MR Imaging Near Metal With 3D UTE-MAVRIC Sequences

Michael Carl[1]
[1]GE Healthcare, San Diego, CA, United States

Introduction: Magnetic resonance imaging around metal implants has been a long-standing challenge. Field inhomogeneities near implants interfere with spatial encoding and cause image degradation. Recently developed sequences such as MAVRIC [1] have enhanced the ability to image near metal implants on clinical scanners. Here we investigate the potential of using the combination of images obtained at different spectral frequencies (bins) as used with MAVRIC with ultrashort echo time (UTE) acquisitions [2,3] rather than the fast spin echo (FSE) acquisitions used with MAVRIC. The potential benefit of this approach lies in the ability of UTE techniques to directly visualize short T2 MSK tissues such as tendons, ligaments, and cortical bone, which could be of critical importance in the vicinity of orthopedic metal implants. Simulations have been used to study different RF pulse types and spectral bin strategies. Phantom and in-vivo studies were also performed.

Theory: With MAVRIC, the wide range of off-resonance frequencies encountered near metal implants is split-up into separate smaller frequency segments (bins), which are independently excited and acquired using FSE sequences. Images obtained at different resonance frequencies containing signal from different locations within the object are added (e.g. sum-of-squares, SOS) to obtain the final image. Imaging of short T2 tissues with UTE sequences is achieved by acquiring the Free Induction Decay (FID) of the MR signal as soon as possible after RF excitation. This can be accomplished by using radial center-out k-space trajectories and starting data sampling on the gradient ramp. Magnitude images can then be reconstructed from the (re-gridded) k-space data. In contrast to FSE-based MAVRIC, UTE does not utilize spin echoes, but instead operates at TE $\rightarrow$ 0, and so minimizes the accumulation of spin de-phasing due to the large static field gradients near metal implants. Symmetric Gaussian RF pulses that are used in FSE-based MAVRIC are not optimal for UTE-based MAVRIC, since they are relatively long and allow time for spin de-phasing. 3D UTE sequences usually use hard RF excitation pulses, which have a sinc-like spectral response. In the vicinity of large field inhomogeneities this can lead to banding (loss of signal) in the image. Therefore, besides hard pulses, truncated sinc RF pulses were studied.

Simulations: Figs.1A and B respectively show a hard pulse and truncated sinc pulse along with their spectral response functions (red). Figs.1C and D show the corresponding SOS combination (black) of the individual spectral responses (blue and red) using spectral bin separations of 1kHz (C) and 2kHz (D), respectively. Despite the sinc lobes of the individual spectral responses in (A), the SOS spectral response for the hard pulses is reasonably flat (C).

Methods: The phantom in our experiments (shown in Fig.3A) was a Cobalt Chromium ($\chi_m \approx$ 900 ppm) hip implant immersed in agarose gel and imaged at 3T (Signa HDx, GE Healthcare). In addition, one small water phantom and three small phantoms containing water doped with different amounts of $MnCl_2$ (resulting in measured T2 values from 3ms – 0.4ms) were placed upright near the agarose phantom. With the phantom placed in a representative orientation (implant stem aligned parallel with B_0), coronal 3D UTE images were acquired at spectral center frequencies ranging from ±8 kHz using the RF and spectral binning shown in Fig.1. Relevant other scan parameters were: TE = 56 μs, BW = ±125 kHz, FOV = 30 cm, and isotropic matrix size 256. For comparison, equivalent clinical 3D FSE images, as well as 3D FSE based MAVRIC images were obtained (BW = ±125 kHz). An in-vivo test was performed in a patient with a Cobalt Chromium hip prosthesis using hard RF excitation pulses of 400 μs duration covering a range from ±3 kHz in 1 kHz increments. Relevant other scan parameters were: TE = 56 μs, BW = ±125 kHz, FOV = 40 cm, and isotropic matrix size 256. For comparison, clinical FSE T1 images were obtained (BW = ±125 kHz).
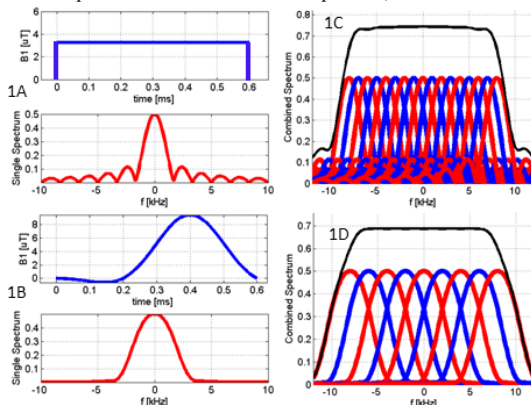

Fig.1: A) and B): RF pulse shapes (blue), and corresponding spectral responses (red). C and D: Corresponding composite multi-pulse responses (black) separated over ±8 kHz.
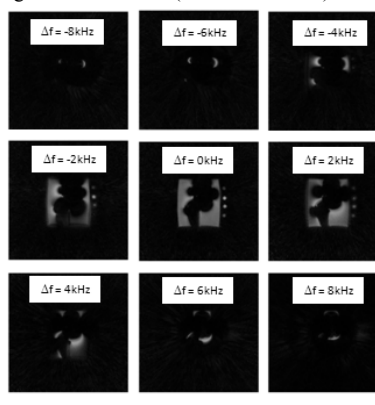

Fig.2: Coronal 3D UTE images of the phantom shown in Fig.3A at various spectral frequencies.

Results: The effects of imaging at different center frequencies are seen in Fig.2, which shows coronal 3D UTE phantom images of the same slice covering a spectral range of ±8 kHz at 2 kHz increments using truncated 600 μs sinc RF pulses (Fig.1B). Due to the dipolar nature of the susceptibility background field near the head of the implant, images obtained at positive off-resonance frequencies tend to contain more signal from the ± z direction (along B_0), while images obtained at negative off-resonance frequencies tend to contain more information from perpendicular to B_0. Fig.3 shows coronal images of the experimental phantom studies. Fig.3A shows the setup used in the phantom. Fig.3B shows a clinical 3D FSE image with large artifacts. Fig.3C shows a 3D FSE based MAVRIC image with significantly reduced artifacts and signal voids near the prosthesis. Fig.3D shows an on-resonance UTE image with hard RF pulses (Fig.1A), resulting in signal banding. For comparison, Fig.3E shows an on-resonance UTE image for the truncated sinc RF pulses (Fig.1B) showing no banding. Fig.3F shows a UTE-MAVRIC image of the SOS combination of the images shown in Fig.2. The shape of the prosthesis is significantly restored compared to the 3D FSE image. Of all the images displayed in Fig.3, only the on-resonance UTE images (D,E) and UTE-MAVRIC combination image (F) show signal from the small short and very short T2 phantoms. Fig.4 shows an in-vivo test of the UTE-MAVRIC sequence. Fig.4A shows a clinical T1 weighted FSE image, while Fig.4B shows the SOS combination of seven 3D UTE images, obtained using 400 μs hard pulses with spectral bins ranging from ±3 kHz in 1 kHz increments. The signal drop-out and artifacts are reduced compared to the FSE image.

Conclusion: We have investigated the potential of combined UTE and MAVRIC sequences to image near metal implants. Our studies show that advantages of the two techniques can be combined. We have shown substantial artifact reduction around Cobalt Chromium at 3T. We have also verified that the 3D UTE based MAVRIC spectral bin combination resulted in images containing information about short T2 components. This may improve the diagnostic capability of MAVRIC when assessing the effects of disease on tendons, ligaments and bones adjacent to implants. We expect to have similar or better success when imaging near less susceptible materials such as titanium ($\chi_m \approx$ 180 ppm) and when operating at lower field strengths such as 1.5T.
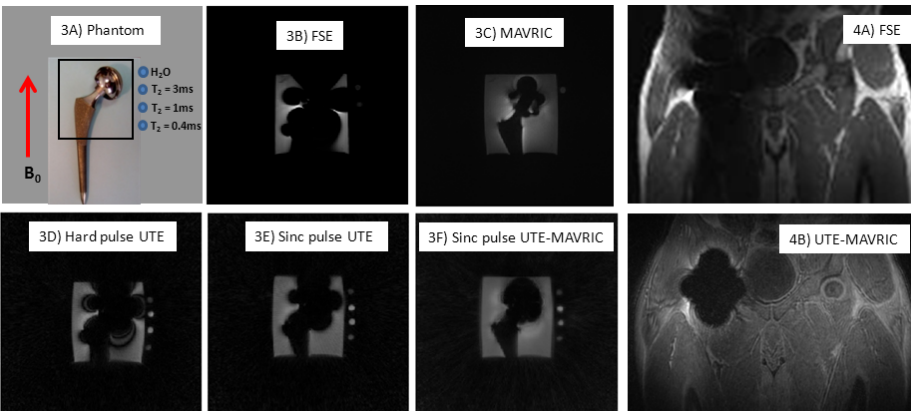

Fig.3: A) Phantom setup. B) Clinical 3D FSE image. C) FSE based MAVRIC image showing reduced artifacts. D) On-resonance UTE image for hard pulses, resulting in signal banding. E) On-resonance UTE image for truncated sinc pulses showing no banding. F) UTE-MAVRIC image of the SOS of the images shown in Fig.2. The short T2 phantoms are only visible on UTE based images.

Fig.4: A) shows a clinical T1 weighted FSE image, while B) shows the UTE-MAVRIC image combination with spectral bins of ±3 kHz.

References: [1] Koch et al, MRM, 61:381–390 (2009)
[2] Rahmer et al, MRM, 55:1075–1082 (2006)
[3] Robson et al, JCAT, 27(6): p. 825-46 (2003)