Assessment of Reduced Encoding Diffusion Spectrum Imaging Implemented with a Bi-Gaussian Model Using Phantoms and Manganese-Enhanced Optic Tracts

Abstract

Diffusion spectrum imaging (DSI) can map complex fiber microstructures in tissues by characterizing their 3-D water diffusion spectra. However, a long acquisition time is required for adequate q-space sampling to completely reconstruct the 3-D diffusion probability density function (PDF). Furthermore, to achieve a high q- or b-value encoding for sufficient spatial resolution, the diffusion gradient duration and diffusion times are usually enlarged on a clinical scanner, which results in a long echo time and low signal-to-noise ratio (SNR) of diffusion images. To bypass long acquisition times and strong gradient requirements, the reduced-encoding DSI with a bi-Gaussian diffusion model (RE-DSI) is presented in this study. The bi-Gaussian extrapolation kernel, which is based on the assumption of a bi-Gaussian diffusion signal curves across biological tissues, is performed to fulfill a high q-value request requirement on the reduced-encoding scheme. Both the intersecting capillary phantoms and the manganese-enhanced rat models were served as standards for accuracy assessment in RE-DSI. The errors of RE-DSI in defining fiber orientation were quantified.
and the results were found to be close to the noise limit. Evidence from a human study demonstrated that RE-DSI significantly decreased the acquisition time, required to meanwhile resolve complex neural fibers. The presented acquisition method facilitates the application of DSI analysis on a clinical magnetic resonance imaging (MRI) system.

*Keywords: diffusion spectrum imaging; phantom model; manganese-enhanced rat model*
Introduction

Diffusion MRI has become an essential tool for contrast imaging of the central nervous system. This has led to a significant improvement in clinical diagnosis. Further advancement to the technique has been made with the introduction of diffusion tensor imaging (DTI) [1, 2]. The technique makes further progress along with the design of diffusion tensor imaging (DTI), which is a feasible valuable technique for identifying anisotropic diffusion as well as non-invasively delineating the principle orientations of white matter tracts non-invasively [3-5]. However, the assumption of a single Gaussian diffusion component in the tensor model results in the ambiguous orientations of fibers in regions where they cross each other containing crossing fibers [6]. Thus, with the typical resolution of a MRI, it may be difficult to interpret the complex neural connections between functional areas of the human brain with under typical resolution of an MRI.

In recent years, various diffusion imaging strategies have been developed to improve the depiction of water diffusion and to resolve the intravoxel fiber orientations. Diffusion spectrum imaging (DSI) [7], for example, utilizes the 3-D spectra of water displacements to characterize the heterogeneities of fiber architectures. DSI was based on the q-space imaging technique, which The theory describes the Fourier relationship...
between echo signal attenuation and the probability density function (PDF) of the displacement of water molecules with the prerequisite of a narrow pulse approximation [8-10].

The DSI technique has been used to map the tissue architecture of biological systems DSI has shown its capability of mapping tissue architectures in biological systems [7, 11]; specifically, it has provided the intravoxel compartment scales of the neural fibers [12], thus allowing and interpreting the physiological and structural conditions of the neural tissues to be interpreted. In addition, 3-D tractography and comparative segmentation of human brain structures have been identified based on DSI and the proceeding orientation distribution function (ODF) [13].

Notwithstanding the utility of DSI, a complete reconstruction of the diffusion PDF requires 515 q-value encoding points distributed on a Cartesian lattice across 3-D q-space. This involves long acquisition times as well as adequate q-values for sufficient resolution. Since the available gradient strength in clinical systems are limited, the latter requirement is achieved by prolonging the diffusion gradient duration (δ) and the diffusion time (Δ) since the available gradient strength in clinical systems is limited. Unfortunately, this leads to a long echo time (TE) and a decline in the SNR level due to a severe T2 decay in the echo. 

Comment [WL4]: CHECK: Do you mean “intravoxel component” here instead of “intravoxel compartment?” This follows on throughout the article, do you mean “component” or “compartment?”

Comment [TK5]: CHECK: This sentence is confusing. It seems you are saying that tractography and comparative segmentation of brain structures were identified with DSI. Then you talk about what appears to be another technique, ODF. However, it is not clear how this technique is related to DSI or was it just another technique used in conjunction with DSI? Consider rewording to clarify. Also, if ODF is another technique (in addition to DSI), which has been helpful in reconstructing tissue structure, it may be good to point that out. At the moment, it doesn’t quite tie into the rest of the paragraph but just appears all of a sudden at the end.

Comment [TK6]: CHECK: The meaning of this portion of the sentence is not clear.

Comment [TK7]: CHECK: Consider not using the word “adequate”. Perhaps use ‘large quantities of’ if that is appropriate.
planar imaging (EPI) sequence. As a consequence of this, the angular accuracy and discrimination would be unavoidably diminished as a consequence [11]. Both lengthy acquisition times, cost and the requirements of the gradient system have retarded the further applications of DSI on clinical scanners. These limitations basically stems from the need to exhaustively sample on a 3-D Cartesian sampling lattice.

A hemispheric encoding scheme (half-q-DSI) can be applied to halve the scan time in DSI since the diffusion contrast is positive and spherical (and thus symmetrical) [7, 14]. However, uncorrected cross-term interactions between diffusion and imaging gradients might result in a misunderstanding of the q-space analysis and inaccurate ODFs in half-q-DSI [15, 16]. Instead of a Cartesian lattice, a body-centered cubic lattice (BCC) sampling scheme was proposed to improve the imaging efficiency of DSI by 30% [17]. Another non-Cartesian q-space encoding scheme, hybrid diffusion imaging (HYDI), has also been employed for DSI-PDF reconstruction. This scheme comprised of five concentric spherical shells and may be applied to multiple types of diffusion analyses [18]. Although it was possible to shorten the acquisition times with all of the above q-space
sampling strategies described above, the need requirement for a large number of high q-values to preserve adequate spatial resolution acquisitions could not be omitted to preserve adequate spatial resolution.

Another category of diffusion imaging techniques utilizes an encoding scheme formed by a single spherical shell with a constant diffusion weighting, as opposed to the 3-D Cartesian lattice with multiple diffusion weightings. These techniques include high angular resolution diffusion imaging (HARDI) [19, 20], q-ball imaging (QBI) [21, 22], persistent angular structure MRI (PAS-MRI) [23], fiber orientation estimation using continuous axially symmetrical tensors (FORECAST) [24], diffusion orientation transformation (DOT) [25], and spherical deconvolution methods [26, 27]. These approaches provide information on the orientation directional information of complex neural fiber networks within a feasible reasonable scan time and may be routinely implemented. The substantially increase in imaging efficiency mainly results from the fewer numbers of diffusion-weighted images (DWIs) needed for data analysis. In addition, the shortened TEs following on a moderate b-value could enhance the SNR of DWIs. These conditions, however, may be insufficient to characterize the 3-D diffusion function that is derived from the multiple q-value diffusion measurements, and would thus be unable to infer tissue microstructural
tissue conditions shape and orientations. However, they might be insufficient to characterize the 3-D diffusion function derived from the multiple q-values diffusion measurements.

In this study, it is proposed that the reduced-encoding DSI implemented complemented with a bi-Gaussian model (RE-DSI), is proposed to be used to trim down the drawbacks of DSI as well as while retaining q-space information. In RE-DSI, a reduced Cartesian sampling scheme, where high q-value acquisitions are omitted, is used to bypass long acquisition times and gradient system demands in DSI. To achieve sufficient resolution for determining the fiber orientations of fibers, the 1-D bi-Gaussian model fitting is performed on this applied to the sampled data at low q-space to regain all diffusion signals at high q-space. Previous studies on animal and human brains have demonstrated that diffusion-attenuated curves could be characterized as a bi-exponential function [28-31]. Accordingly, we hypothesized that the diffusion signal attenuation along each radial direction in q-space was a bi-Gaussian function. This assumption is similar to that used in the DOT technique [25], which straightforwardly converts the diffusivity function into displacement probabilities at a particular distance away from the origin, while RE-DSI tends to reconstruct a diffusion PDF from q-space signals.
The performance of RE-DSI in terms of clearly defining the orientations of coherent and heterogeneous fiber orientations was assessed using the capillary phantom models and the manganese-enhanced (ME) rat models, both of which these models have been previously utilized to validate the DTI and DSI methods/techniques [5, 11]. The magnetic resonance (MR) images of capillary phantoms and the ME rat optic tracts were served as the standard for the measurement of angular uncertainties in RE-DSI. The results showed that the consequential resulting PDF profiles and ODF patterns reconstructed by RE-DSI were comparable to those achieved by DSI. The merits of RE-DSI in a clinical environment are twofold: (a) the scan time can be remarkably shortened to a half or a quarter of its original time; and (b) DWIs can be acquired with better SNR under limited gradient intensity. To demonstrate the clinical feasibility of RE-DSI, this technique was performed on a healthy subject at the end of this study.

Comment [WL20]: CHECK: This is the first time “MR” is used apart from “MRI”. I assume it stands for “magnetic resonance”.

Comment [TK21]: CHECK: Consider revising. Suggestions: ‘DWIs can be acquired with a limited gradient intensity, leading to a better SNR’ OR ‘better SNR allows for acquisition of DWIs under limited gradient intensity’
Materials and Methods

Reduced-Encoding Scheme

In DSI, the q-space sampling scheme consists of 515 diffusion wave vectors (q), where \( q = \gamma gT \) \( (\gamma \) gyromagnetic ratio, g: gradient vector, \( T \) duration of diffusion gradient), placed on a Cartesian lattice within a sphere that has a radius of five lattice units. The framework of the q-space acquisition scheme in RE-DSI is the same as that of DSI, except that the encoding wave vectors are within a radius of three or four lattice units. The resultant encoding numbers are 257 and 123, respectively.

Bi-Gaussian Model

The bi-Gaussian model assumption in RE-DSI was established according to several studies on bi-exponential analyses of high b-value diffusion data [28-31]. The bi-exponential diffusion model, also called the or so-called two-compartment model, ascribes the contribution of MR signal attenuation to the weighted sum of fast and slow water diffusion. The general formula is shown as follows [32]:

fitting of undersampled data. The slightly increased error of 4.70° ± 3.51° for the case of 45° crossing in RE-DSI [123] might have resulted from fewer q-space acquisitions.
CHECK:

Do you mean “can enhance” or “could enhance”? If they always enhance it, use “can enhance”.