Fast direct image reconstruction for MRI and fMRI in the presence of field inhomogeneities and T2* decay.

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Introduction. In order to accurately interpret functional magnetic resonance imaging (fMRI) data, it is imperative the underlying images be free from reconstruction artifacts. It is well known that presence of field inhomogeneities causes geometric distortions, while T_2^* decay causes blur and drop-outs. We have developed an algorithm for reconstructing MR images from data acquired with rectilinear EPI trajectories. This algorithm simultaneously corrects for the phase evolution caused by the magnetic field inhomogeneity, and the signal decay caused by T_2^* relaxation. We have implemented a multi-echo FLASH sequence to collect accurate field and T_2^* maps needed as inputs to the algorithm. We have applied this method to the reconstruction of data acquired using a single-shot EPI sequence at 3T, demonstrating the methods' artifact reduction capabilities.

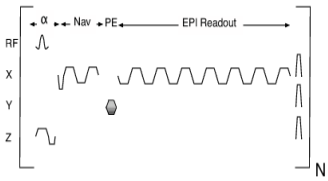

Figure 1. Schematic diagram of the field and decay mapping pulse sequence.

Mapping B0 and T_2^*. The field inhomogeneity and T_2^* mapping sequence was implemented as a 32-echo FLASH sequence, consisting of a slice-selective excitation followed by three navigator readouts, a y-phase encoding, and train of readouts 1 ms apart (figure 1). Complex images were reconstructed from each echo. For each pixel, the local field B(x) and decay constant $T_2^*(x)$ were estimated in Matlab using non-linear least squares fitting of the data to the model $\rho(x)\exp(i\,B(x)t - t/T_2^*(x))$.

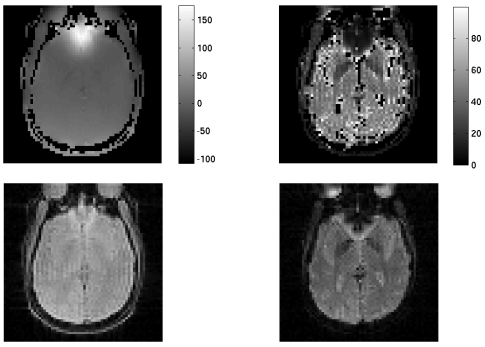

Figure 2. Top: field and T_2^*.maps, bottom: magnitude images of echoes corresponding to TE=7 and 30 ms.

Image reconstruction. We consider image reconstruction of data collected using a single-shot EPI sequence. Given maps of B(x) and T_2^* (x), the image reconstruction proceeds in a manner similar to that proposed by Kadah [1]. Each frequency encode line is reconstructed separately ignoring field inhomogeneity induced phase evolution and decay. This results in a semi-reconstructed image ρ(x,ky). Then for each x, the pseudo-inverse of the forward model containing decay and phase evolution in the phase encode direction (ky) is applied. In other words, we are assuming that the signal evolution during the acquisition of one line is due to the applied gradients only. This separation of the evolution into two time scales is reasonable because: (1) the field inhomogeneity is much smaller than the bandwidth per pixel (~100 Hz versus ~1000 Hz); and (2) the signal decay is small, (time constant is of the order of several tens of milliseconds, i.e. the signal decay during a readout of a single line is on the order of 1%). Using this approach for obtaining 64x64 images, the reconstruction problem is reduced to 64 sub-problems, each of which requires inverting a 64x64 matrix, a definite advantage over the full inversion of a 4096x4096 matrix. For the reconstruction of a time series of such images, the time consuming matrix inversion preparation step is performed only once, the matrices are stored, and the reconstruction proceeds as above. The algorithm is implemented in Matlab.

Results and Discussion. Data were collected from a human using a Siemens Allegra 3T scanner. The field and T_2^* mapping sequence parameters were TR=2500 ms, TE=23 ms, field of view 192x192 mm with 64x64 resolution, slice thickness 3 mm. Figure 2 provides example images and maps. Single-shot EPI sequence was employed to collect data with the same slice prescription, TE=30 ms, bandwidth=2894 Hz/pixel and echo spacing of 400 ms (figure 3). It is evident from figure 3 that correcting for field inhomogeneity removes distortions, and correcting for T_2^* decay adjusts the contrast, as expected. Note the accuracy of the inhomogeneity correction as reflected in the phase images. The EPI readout starts 18 ms after the excitation pulse, therefore the fast decaying signal components are lost prior to acquisition, e.g. the signal from the basal ganglia cannot be recovered.

In order to eliminate the field inhomogeneity and decay reconstruction artifacts, we measured and incorporated them into the image reconstruction algorithm. The method does not assume that the inhomogeneity and decay maps are smooth. Figure 3 demonstrates the effectiveness of the approach. Although many groups have reported algorithms for correcting for field inhomogeneity (see [2] and references therein), to our knowledge, less attention has been paid to T_2^* blur. The importance of our proposed method for fMRI, is highlighted in the simulation results in figure 4 where negative and non-local artifacts are present when images are obtained with the conventional reconstruction algorithm. This artifact appears as a modulation of image intensity in regions surrounding the actual activity, and may be misinterpreted as a positive/negative BOLD effect correlated with the stimulus. It is seen that the artifacts are reduced when images are reconstructed with the presented algorithm.

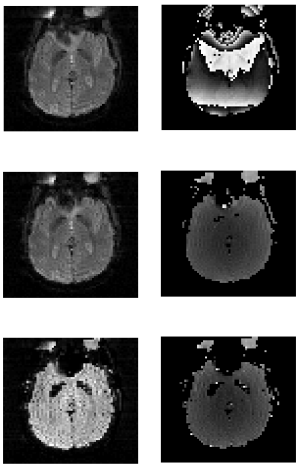

Figure 3. Left to right: magnitude and phase images. Top to bottom: images produced by conventional reconstruction, correcting for field inhomogeneity only, correcting for both inhomogeneity and decay. Phase encode direction is left to right. (The same slice as in figure 3 is presented, phase images are masked at 0.1 of magnitude.)

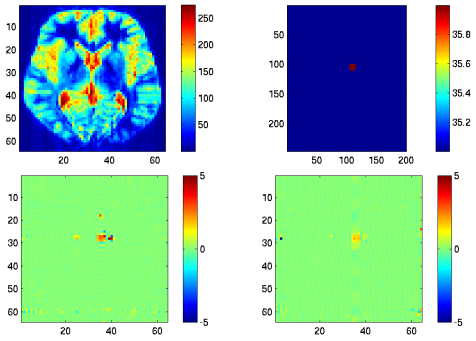

Figure 4. Simulation of fMRI experiment. Top panels show fine structured object and the region where simulated BOLD activity takes place (decay constant changes from 35 ms to 36 ms). Bottom panels show recovered percent change activity at 64x64 resolution, left: conventional reconstruction produces non-local and negative BOLD artifacts, right: artifacts are diminished when image is reconstructed with the presented algorithm.

Conclusion. We have presented a pulse sequence for accurate measurement of field and decay maps and a fast direct EPI image reconstruction algorithm utilizing them to eliminate geometrical and blur artifacts. We have identified non-local image reconstruction artifacts that may occur in fMRI studies if the T_2^* decay is ignored in the reconstruction. The field and T_2^* mapping sequence itself can find a stand-alone use in clinical applications for quantitative perfusion studies.

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References.

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