Geometrically undistorted imaging of orthopedic implants using compressed sensing accelerated phase encoded imaging

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Purpose: Clinical MR images suffer from geometrical distortions and incomplete signal excitation in the vicinity of paramagnetic objects (e.g. orthopedic implants, needles), due to the presence of strong field inhomogeneities [1-4]. Sophisticated methods have been developed in recent years (i.e. MAVRIC [1], SEMAC [2]), which improve on clinically used imaging sequences. Although these methods have been shown to significantly reduce geometric distortions, the image quality close to paramagnetic objects can still be improved by geometrically undistorted MR imaging. This is of interest for orthopedic imaging, where it may improve the detection of osteolysis, inflammation and (pseudo-) tumors near hip implants [3,4]. In this work 3D imaging of a titanium hip implant will be explored by comparing a conventional frequency encoded spin-echo (SE) technique to a purely phase encoded SE sequence to investigate the advantages of phase encoded imaging [5,6] with respect to geometrical accuracy. To decrease the inherently long acquisition time of phase encoded imaging all three spatial encoding dimensions will be undersampled, both retro- and prospectively, followed by CS reconstruction [7].

Methods: To omit frequency encoding and slice selection, a non-selective 3D SE spectroscopic imaging (SE-SI) sequence was utilized, enabling purely phase encoded imaging in all dimensions. The sequence was adapted to acquire images at short TR and TE by reducing the acquisition window, applying crusher and spoiling gradients in all dimensions to avoid spurious echoes and replacing the standard spectroscopy excitation and refocusing pulses with sinc-gauss excitation and block refocusing pulses. Prospective undersampling of the purely phase encoded images was implemented using pseudo-random sampling patterns on a Cartesian grid with a second order variable density weighting in all three spatial dimensions. Undersampled data was reconstructed using an adaptation of the CS algorithm by Lustig [7], which was adapted to enable reconstruction of 3D data. CS reconstructions were performed using both total variation and L1 wavelet norm regularization (Daubechies 6).

Imaging: Images of a 2% agarose gel with 30.5 mg/L $MnCl_2.H_2O$ containing femoral stem of a hip implant and a grid to visualize geometric distortions were acquired on a 1.5T MR system (Philips Achieva) with an 8 channel head coil. Imaging parameters for both 3D SE-SI and frequency encoded 3D spin-echo (SE) acquisitions included: acquisition matrix 128x128x16, voxel size 2x1.5x5 mm, FOV 256x192x80 mm, $B_{1,max}$ = 23 μT, TE/TR = 20/100 ms. Frequency encoded 3D-SE images were acquired in 3 min 36 sec using a 65 kHz bandwidth for the readout gradient and either selective 90 - 180^0 pulses with a 3.2 KHz excitation and 3.8 kHz refocusing bandwidth or non-selective 60 - 60^0 pulses with a 5 kHz excitation and 5.8 kHz refocusing bandwidth. The 60 - 60^0 pulse settings were chosen to excite as many off-resonance spins as possible in a single image acquisition ($\Delta\chi_{titanium}$=180 ppm, ~11 kHz off-resonance at 1.5T). Non-selective 3D SE-SI images were acquired with a spectral width of 8000Hz, 64 data points and 60 - 60^0 pulses with 5 kHz excitation bandwidth. Both fully sampled and 6x undersampled 3D SE-SI images were obtained in 7 h 16 min and 1 h 12 min 49 sec, respectively. To visualize the extent of off-resonance effects, a frequency map was created by linearly fitting ten subsequent time points of the SI phase data, which were chosen symmetrically around the echo time point. As a reference, a clinical CT scanner (Brilliance, Philips Medical Systems, Best, The Netherlands) was used to image the phantom with a FOV of 183 x 183 mm, matrix = 512x512 ,slice thickness = 1mm, gap 0.5 mm, collimation of 64 x 0.625 mm, pitch 0.671, rotation time = 0.5s. X-ray tube voltage and current were 140 KV and 337 mA with mAs/slice = 250 mA. The CT image was reformatted to have the same slice thickness as the MR images.

Results: Large distortions with respect to the grid can be observed in conventional SE images (fig 1A-B), as indicated by the arrows. The distortions are present in the frequency encoding direction and coincide with the location of the high field offsets in the frequency map (fig 1E). Increasing the excitation bandwidth from 3.2 (fig 1A) to 5 kHz (fig 1B, 1D), increases the excited frequency range from ±1.6 kHz to ±2.5 kHz around f_0. Therefore, the additionally excited spins, which range between -1.6 to -2.5 kHz or +1.6 to +2.5 kHz off-resonance, will be shifted 3.2 - 5 pixels against (-) or in (+) the readout gradient direction, respectively, taking into account the readout bandwidth of 0.5 kHz/pixel. This can be observed as additional signal pile-up in fig 1B as indicated by arrows. Other image artifacts in the non-selective SE image may be explained by the absence of additional crusher gradients. The increased excitation bandwidth resulted in nearly identical dimensions for the 3D-SE-SI image (fig 1D) as the CT reference scan (fig 1C), ensuring that the majority of off-resonance spins were excited. Furthermore, the 3D SE-SI image (fig 1D) was not affected by image artifacts, signal pile-up or bending of the grid. A factor six retrospective (fig 2A) and prospective (fig 2C) undersampling did not affect the geometric accuracy of the sampled 3D SI-SE image and can be used to decrease the acquisition time a factor six while introducing only modest errors (fig 2B).

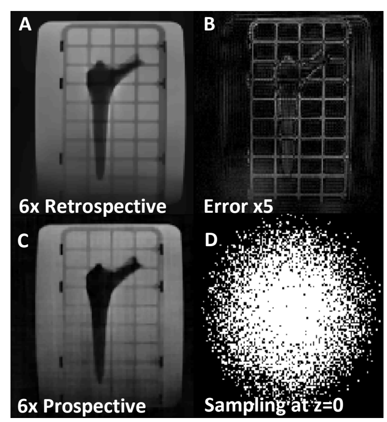

Fig 2: 6x retrospective (A) and prospective (C) CS accelerated phase encoded images. The difference of the retrospective and fully sampled images is scaled a factor 5 for visualization purposes (B). The sampling scheme for z=0 is shown (D).

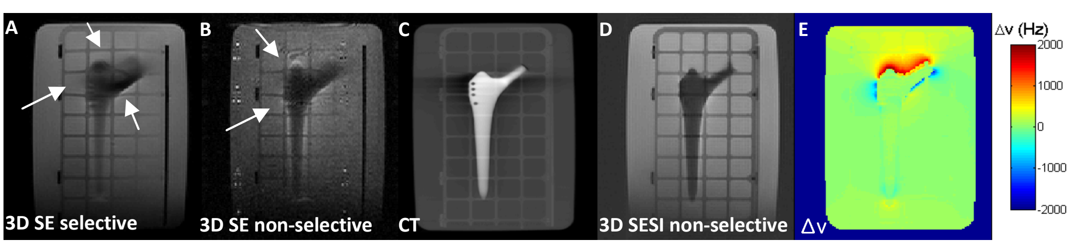

Fig 1: Conventional SE images with selective 90-180^0 (A) and non-selective 60-60^0 (B) pulses. The actual dimensions of the implant are visible on the CT reference scan (C) and can be compared to the non-selective 60-60^0 3D SE-SI image (D). Time resolved information of the 3D SE-SI data was used to obtain a frequency map (E).

Conclusion and discussion: In this work it was shown that purely phase encoded imaging can be used to obtain geometrically undistorted images near a titanium orthopedic implant. An excitation bandwidth of 5 kHz appeared to be sufficient to excite the majority of spins in the presence of the implant. Furthermore, a total acceleration factor of six was obtained using CS, which was demonstrated not to affect the geometric accuracy. The applicability of the presented technique may be extended to different metallic alloys, exhibiting larger off-resonance effects, by combination with a multi-spectral approach, as is known from MAVRIC [1]. Finally, in order to realize clinically feasible imaging times in the near future, it may be possible to limit the encoding matrix and the FOV surrounding orthopedic implants, using surface coils. Parallel imaging may also contribute to reduction of the acquisition time, however, we hypothesize that the combination of limited spatial sensitivity by use of surface coils, turbo acceleration [6] and CS will dramatically reduce total acquisition times and enable translation into the clinic.

References:

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