Under-Relaxed Image Restorative Technique for Na^{23} MRI

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Abstract

To improve signal-to-noise ratio in sodium image, short echo time (2-3 ms) and long data acquisition (10-20 ms) protocols are used. Sodium in biological specimens demonstrates a bi-exponential decay of transverse magnetization and the fast decaying component of the sodium signal results in the reconstruction of images which are blurred significantly. The spatially-dependent nature of the blurs are due mainly to the presence of short local transverse relaxation values (0.7-3 ms) of sodium in tissue. We present an algorithm that corrects for object-dependent blurs due to fast-decaying T2 and improves the computational behavior of the algorithm by incorporating a relaxation parameter into the iterative process.

Introduction

All magnetic resonance images are blurred as a result of an inherent decaying of nuclear magnetic signals during data acquisition due to spatially-varying and object-dependent transverse relaxation (T2). The extent of the blur depends on the distribution of transverse relaxation time of the object and the data acquisition time used in the pulse protocol. Compared to the strength of proton magnetic resonance signal from a biological organism, sodium signal is inherently weak. A method of improving signal-to-noise ratio in sodium magnetic resonance imaging is to perform asymmetric sampling of gradient-echo signal so that images with short echo time (2-3 ms) and narrow bandwidth may be acquired [1].

However, a rapid biexponential decay of sodium signal during long data acquisition period, especially due to the presence of fast transverse relaxation in natural endogenous sodium in tissue, results in fast T2-dependent blurring of reconstructed images.

For short echo time (TE) and extended total data sampling period, two phase-encoded gradient echo sodium signals (using positive and negative readout gradients, respectively) are acquired separately, and then synthesized to form a full gradient echo signal. Such a signal is expressed in complex phasor notation

$$\iint \left[I_f(x,y)\, e^{-(|t|/T2_f(x,y))} + I_s(x,y)\, e^{-(|t|/T2_s(x,y))} \right] \cdot e^{-(i\gamma G_x xt)}\, e^{-(i\gamma G_y ym)}\, dxdy \tag{1}$$

where

$$I_f(x,y) = M_f(x,y)\, e^{-(TE/T2_f(x,y))} \tag{2}$$

is the fast $T2_f$-weighted spin density distribution, and

$$I_s(x,y) = M_s(x,y)\, e^{-(TE/T2_s(x,y))} \tag{3}$$

is the slow $T2_s$-weighted spin density distribution. M_f and M_s are spin density distributions associated with $T2_f$ and $T2_s$, respectively. t is a psuedo-time variable where

$t=0$ at the center of the echo. The terms $e^{-|t|/T2_f(x,y)}$ and $e^{-|t|/T2_s(x,y)}$, which are responsible for the blurs in the reconstructed image, describe the decay of signal during data acquisition due to the presence of local fast and slow T2, respectively. Both simulation and experimental studies have shown that the first term is the main contributor to the blur and the second term negligible.

We have previously described a method that reduces T2-dependent blurring in images due to mono-exponential decay of transverse magnetization [2]. The general idea behind the method is to iteratively refine our estimate of the part of the signal which give rise to spatially-varying T2-dependent blurring in the images -- using estimates of the distribution of spin density and T2 -- and subtracting that signal from the original signal.

The algorithm presented here adheres with the general scheme described above but with the addition of a relaxation parameter u of the Iterative Data Refinement (IDR) [3]. We have translated the algorithm into the general framework of IDR, and a relaxation parameter introduced into the algorithm [4]. The relaxation parameter results in many cases in a much improved computational behavior of the algorithm. Under such a scheme, the Riemann sum approximation of the integral

$$\iint \left[I_f^{(k)}(x,y)\left\{u\, e^{-(|t|/T2_f^{(k)}(x,y))} - 1\right\} + I_s(x,y)\right] \cdot e^{-(i\gamma G_x x p)} e^{-(i\gamma G_y y m)}\, dxdy \qquad (4)$$

is calculated and subtracted from the measured signal (1) which has been multiplied by a constant u. Images reconstructed by inverse Fourier transform using the subtracted signal should be an improved estimate of the T2-weighted spin density distribution, with reduced blurring. The iterative procedure proceeds by calculating (k-th) estimates of $M_f(x,y)$ and $T2_f(x,y)$ from the corrected images to yield the next improved estimate of (4), to be used for further refinement.

Method

A sodium phantom was designed so that it consists of objects that exhibit anisotropic blurring, an object that does not blur and a non-object (a hole), all surrounded by a background material which produces a weak bi-exponential sodium signal. A mixture of saline and agarose gel simulates the bi-exponential relaxation of sodium in biological tissue. A schematic diagram illustrating the construction of the sodium phantom is shown in Figure 1.

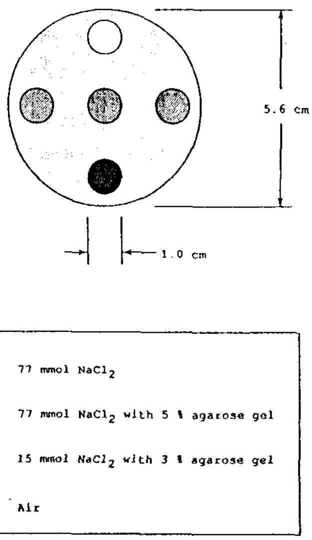

Fig. 1. Schematic diagram of construction of a sodium phantom.

The background material, a mixture which consists of 14 mmol aqueous solution of $NaCl_2$ and agarose concentration of 3% (by weight), was filled into a 5.6 cm diameter plastic cylindrical container. The material was measured to have a fast and a slow T2 values of approximately 6 and 36 ms, respectively. A projection technique was used to determine the T2 values of the sodium phantom. A row of three test tubes (1.0 cm in diameter and 4 cm in length) filled with a mixture of 77 mmol of aqueous solution of $NaCl_2$ and agarose concentration of 5% were embedded in the background material. These phantoms were measured to have a fast and a slow T2 values of approximately 5 and 36 ms, respectively. This row of phantoms would demonstrate anisotropic blurrring due to fast T2 effects. Two test tubes, one filled with 77 mmol of aqueous solution of $NaCl_2$ and another with air, were also embedded in the background material. 77 mmol saline solution was measured to have a T2 value of 56 ms.

All MR imaging was performed on a 1.9 Tesla superconducting magnet with a 16 cm clearance for imaging. The system operates at 21.41 MHz for sodium imaging. A RF rotating-field sodium coil was used to image the phantom. A pulse sequence which acquires phase-encoded asymmetric gradient-echo signals was implemented. In one pulse sequence, 64 data samples of an asymmetric gradient-echo signal are collected over a period of 10 ms. 128 steps of phase-encoding are performed. In a separate sequence (with the x-gradients reversed), the other half of the asymmetric signal is acquired. As before, 64 data samples are collected and 128 steps of phase-encoding are done. The pulse sequences may be repeated to average signals in order to improve SNR. The samples of one half of the asymmetric signal (in terms of Fourier transform, the negative spatial-frequency component of the signal) are "time-reversed" and the two half echoes acquired separately are combined together to form one complete gradient-echo signal.

Asymmetric sodium signals were obtained at echo times 1.7, 2.2, 2.7, 3.2, 3.7, 4.2, 4.7, 5.7 and 6.7 ms. Sodium gradient echo signals with longer echo times were not acquired because the images were beginning to show noticeable distortions and loss of signals from the effects of magnetic field inhomogeneities. In the pulse sequence, a narrow bandwidth of the signal is achieved by using a weak readout gradient over an extended time period, and in this situation, weak gradient echo signals at long echo time are more susceptible to contaminations from the effects of field inhomogeneities. Repetition time (TR) of 300 ms and bandwidth of 3.2 kHz were used. All images were obtained transaxially, with slice thickness of approximately 1 cm. Averaging of 4 signals were done to improve SNR. The imaging time required to collect all the signals was approximately 90 min.

If any meaningful values of fast and slow T2 are to be determined from a least-squares-fit method, data points at echo times in the range of both T2 must be available. Using only the 9 available images at relatively short TE, the least-squares-fit would give erroneous values of slow $T2_s$. Since the integral aim of the experiment is to reduce the fast T2-dependent blurs in the sodium images, we need only the estimates of fast $T2_f$ and M_f (spin density associated with $T2_f$) from the images to carry out the blur-correcting iterative process. Decomposing sodium images into fast and slow T2-weighted images were not possible with the data available.

Sodium signals from a mathematical sodium phantoms which mimiced the physical phantom described above were also simulated. For our simulation studies, we made the assumption that noise in the MR signal is Gaussian and additive. Although the assumption is not necessarily correct, it is a reasonable one for the purpose of simulating a noisy MR signal. A Gaussian noise with a mean zero and a stardard deviation of one was added to the simulated signals so that the average background noise in the reconstructed simulated (magnitude) images was approximately the same as the level of background noise in the experimental sodium images. The same iterative method to reduce the fast T2 blurs in the experimental images was also used in the simulated images.

Results and Discussion

A method of monitoring the convergence of an iterative algorithm is to evaluate a Figure of Merit (FOM) which is defined to be the logarithm of a normalized sum of squares distance (between the ideal and corrected signal). For simulation studies, since we are able to calculate the ideal signal (the sodium signal with no effects from the decay of signal during data acquisition), such FOM may be calculated for every iteration. Figure 2 shows the FOM plots for the over-relaxation (ω=1.5), no-relaxation (ω=1.0) and under-relaxation (ω=0.5).

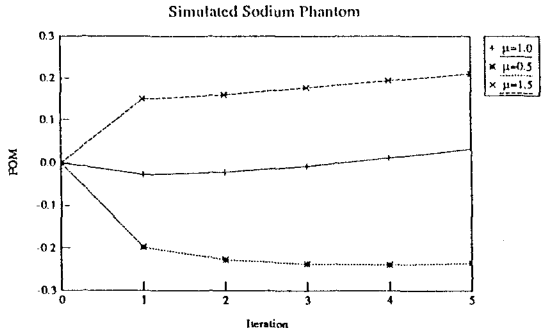

Fig. 2. Plots of Figure of Merit (FOM) versus number of iterations for the case of over-relaxation (ω=1.5), no-relaxation (ω=1.0) and under-relaxation (ω=0.5).

Over-relaxation of the algorithm resulted in deterioration of FOM while no-relaxation shows little improvement after one iteration. Consistent improvement of FOM is observed for the case of under-relaxation only. Although not shown here, the increased high frequency noise in the images is evident especially when ω=1.0 and 1.5, and after 5 iterations, structured image artifacts (in the background phantom) are clearly seen in the images.

Figure 3 shows the result of the experimental studies for images obtained at TE=1.7 ms. Uncorrected image (top left) shows the three saline/agar phantoms in a row exhibit horizontal blurring, as expected, due to T2 effects. Corrected images after 1 iteration using ω=1.0 (top right), ω=0.5 (bottom left) and ω=1.5 (bottom right) are shown. As in the simulation studies, the under-relaxation of the algorithm yielded images with reduced T2-dependent blurring and minimal high frequency artifacts.

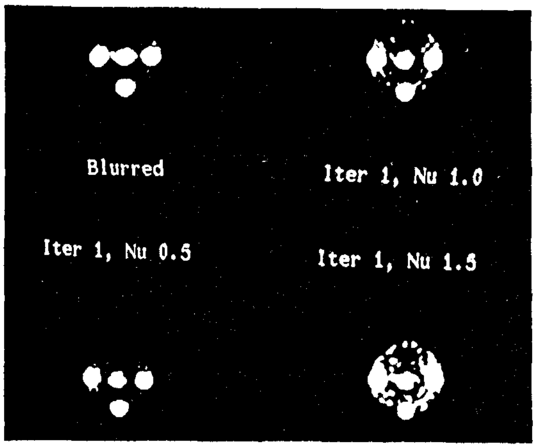

Fig. 3. Experimental sodium images are displayed using a narrow display window level to demonstrate the reduction of blurs in the corrected images.

Conclusion

The results of the algorithm using both the mathematical phantom and the physical phantom complement each other in that similar conclusion may be drawn from both studies. The two studies, done is parallel, show that the algorithm with under-relaxation produces desirable results, in the sense that, the fast T2 dependent blurs are reduced with small enhancement of high frequency noise in the images.

References

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