Experimental Verification of Increased Diffusion Sensitivity with Hyperechos

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Introduction

For a given b-value, the diffusion sensitivity is optimal when the signal is at a maximum for a given set of relaxation rates T_1, T_2 and the local diffusion D of the tissue in question. The influence of these parameters in a pulse sequence depends not only upon the pulse sequence timing parameters, but also on the coherence pathways that contribute to the final signal [1]. Hennig introduced the concept of the hyperecho [2], in which a sequence of radio-frequency (RF) excitation pulses of arbitrary flip angle, phase, and intermediate gradient pulses following an initial 90° pulses will refocus to a full echo following a 180° inversion pulse if they are applied with conjugate symmetry. The simplest hyperecho sequence is equivalent to a standard diffusion-weighted stimulated sequence with the introduction of a 180° refocusing pulse into a standard diffusion weighted stimulated echo sequence. In a recent paper, we presented theoretical results that a two-pulse diffusion weighted hyperecho sequence has the advantage of producing equal or greater signal-to-noise (SNR) than a standard diffusion weighted stimulated echo sequence with equivalent diffusion weighting [3]. Thus, in theory, the two-pulse hyperecho sequence produces greater diffusion sensitivity than the stimulated echo sequence. This study presents experimental verification of the improved diffusion sensitivity produced by the two-pulse hyperecho sequence.

Results are shown in the application to high angular resolution diffusion encoding (HARD) [3-5] in two normal human brains. To demonstrate diffusion sensitivity, representative fractional anisotropy (FA) maps are shown for each imaging sequence and for each subject. When compared with an equivalent stimulated echo acquisition, data acquired with a hyperecho acquisition produce higher fractional anisotropy values in white matter regions.

Method

Images were acquired on a 1.5 T GE Signa LX MRI scanner equipped with a 22 mT/m, actively shielded, 120 mT/m/ms gradient system located at the VA San Diego Healthcare System. Diffusion images were acquired on two normal human volunteers, with approval from the Humans Subject Committee at UC San Diego using a diffusion sequence, which employs a spiral readout. The spiral acquisition designed by Li and Kim [6,7], permits high-resolution diffusion-weighted MRI using a variable density spiral, which allows self-navigating due to the oversampling of the center of k-space. The pulse sequence was designed to operate in stimulated and hyperecho modes that enable keeping all imaging and diffusion weighting parameters the same except for those defining the excitation modes.

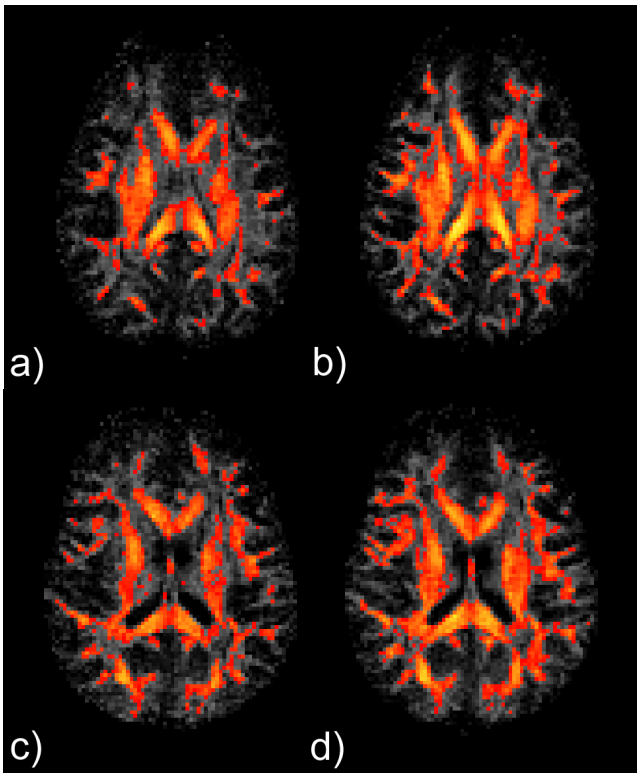

Figure 1. Fractional anisotropy maps derived from HARD encoded (a,c) stimulated echo and (b,d) hyperecho images acquired with identical diffusion weighting for volunteer 1 (a,b) and volunteer 2 (c,d).

High angular diffusion encoding (HARD) was achieved by generating gradient directions equally spaced on a sphere by tessellations of an icosahedron [3,8]. Images were acquired at 3 slices with the following parameters: FOV = 24 cm, slice thickness = 3.8 mm, image matrix 128×128 (voxel size 1.875 mm × 1.875 mm × 3.8 mm), TR = 750 ms, TE = 76 ms, TM = 35.7 ms, and four interleaves. The diffusion parameters were: diffusion gradient duration, δ = 30 ms, mixing time Δ = 72.4 ms, and b ≈ 2000 s/mm2. Fifteen averages at each diffusion direction were collected to ensure high signal-to-noise ratios, and resulted in a total scan time of approximately 33 minutes for each sequence.

Results

Fractional anisotropy maps were determined from the diffusion tensor for each data set. The diffusion tensor was estimated by fitting the full data from the six unknown in the diffusion tensor via linear regression. Representative FA maps created from data acquired with the stimulated echo sequence (Figures 1a,1c) and the hyperecho sequence (Figures 1b,1d) is shown for each volunteer. Each figure consists of a gray scale image overlayed with a color map displaying FA > 0.4.

The hyperecho FA maps show that more pixels pass the 0.4 threshold in fiber structures than the stimulated echo maps. This is clearly represented in regions, such as the corpus callosum, in which the intravoxel fiber orientation is essentially unidirectional. When compared with an equivalent stimulated echo sequence, data acquired with a hyperecho acquisition produce an improved representation or sensitivity of the local anisotropy as seen by the increased FA values.

Conclusion

A common theme in DW MRI is the necessary tradeoff between diffusion sensitization and the resulting loss in SNR that degrades the data. While Hennig's hyperecho concept is quite general [2], we have focused this work on the simplest case of a two-pulse hyperecho because it can be directly compared with standard diffusion echo. We have experimentally shown that there is improved diffusion sensitivity in the two-pulse diffusion weighted hyperecho sequence when compared with the standard stimulated echo sequence, which supports theoretical results [3].

References

1. Hennig J, MRM, **78**:97-407 (988).
2. Hennig J, *et al*., MRM, **46**:6-12 (2001).
3. Frank LR, *et al*., MRM, **49**:1098-1005(2003).
4. Frank LR, MRM, **45**:935-939 (2001).
5. Frank LR, MRM, **47**:1083-1099 (2002).
6. Li T, *et al*., ISMRM Scientific Meeting, **69** (2003).
7. Kim D, *et al*., MRM **50**:214-219 (2003).
8. Tuch DS, *et al*, ISMRM Scientific Meeting, **321** (1999).