High angularly resolved diffusion imaging with accelerated multi-shot acquisition and compressed sensing

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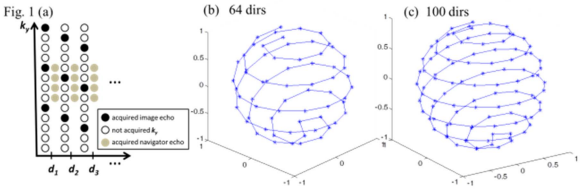
Target Audience: MR physicists and brain researchers with an interest for novel diffusion-imaging schemes.

Purpose: High Angular Resolution Diffusion Imaging[1] (HARDI) is a diffusion encoding scheme for resolving intra-voxel fiber structures and crossings. It typically involves acquiring as many as a few hundred diffusion weighted images using a single-shot EPI sequence. The main factors limiting its usefulness and clinical applicability may be the length of the acquisition process and the vulnerability of the single-shot EPI sequence to geometric distortions, especially at high field strengths. The purpose of the present work is to introduce a fast HARDI method mostly insensitive to distortion by accelerating the acquisition both in k-space (to reduce distortion) and in the diffusion-encoding space (to reduce acquisition time).



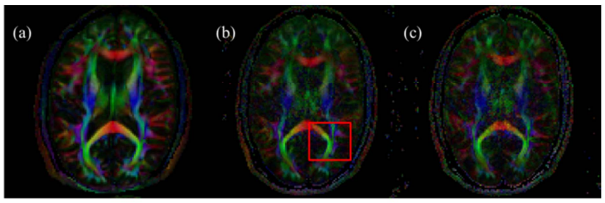


Fig. 2 R = 2 ; 100 dirs R = 4 ; 100 dirs R = 4 ; 64 dirs

Methods: The present approach was inspired by work done with compressed sensing to reduce acquisition time in diffusion imaging[2], and by a recently-published acceleration strategy for multi-shot EPI to reduce distortion[3]. Both approaches were conceptually combined here to achieve considerable reductions both in distortion and scan time by exploiting both signal sparsity and correlations among diffusion signals. Diffusion-weighted images are first reconstructed with motion correction[3,4] and artifact removal[3] according to:

$$o = \text{argmin}_o \|Eo - s\|_2^2 - \lambda^2 \|L^{-1} o\|_2^2 \ , \qquad [1]$$

where s is the partially sampled k-space of diffusion images, o the reconstructed images, E the encoding operator and L the regularization information from navigator echoes.[3] The resulting diffusion-weighted images are treated with the compressed sensing inspired CFARI[2] algorithm to resolve the intravoxel fiber orientations:

$$f = argmin_{f:f_i \in [0,\infty)} \|Df - o\|_2^2 + \beta \|f\|_1, \quad [2]$$

where D denotes a set of modeled diffusion weighted signals and f_i represents the weighting of the i-th component in D (512 prolate-shaped bases were used, the ratio of the three principal diffusion coefficients was set to 4:1:1, bases were homogenously distributed over a unit spherical shell, a first pass through the algorithm identifies 64 dominant bases while a second pass improves accuracy by considering only these bases along with similarly-oriented bases).

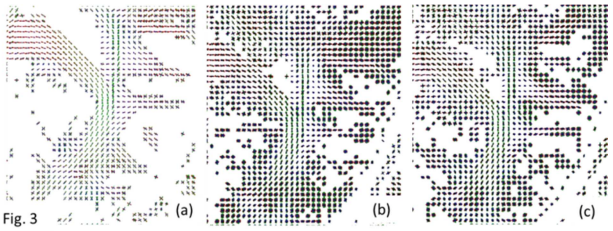


Fig. 3

Volunteer experiments were performed on a 3 T system following informed consent. A b-value setting of 1000 s/mm^2 was chosen here. A reference protocol was designed to provide relatively low distortion levels using a commercially-available sequence: 100 diffusion directions, two fully-sampled b = 0 images, two-fold parallel imaging acceleration, TE/TR = 72/3000 ms, FOV = 22 cm, slice thickness = 4 mm, matrix size = 128×128, Q-ball imaging reconstruction[1]. The protocol was then modified to further reduce distortion and scan time: Acceleration in k-space was increased from two- to four-fold (see black dots in Fig. 1a), the number of sampled diffusion directions was either 100 or 64, one navigator signal per slice per TR for motion compensation and data were reconstructed as described above. Figures 1(b) and 1(c) illustrate the sampling trajectories that were employed to obtain 64 and 100 diffusion directions, respectively. The trajectories in Figs 1(b) and 1(c) were designed intending to provide smooth signal variation along *k-d* space.

Results: Color FA maps (Fig.2) of one slice and orientation maps (Fig.3) of the indicated region in Fig. 2 are shown from a same volunteer for all three different settings. The distortion with two-fold k-space acceleration in Fig. 2(a) is slightly higher than that of Fig. 2(b) and 2(c), with four-fold acceleration. Acquisition times for data in Fig. 2(a) and Fig. 2(b) both took 5:45, while that for data in Fig. 2 (c) was 3:50(about 40% shorter). As could be expected, SNR drops occurred the further the acquisition was accelerated.

Fig. 3 shows the reconstructed orientation from the region highlighted in Fig. 2(b). Overall, similar image resolution, FA, and orientation distribution was retained through the addition of k-space and diffusion-space acceleration. Similar peak orientation was maintained in both uni-structural fiber-crossing regions, although some degree of shape swelling can be noticed in the accelerated versions in Figs 3(b) and 3(c).

Discussion: Results in Figs. 2 and 3 suggest that the proposed method could accelerate HARDI acquisition and retain sufficient quality to resolve fiber crossings. Distortion levels were decreased by four-fold with the proposed approach compared to non-accelerated EPI acquisitions, while total scan time was reduced by 40%. The regularization weightings in the reconstruction have not been optimized yet, and could offer a means to trade-off some spatial resolution for further SNR.

Conclusion: A method was presented for faster and less distorted HARDI imaging through acceleration in k-space and diffusion-encoding space.

Reference: [1] D. S. Tuch, Mag Res Med, 52:1358-72, 2004. [2] B. A. Landman et al, NeuroImage, 59:2175-86,2012. [3] B. Madore, Mag Res Med, [Epub], 2012. [4] D. Atkinson, et al, Mag Res Med,56:1135-39, 2006.

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