Rapid Deblurring for spiral fMRI

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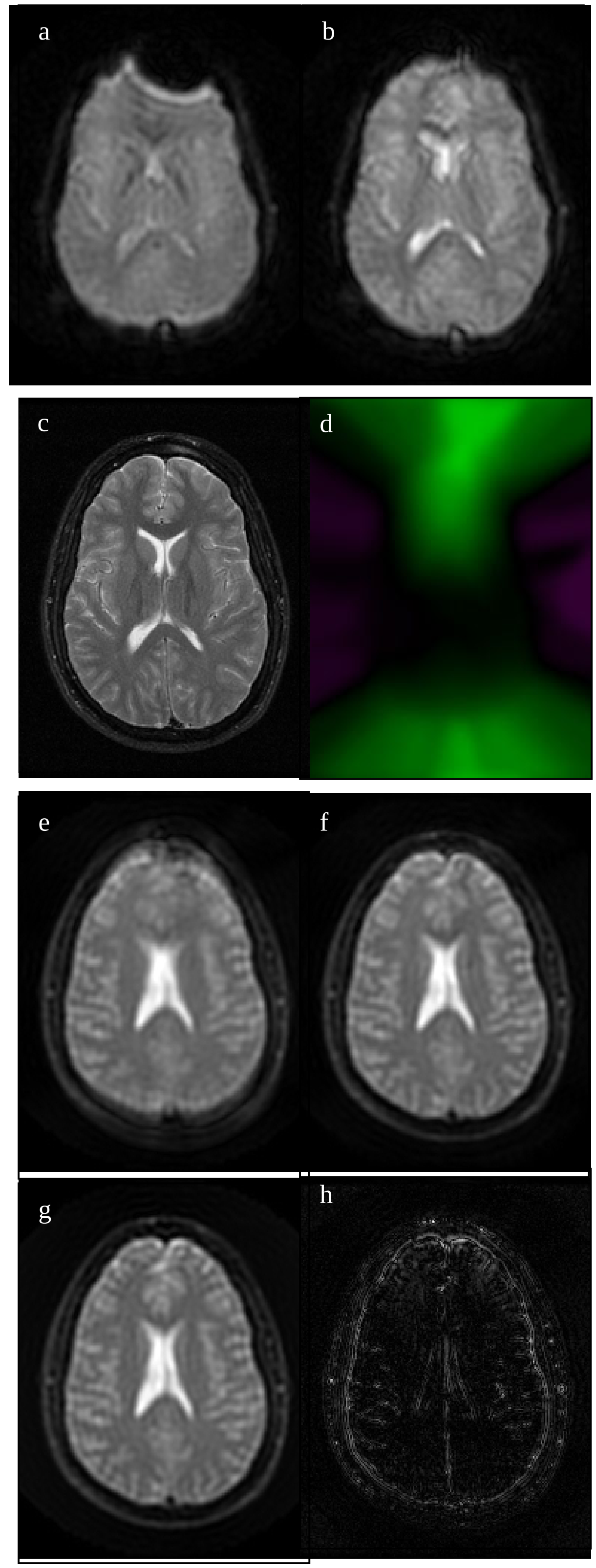
Introduction: Spiral imaging gives fMRI data without geometric warping, although blurring that occurs due to off-resonance. This work builds off a one-dimensional convolution correction algorithm that is orders of magnitude faster than traditional conjugate phase methods, and presents the efficacy of this method under different conditions. Typical fMRI datasets require thousands of images and despite current speedups, many other approaches to this problem are computationally expensive.

Theory: In nearly constant off-resonance fields, the acquired signal in *k*-space has merely accumulated an additional phase shift according to the time of acquisition of the sample. In spiral imaging the time of acquisition of a sample can be approximated as a function of distance from the center of *k*-space (i.e. $t = c\,(k_x^2 + k_y^2)$); therefore *k*-space has been multiplied by a complex exponential function with an approximately parabolic phase. This is equivalent to convolving the data in image space with a separable blurring kernel. Deconvolution of this blurring was introduced in [1]. The work discussed here adds to the deconvolution method by also convolving in progressive steps, as well as iteratively deblurring the field map. The former is done by deblurring up to (e.g.) 10 Hz and then deblurring again up to another 10 Hz where appropriate, etc. Areas that are more on-resonance finish deblurring first. It attains a very good correction of the image where fields are smooth, no matter the magnitude. Though breakdown of the algorithm occurs where magnetic fields vary quickly, such areas often coincide where the signal that would be blurred is lost due to T2* decay, therefore preserving the integrity of the algorithm. The number of computations to implement the algorithm is $2k_{ave}N^2$ beyond typical reconstruction, where k_{ave} is the average deconvolution kernel length. This average kernel length varies from 1 to 22 pixels in fields of 60 Hz off-resonance or less.

Methods: Data were collected on a 1.5 T GE EXCITE scanner with echospeed-plus gradients. The first two scans of each slice had different TE's, from which field maps were calculated. The average off-resonance was removed by subtracting the average accumulated phase of each sample from the original signal (this value could also be calculated to minimize faster changing fields). Images are then reconstructed and progressively convolved with the deblurring kernels in both directions, producing the corrected images (this is first done with the field map images, and the field map is regenerated). Also, data were simulated using the off-resonance MR equation directly into a spiral *k*-space trajectory and then reconstructed via the algorithm.

Results: Datasets have been collected from phantom and in vivo data as well as using simulations of illustrative data and simulations of acquired in vivo data. In vivo datasets were collected and a sample is illustrated in Fig. 1. A comparison T2 weighted image is also included for reference.

References: [1] Mag Res Med 44:491.



Figures 1. Blurred in vivo image (a), deblurred version (b), comparison T2 weighted image (c) (also used for simulation), associated field map (d), simulated blurred image using same field map (e), deblurred version (f), simulated image with no off-resonance (g), difference (x2) between deblurred simulation (f) and reference (g) (h).