 24, acquisition time = 3:23 min/5:41 min.

The UTE-FE imaging parameters at both 1.5 and 3 T were as follows: RF pulse width = 100 μs , TE = 100 μs , FA = 10° , FOV = $240 \times 240 \times 240$, number of projections = 18000, acquisition time = 1:56 min.

The UTE-FPE reference images of the 3D printed plastic replica total hip replacement phantom were also acquired at 1.5 and 3 T. The acquisitions parameters were identical to the metallic total hip replacement phantom with RF pulse widths = 24 μs and $G_{RF} = 0\text{-mTm}^{-1}$ acquisitions.

In Vivo Knee Experiments

The UTE-FPE, MAVRIC-SL, and 2D-FSE acquisitions were performed in a subject with a cobalt-chromium-molybdenum total knee replacement (NexGen, Zimmer Inc, Warsaw, IN) at 3 T (MR 750, GE Healthcare) using a 16-channel wrap coil (NeoCoil). The UTE-FPE

acquisitions using both 8- and 24- μ s RF pulses were acquired to demonstrate the effect of RF pulse width on the maximum FAs, and therefore T_1 weighting.

The UTE-FPE imaging parameters were as follows: $R = 2 \times 2 \times 1$, $TR = 1.5$ ms, $TE = 370/378$ μ s (8/24 μ s), matrix = $224 \times 170 \times 40$, $FOV = 224 \times 170 \times 180$ mm, $FA = 2^\circ/6^\circ$ (8/24 μ s), $G_{RF} = 0$ mTm $^{-1}$ bandwidth = ± 250 kHz, $G_{max} = 45$ mTm $^{-1}$, and $t_{acq} = 5:32$ min.

The MAVRIC-SL acquisition parameters were as follows: matrix = $224 \times 170 \times 40$, $FOV = 224 \times 170 \times 180$ mm, $R = 2 \times 2$, interleaves per pass = 4, number of passes = 6, $TR = 1245$ ms, $TE = 6.7$ ms, $FA = 90^\circ$, $ETL = 24$, and $t_{acq} = 5:22$ min.

The 2D-FSE acquisition parameters were as follows: $TR = 4000$ ms, $TE = 8.8$ ms, $FA = 111^\circ$, $ETL = 24$, matrix = $256 \times 256 \times 30$, $FOV = 224 \times 224$ mm, slice thickness = 4 mm, slice spacing = 1 mm, number of excitations = 2, and $t_{acq} = 3:28$ min.

In Vivo Bilateral Hip Experiments

The UTE-FPE, MAVRIC SL, and 2D-FSE acquisitions were performed in a subject with bilateral cobalt-chromium-molybdenum total hip replacements (M2a Magnum, Biomet Orthopedics Inc, Warsaw, IN) at 3 T (MR 750, GE Healthcare) using a 32-channel cardiac coil (Invivo, Gainesville, FL).

The UTE-FPE acquisitions parameters were as follows: $R = 2 \times 2 \times 2$, $TR = 1.6$ ms, $TE = 286$ μ s, bandwidth = ± 250 kHz, matrix = $160 \times 160 \times 80$, $FOV = 400 \times 400 \times 240$ mm, $FA = 6^\circ$ (24- μ s RF pulse), $G_{RF} = 0$ mTm $^{-1}$, $G_{max} = 45$ mTm $^{-1}$, and $t_{acq} = 5:17$ min.

The MAVRIC-SL acquisition parameters were as follows: matrix = $160 \times 160 \times 80$, $FOV = 400 \times 400 \times 240$ mm, $R = 2 \times 2$, interleaves per pass = 4, number of passes = 6, $TR = 814$ ms, $TE = 5.9$ ms, $FA = 90^\circ$, $ETL = 16$, and $t_{acq} = 8:38$ min.

The 2D-FSE acquisition parameters were as follows: $TR = 600$ ms, $TE = 14.1$ ms, $FA = 111^\circ$, $ETL = 24$, matrix = $256 \times 256 \times 36$, $FOV = 400 \times 400$ mm, slice thickness = 4 mm, slice spacing = 1 mm, number of excitations = 2, and $t_{acq} = 5:22$ min.

Image Reconstruction and Analysis

All UTE-FPE acquisitions were reconstructed using MATLAB (v2014, The MathWorks, Natick, MA). For each data set, images were reconstructed at a variety of TEs spanning the plateau of the phase-encoding gradient. Because of the “zoom-in” effect, the FOV and resolution changes as a function of echo time. The echo time with the prescribed FOV was chosen by fitting the rate that the FOV changes using the method proposed by Jang et al. (23). Each TE image of every acquisition was reconstructed using a 3D undersampled generalized autocalibrating partial parallel acquisition (26) reconstruction with a kernel size of $7 \times 5 \times 5$.

RESULTS

The UTE-FPE images of the total hip replacement phantom acquired at three RF pulse durations (8, 24, and 144 μ s) and three maximum G_{RF} values (0, 10, and 20 mTm^{-1}) are shown at 1.5 T (Fig. 2) and 3 T (Fig. 3). Both 1.5 and 3T images showed the same general trend (Fig. 4). Specifically, for a fixed maximum G_{RF} , as the pulse duration increased, the volume of signal loss increased as a result of incomplete excitation of the metal induced off-resonance. As the maximum G_{RF} increased, signal loss slightly decreased as a result of shorter realized echo times. This effect continued until the RF bandwidth was insufficient to excite the total off-resonance caused by the combination of gradient-induced and metal-induced field perturbations. After this point, the RF pulse is no longer nonselective for all k-space locations, which results in selectivity artifacts described in (23). The 8- μ s RF pulse has a sufficiently large RF bandwidth to excite the gradient-induced and metal-induced field perturbations in the presence of 10- mTm^{-1} gradients, but not the 20- mTm^{-1} gradient. The longer RF pulses (both 24 and 144 μ s) did not have sufficiently large RF bandwidths to excite the gradient-induced and metal-induced field perturbations. As a result, the 8- μ s pulse provides excellent image quality for G_{RF} values of 0 and 10 mTm^{-1} , and starts to blur at 20 mTm^{-1} . The longer pulses present with varying degrees of signal loss and selectivity artifact when higher G_{RF} values greater than 0 mTm^{-1} were used.

A comparison of metallic implant images to the identical plastic replica images demonstrated a substantial increase in signal loss at higher field strengths for both UTE and MAVRIC-SL (spin echo) methods (Supporting Fig. S1). Although MAVRIC SL outperformed UTE at 1.5 T by capturing signal closer to the implant, their performance at 3 T was comparable. At both 1.5 and 3 T, the UTE-FE images demonstrated substantial bulk distortions resulting in a ring of signal pileup.

Improved T_1 weighting can be achieved using longer RF pulses with larger maximum attainable FAs. Peak B_1 and specific absorption rate safety constraints limit the maximum attainable FAs. Figure 4 shows UTE-FPE acquisitions using an 8- μ s ($FA = 2^\circ$) and 24- μ s ($FA = 6^\circ$) pulse in a subject with a total knee replacement. The 8- μ s RF pulse results in a proton density-weighted image, whereas significant T_1 weighting can be achieved using the 24- μ s RF pulse, as seen by the improved contrast between fat and muscle.

Both T_1 -weighted MAVRIC-SL and UTE-FPE imaging showed substantial improvements over commonly used 2D-FSE to image tissue directly adjacent to the knee prosthesis (Fig. 4). A comparison of UTE-FPE images to T_1 -weighted MAVRIC SL with matching resolution showed both techniques had similar contrast, ability to image tissue directly adjacent to the knee prosthesis, and acquisition efficiency. The arrows in this figure highlight signal pile-up and ripple artifacts seen in the MAVRIC-SL image.

Figure 5 shows a comparison of UTE-FPE to T_1 -weighted MAVRIC SL and 2D FSE in a subject with bilateral hip implants. For matching image resolution, MAVRIC SL displayed higher signal-to-noise ratio than the UTE method, although with 63% longer scan time ($t_{UTE-FPE} = 5:17$ min; $t_{MAVRIC\ SL} = 8:38$ min). The MAVRIC-SL acquisitions also

demonstrated less signal dropout than the UTE FPE, particularly at the region lateral to the femoral head.

DISCUSSION

In this work, we developed a UTE-FPE approach for imaging near metallic implants. Broadband UTE FPE allowed for a single acquisition spanning a large range of frequencies, while 3D parallel imaging enabled clinically feasible acquisition times. This approach was validated and optimized using phantom and preliminary in vivo studies. Comparisons with an existing spin echo–based 3D-MSI technique (MAVRIC SL) and 2D FSE were performed in phantoms and subjects with total knee and bilateral hip replacements.

The proposed UTE-FPE technique has similarities to the SPRITE technique (21,22). However, SPRITE requires extremely short pulses at FAs not possible on a clinical MRI for safety reasons (peak B_1 and specific absorption rate safety limitation) and high performance gradient amplitudes not available on a clinical MRI. To overcome the limitations of SPRITE, a ramped approach based on the method described by Jang et al. (23) was used. In this approach, the maximum amplitude of the gradient during RF excitation is limited. This ramped approach provides flexibility in choosing an RF pulse width that can be used to increase FAs and T_1 contrast. This ramped approach also has benefits in applications in which the gradient-induced off-resonance is large, including large FOV and off iso-center acquisitions. In these applications, a lower G_{RF} value can be used to prevent signal excitation and slice-selectivity artifacts.

The setting of optimal G_{RF} values in vivo is challenging. Factors including the FOV, the position of the FOV relative to iso-centers, and the position of the implant in the FOV all affect the maximum G_{RF} that can be achieved. In the case in which a short pulse and a large G_{RF} are used, it can safely be assumed that the primary source of off-resonance comes from the gradients being on during excitation. In the case of longer pulses with a small G_{RF} , the off-resonance from both gradients and metallic implant must be considered. A prescan such as the one proposed by Kaushik et al. (27) could be used to measure the off-resonance induced by the implant and calculate optimal G_{RF} . Alternatively, nonoptimal G_{RF} values could be used to ensure that the bandwidth of the RF pulse is sufficient to cover the off-resonance induced by the gradients and the metallic implant over the entire FOV.

The proposed UTE-FPE acquisition uses a constant TE for all phase encodes. This has the primary advantage of making UTE-FPE immune to metal-induced distortions, as the phase accumulation caused by metallic implant is identical for each phase encode. Another advantage of a constant TE acquisition is that dephasing as a result of T_2^* is the same for all k-space positions, and as a result, the resolution of the acquisition is independent of T_2^* . Conventional frequency-encoded UTE methods exhibit T_2^* -driven resolution loss (23). Short T_2^* components, the signal generally of interest in a UTE acquisition, often decay well before the readout of a center-out acquisition can be completed, resulting in the loss of high-resolution information in short T_2^* components.

The performance of the proposed UTE-FPE acquisition is highly dependent on the gradient performance. Higher performance gradients (gradient slew rate and maximum gradient strength) allow shorter TEs, TRs, and acquisition times. The effect of gradient performance on minimum attainable TE can be seen in the phantom results obtained at 1.5 T ($G_{\max,1.5T} = 33 \text{ mTm}^{-1}$; slew rate_{1.5-T} = $120 \text{ mTm}^{-1}\text{ms}^{-1}$) and 3 T ($G_{\max,3.0T} = 50 \text{ mTm}^{-1}$; slew rate_{3.0T} = $200 \text{ mTm}^{-1}\text{ms}^{-1}$). The improved gradient performance of the 3T system used in this study allowed the range of TEs to decrease from 386–622 μs (1.5 T) to 284–482 μs (3 T) at matching resolution and maximum G_{RF} . Another advantage of stronger gradient performance is substantially shorter TRs ($\text{TR}_{1.5T} = 2.8 \text{ ms}$, $\text{TR}_{3.0T} = 1.6 \text{ ms}$). These shorter TRs were enabled by the combination of shorter TEs and shorter spoiler gradients. Scanners with even higher performance gradients could acquire UTE-FPE images with shorter TEs, shorter TRs, and shorter acquisition times.

Safety compliance limits the FAs attainable using broadband RF pulses. In all acquisitions in this study, receive-only RF coils with body transmit were used. The use of transmit/receive RF coils would allow the use of higher FA safety-compliant RF pulses for improved T_1 weighting.

Long acquisition times have been a primary limitation of in vivo applications of fully phase-encoded methods near metallic implants. To overcome this limitation, broadband excitations and 3D undersampling schemes were combined to accelerate UTE-FPE acquisitions to clinically feasible acquisition times. Unlike multispectral approaches, which require multiple acquisitions to excite the off-resonance spectrum induced by a metallic implant, broadband excitations can excite a large off-resonance spectrum in a single acquisition. Three-dimensional undersampling scheme (i.e., undersampling in three phase-encoding directions rather than the traditional two phase-encoding directions) were used in conjunction with broadband excitation to accelerate UTE-FPE acquisitions into clinically feasible acquisition times. More advanced sampling schemes like CAIPIRI-NHA (28) offer the potential to improve signal-to-noise ratio or reduce acquisition times, and may be an important enhancement of the UTE-FPE approach.

The acquisition times of UTE-FPE acquisitions demonstrated in the knee and hips were comparable or faster than MAVRIC-SL acquisitions with matching resolution. In the examples, MAVRIC SL were prescribed to match the UTE-FPE acquisitions. Clinical MAVRIC-SL protocols are typically acquired with higher in-plane resolution, thinner slabs, and thicker slices (e.g., MAVRIC-SL Matrix Size_{Knees} = $320 \times 256 \times 24$, Matrix Size_{Hips} = $512 \times 256 \times 24$). The MAVRIC-SL acquisition times with high in-plane resolution and few slices tend to be faster than UTE FPE.

Perturbations to the B_1 magnetic field induced by metallic implants cause significant artifacts in both UTE-FPE and MAVRIC-SL images. These B_1 inhomogeneities result in signal shading and brightening artifacts, an effect that worsens at higher field strengths (29). Regions where long metal components run parallel to the magnetic field (i.e., stem of a hip prosthesis) are particularly susceptible (30). Although B_1 artifacts are visible in most figures, 3T images of the hip replacement phantom and subject (Figs. 3 and 5, Supporting Fig. S1) have prominent B_1 artifacts, causing both signal shading and brightening near the stem of

the hip prosthesis. Although both UTE-FPE and MAVRIC-SL images are susceptible to B_1 artifacts, they appear to manifest differently in UTE-FPE images than MAVRIC-SL images. B_1 artifacts near metallic implants can be investigated using a recently developed fully phase-encoded B_1 mapping method that is robust to both geometric distortions and signal (31).

Metal insensitive UTE techniques, such as the proposed method, excel at imaging short T_2 tissues and offer opportunities that include the examination of bone integrity or tendon and ligament ruptures in failed hip and knee replacements. Applications in which the examination of short T_2 tissues including tendons, ligaments, and cortical bone is required should use a UTE-based approach instead of 3D-MSI approaches, as short T_2 tissues will give little or no signal when using 3D-MSI techniques. Short T_2 tissues can be detected using UTE sequences, but limited contrast between short T_2 tissues and nearby long T_2 tissues often presents a challenge for UTE imaging. The addition of preparation pulses (32,33) to improve T_2 contrast would greatly improve the utility of UTE FPE. A requirement of these long T_2 suppression preparation strategies for metallic implant applications is robustness to the presence of large B_0 inhomogeneities.

CONCLUSIONS

Three-dimensional undersampled UTE-FPE acquisition with broadband excitation can generate distortion-free images near metallic implants in clinically feasible acquisition times. The trade-offs of shorter and longer RF pulses were demonstrated in a total hip replacement phantom and a subject with a total knee replacement. Shorter RF pulses captured signal closer to the implant by improving spectral coverage and allowing shorter TEs, whereas longer pulses improved T_1 weighting through larger maximum attainable FAs.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

Acknowledgments

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FIG. 1.

The proposed UTE-FPE acquisition uses phase encoding in all three spatial dimensions to prevent frequency encoding–related distortions and a gradient during RF excitation to reduce the encoding time/TE. In this acquisition, k-space is sampled at one location per TR, and k-space is sampled by cycling through all possible combinations of phase-encode gradients in three dimensions (G_x , G_y , and G_z). A maximum allowable G_{RF} prevents slice-selectivity artifacts, while still allowing excitation of a broad off-resonance spectrum near metallic implants. Temporal samples are acquired throughout the data acquisition, allowing reconstruction of images with varying FOV.

FIG. 2.

Shorter RF pulses have improved spectral coverage and shorter TEs (larger maximum G_{RF}), allowing signal closer to the implant to be captured. The UTE-FPE acquisitions acquired at 1.5 T are shown of a hip-replacement phantom for three different RF-pulse durations (8, 24, and 144 μ s) and three different maximum G_{RF} values (0, 10, and 20 mTm^{-1}), with TEs ranging from 386 to 622 μ s. The RF bandwidth and maximum G_{RF} bandwidth over all dimensions are shown for each RF pulse and G_{RF} value.

FIG. 3.

Signal excitation loss increased at 3 T through larger metal-induced B_0 inhomogeneities and T_2^* decay. Shorter RF pulses have improved spectral coverage and shorter TEs (larger maximum G_{RF}), allowing signal closer to the implant to be captured. The UTE-FPE acquisitions acquired at 3 T are shown of a hip-replacement phantom for three different RF pulse durations (8, 24, and 144 μ s) and three different maximum G_{RF} values (0, 10, and 20 mTm^{-1}), with TEs ranging from 284 to 482 μ s. The RF bandwidth and maximum G_{RF} bandwidth over all dimensions are shown for each RF pulse and G_{RF} value.

FIG. 4.

Longer RF pulses allow for improved T_1 weighting through larger maximum attainable FAs. The UTE-FPE acquisitions with a 24- μ s RF pulse (6°) of the knee offer comparable image contrast and acquisition times to a T_1 -weighted MAVRIC-SL acquisition. The UTE-FPE (8- and 24- μ s RF pulses), T_1 -weighted MAVRIC-SL, and 2D-FSE acquisitions are shown in a subject with a cobalt-chromium- molybdenum alloy knee replacement. Red arrows highlight the signal pile-up and ripple artifacts seen in the MAVRIC-SL image that are not seen in the UTE-FPE acquisition.

FIG. 5.

The UTE-FPE bilateral hip acquisitions are feasible in clinically relevant acquisition times. The UTE-FPE ($t_{\text{acq}} = 5:17$ min), T₁-weighted MAVRIC-SL ($t_{\text{acq}} = 8:38$ min), and 2D-FSE acquisitions ($t_{\text{acq}} = 5:22$ min) (a) and zoom-ins of region enclosed by red box (b) are shown of a subject with bilateral hip replacements.