Three-dimensional T1-weighted MR Imaging using a One-point Dixon Technique with Arbitrary Echo Time

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Introduction

Three dimensional (3D) T1-weighted gradient echo pulse sequence is widely used in oncological and other MR imaging applications [1,2]. As in conventional spin echo T1-weighted imaging, fat suppression often improves image contrast, particularly for post contrast agent injection. Because of the short repetition time (TR) that is typically used for 3D imaging, fat suppression is achieved by applying either a chemical shift selective or spectrally selective inversion recovery RF pulse to only a small central part of the k-space. Nevertheless, these RF pulses require additional scan time, may cause SNR loss due to field inhomogeneities, and at best suppress only the low-resolution component of the fat signals.

The objective of this work is to develop a one-point Dixon technique that is capable of obtaining separate water and fat images from 3D gradient echo data without fat suppression and with an arbitrary echo time. Our hypothesis is that the one-point Dixon technique can be used to overcome the above-mentioned limitations of the existing techniques. We applied the technique in breath-hold abdominal imaging and were able to achieve clean water and fat separation with improved image quality and increased scanning efficiency.

Methods

Assuming the relative phase shift between the water and fat signals are set at 90^o and the phase errors are successfully removed, previous investigators (for example, in Ref. [3]) have shown that separate water and fat images can simply be taken as the real and imaginary part of the quadraturely-detected signal. In the presence of the phase errors, a gradient echo image acquired with an arbitrary echo time can be expressed as follows:

$$S = (W + e^{i\theta} F)e^{i\phi} \qquad [1]$$

in which W and F represent the water and fat signal intensity, respectively. ϕ is the phase error due to the field inhomogeneity and other system imperfections. θ is a user-controllable relative phase shift between the water and fat signals and is proportional to TE and the chemical shift difference between water and fat.

In order to determine W and F from Eq. [1], we extended a previously published region-growing algorithm [4] to first determine the phase factor term ($e^{i\phi}$). The algorithm uses two pre-calculated phase gradients and a series of numbered pixel stacks to help guide an optimal growth path so that regions with smoother phase changes are grown before regions with larger phase changes. At each step of the region growing, the phase factor of a given pixel is determined based on a value estimated using both the amplitude and phase of the neighboring pixels. To start the region growing, an initial seed is selected from an area with the most uniform phase (i.e. the smallest phase gradients). This area is then identified as either water or fat by monitoring whether the abrupt phase changes due to the intrinsic phase θ encountered during the following region growing is leading or lagging the initial seed pixel. All the abrupt changes due to the intrinsic phase θ are removed in the region growing process [4]. Once the phase factor is determined on a pixel-by-pixel basis, it is smoothed with low-pass filtering. Afterwards, the phase factor is eliminated from Eq. [1], resulting in a phase corrected image $S' = Se^{-i\phi}$. Finally, the water and fat images are decomposed from S' as follows:

$$F = \mathrm{Imag}(S')/\sin(\theta) \qquad [2]$$

$$W = \mathrm{Real}(S') - F\cos(\theta) \qquad [3]$$

Experiments and Results

The image reconstruction program using the phase correction algorithm is implemented in Matlab (MathWorks, Natick, MA). Data using a fast spoiled 3D gradient echo sequence with and without spectrally selective inversion recovery fat suppression were acquired on a 1.5T whole body clinical scanner (Waukesha, Wisconsin) and with a 4-channel torso-phased array coil. Breath hold time for fat suppressed and non-fat suppressed acquisitions was 21 seconds and 14 seconds, respectively. Other scan parameters were kept identical and listed as follows: TR/TE = 4.2/1.7ms, flip angle = 15^o, acquisition matrix =384x110x32, receiver bandwidth = ±83.3 kHz, FOV= 36x24.75cm, slice thickness = 5mm.

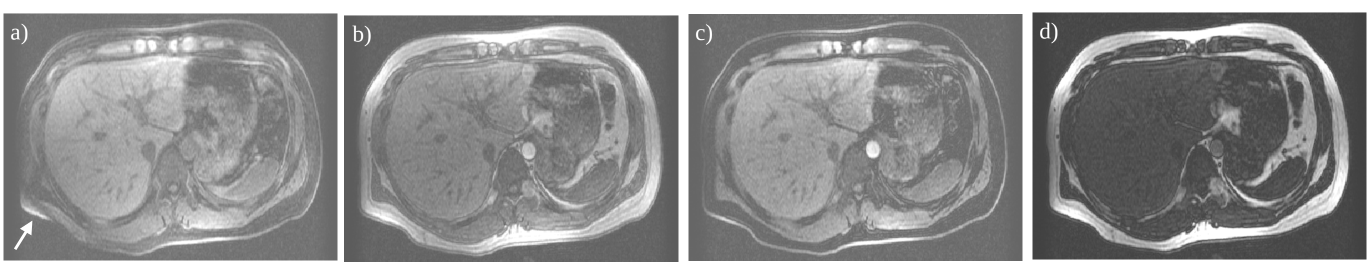

Fig. 1a-d) show for a representative slice, the fat suppressed, non fat-suppressed, processed water-only, and processed fat-only images, respectively. Because of the large FOV, fat suppression in Fig. 1a) is not uniform (for example, at the location by the arrow). Additionally, the image appears somewhat “foggy” because the fat suppression RF pulse is applied only at the center part of the k-space (by default, every 32 TR per RF pulse). In comparison, the two processed images (Fig. 1c and 1d) show very clean water and fat separation. Details of the organ structures are better depicted than in Fig. 1a), particularly for anatomy outside the liver.

Conclusions

Typical scan time savings for T1-weighted 3D images without fat suppression vs. with fat suppression average around 4-7 seconds, which can be important for effective breath hold for a patient. Here, we demonstrated that with proper phase correction, the data collected without fat suppression can be further processed into separate water-only and fat-only images. In comparison to the image acquired with conventional fat suppression, the processed water-only image shows better uniformity in fat suppression and noticeably improved image quality.

References

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