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Development and initial evaluation of 7-T *q*-ball imaging of the human brain

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Abstract

Diffusion tensor imaging (DTI) noninvasively depicts white matter connectivity in regions where the Gaussian model of diffusion is valid but yields inaccurate results in those where diffusion has a more complex distribution, such as fiber crossings. *q*-ball imaging (QBI) overcomes this limitation of DTI by more fully characterizing the angular dependence of intravoxel diffusion with larger numbers of diffusion-encoding directional measurements at higher diffusion-weighting factors (*b* values). However, the former technique results in longer acquisition times and the latter technique results in a lower signal-to-noise ratio (SNR). In this project, we developed specialized 7-T acquisition methods utilizing novel radiofrequency pulses, eight-channel parallel imaging EPI and high-order shimming with a phase-sensitive multichannel B_0 field map reconstruction. These methods were applied in initial healthy adult volunteer studies, which demonstrated the feasibility of performing 7-T QBI. Preliminary comparisons of 3 T with 7 T within supratentorial crossing white matter tracts documented a 79.5% SNR increase for b=3000 s/mm^2 (P=.0001) and a 38.6% SNR increase for b=6000 s/mm^2 (P=.015). With spherical harmonic reconstruction of the *q*-ball orientation distribution function at b=3000 s/mm^2, 7-T QBI allowed for accurate visualization of crossing fiber tracts with fewer diffusion-encoding acquisitions as compared with 3-T QBI. The improvement of 7-T QBI at *b* factors as high as 6000 s/mm^2 resulted in better angular resolution as compared with 3-T QBI for depicting fibers crossing at shallow angles. Although the increased susceptibility effects at 7 T caused problematic distortions near brain–air interfaces at the skull base and posterior fossa, these initial 7-T QBI studies demonstrated excellent quality in much of the supratentorial brain, with significant improvements as compared with 3-T acquisitions in the same individuals.

Keywords: Diffusion tensor imaging (DTI); High-field MR imaging; Fiber tractography; High angular resolution diffusion imaging (HARDI); *q*-ball imaging (QBI); White matter

1. Introduction

There has been continuous progress in magnetic resonance (MR) diffusion imaging technology over the past two decades, revolutionizing scientific studies of the human central nervous system and the clinical practice of neuroradiology. However, despite the development of higher-performance gradient systems, multielement phased array coils and parallel imaging, as well as the latest generation of clinical 3-T high-field scanners, diffusion MR imaging of the human brain in vivo has remained limited by signal-to-noise ratio (SNR) constraints. This is because the progression of new applications for diffusion imaging has created ever-increasing demands for spatial resolution and angular resolution. Much recent interest has focused on diffusion tensor imaging (DTI) [1,2] and its use in the noninvasive investigation of white matter connectivity with fiber tractography [3–5], which requires high resolution in three spatial dimensions. Moreover, the shortcomings of DTI for accurately characterizing diffusion in complex white matter, where fiber tracts with different orientations intersect or are otherwise partial volume averaged within a voxel, have led

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to the advent of high angular resolution diffusion imaging (HARDI). HARDI methods attempt to recover more information about the angular structure of intravoxel diffusion by acquiring larger numbers of diffusion-encoding directional measurements at higher diffusion-weighting factors (b values) than are routine for DTI [6–8]. There has been a growing list of proposed techniques for reconstructing fiber orientations from HARDI data, including spherical harmonic modeling of the apparent diffusion coefficient profile [7,8], multitensor modeling [9], generalized tensor representations [10,11], persistent angular structure [12], circular spectrum mapping [13], q-ball imaging (QBI) [14,15], spherical deconvolution [16], fiber orientation estimated on continuous axially symmetrical tensors [17] and the diffusion orientation transform [18]. However, all of these HARDI approaches remain SNR limited even on 3-T scanners with parallel imaging capability because of the need for both high spatial resolution and very strong diffusion weighting, resulting in long examination times and/or suboptimal discrimination of intravoxel crossing fiber tracts. Indeed, the low SNR of HARDI at 3 T has motivated the introduction of different forms of regularization to improve the reliability of fiber orientation determination [19–21]. Proposed methods for more fully sampling q-space often require even higher diffusion-weighting factors than are typical for HARDI [22–24]; hence, these techniques might derive even greater benefits from better SNR.

The most straightforward strategy for improving SNR is to boost the magnetic field strength B_0, thereby increasing proton spin polarization. However, to our knowledge, there has been no report in the peer-reviewed literature of diffusion imaging of the living human brain at field strengths greater than 3 T. Challenges to diffusion-weighted imaging (DWI) at ultra-high field include T_2 and T_2^* shortening, large chemical shifts, poorer field homogeneity and increased susceptibility artifacts. The goals of this project were to develop specialized 7-T single-shot spin-echo (ssSE) EPI q-ball acquisition methods and to investigate feasibility and performance in initial human studies.

2. Materials and methods

2.1. Image acquisition

HARDI was performed on three healthy adult male volunteers (ages 26, 36 and 38 years) using a 3-T Signa EXCITE scanner (GE Healthcare, Waukesha, WI, USA) equipped with a commercial eight-channel phased array head receiver coil (MRI Devices, Waukesha, WI, USA). The volunteers also underwent HARDI on a 7-T MR scanner (GE Healthcare) equipped with a 30-cm volume excite coil and an eight-channel phased array receiver (Nova Medical, Wilmington, MA, USA). The imaging protocols were approved by the institutional review board of our medical center, and written informed consent was obtained from all participants. At both 3 and 7 T, a multislice axial ssSE echoplanar (EP) pulse sequence was employed for HARDI. The 7-T version of the sequence incorporated a custom-designed high-bandwidth fat-saturation pulse for effective lipid suppression. Automated high-order shimming was performed prior to the diffusion-weighted acquisitions using a specialized multichannel B_0 field map reconstruction that preserves phase information when combining signals from the different phased array coil elements [25]. Both the 3- and 7-T scanners were equipped with gradient coil sets that have a maximum amplitude of 40 mT/m and a maximum slew rate of 150 T/m/s. At each field strength, conventional Stejskal–Tanner diffusion encoding was applied in each of two HARDI protocols, one with 36 diffusion directional measurements at b=3000 s/mm^2 [repetition time (TR)=5 s; echo time (TE)=89 ms; number of excitations (NEX)=1], and the other with 131 diffusion directional measurements at b=6000 s/mm^2 (TR=6 s; TE=108 ms; NEX=1). For both protocols, an additional acquisition without diffusion weighting at b=0 s/mm^2 was also obtained. The total scan time for the 36-direction experiment was 6 min 30 s; that for the 131-direction experiment was 26 min 40 s. For SNR calculation at 3 and 7 T, two identical 6-direction acquisitions were performed at b=3000 s/mm^2 and b=6000 s/mm^2 using the same sequence parameters as described above for the two HARDI protocols. In all cases, the diffusion-encoding directions were distributed uniformly over the surface of a sphere using electrostatic repulsion [26]. Parallel imaging of the diffusion-weighted data acquired with the eight-channel phased array head coils at 3 and 7 T was accomplished using the array spatial sensitivity encoding technique (ASSET; GE Healthcare) with an acceleration factor of 2. For all acquisitions, spatial resolution was 2 mm in all three dimensions (field of view=256×256 mm; 128×128 matrix; 2-mm-thick interleaved slices with no gap). In addition, for all acquisitions, the supratentorial brain from the vertex to below the level of the corpus callosum (CC) was covered with axial slices oriented along the plane passing through the anterior and posterior commissures.

2.2. Image postprocessing

Tensor analysis of the HARDI data sets was performed in DtiStudio v2.4 [27], including generation of color fractional anisotropy (FA) maps using the standard red–blue–green directional encoding convention. The q-ball orientation distribution function (ODF) at each voxel was computed from the HARDI data using spherical harmonic basis functions with methods we have described in detail previously [19]. Truncating the spherical harmonic series at a low order enables q-ball ODF reconstruction from relatively few diffusion-encoding directional measurements with little noise amplification, but achieving a higher angular resolution with larger numbers of directional measurements requires higher harmonic orders in the reconstruction. The penalty of including higher harmonic orders is greater noise amplification. Hence, in the fast acquisition regime at b=3000 s/mm^2