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Original contribution

A theoretical validation of the B-matrix spatial distribution approach to diffusion tensor imaging

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ABSTRACT

The recently presented B-matrix Spatial Distribution (BSD) approach is a calibration technique which derives the actual distribution of the B-matrix in space. It is claimed that taking into account the spatial variability of the B-matrix improves the accuracy of diffusion tensor imaging (DTI). The purpose of this study is to verify this approach theoretically through computer simulations.

Assuming three different spatial distributions of the B-matrix, diffusion weighted signals were calculated for the six orientations of a model anisotropic phantom. Subsequently two variants of the BSD calibration were performed for each of the three cases; one with the assumption of high uniformity of the model phantom (uBSD-DTI) and the other taking into account imperfections in phantom structure (BSD-DTI). Several cases of varying degrees of phantom uniformity were analyzed and the distributions of the B-matrix obtained were used for the calculation of the diffusion tensor of a model isotropic phantom. The results were compared with standard diffusion tensor calculation.

The simulations confirmed the improvement of accuracy in the determination of the diffusion tensor after the calibration. BSD-DTI improves accuracy independent of both the degree of uniformity of the phantom and the inhomogeneity of the B-matrix. In cases of a relatively good uniformity of the phantom and minor distortions in the spatial distribution of the B-matrix, the uBSD-DTI approach is sufficient.

1. Introduction

Diffusion tensor imaging is a powerful technique with multiple clinical applications. It can provide additional information about anatomical details which are not achievable by means of any other imaging techniques [1]. White matter fiber tracking and fractional anisotropy (FA) maps can be useful during pre-surgical planning as well as in intraoperative MRI, especially in the case of brain and spinal cord tumors [2–5]. DTI can also improve the diagnostic capabilities of MRI in cases of neurodegenerative disorders like epilepsy [6,7], multiple sclerosis [7,8], Alzheimer's disease [9–11], ischemic stroke [12,13] and other brain injuries which cause changes in the diffusion properties in a given brain region [1]. A novel potential application of DTI emerges in the field of brain functional MRI, which can be faster and more accurate than BOLD fMRI [14].

The DTI experiment requires at least six diffusion weighted imaging (DWI) measurements with non-collinear directions of the diffusion sensitizing gradients and one additional reference measurement, usually performed without any diffusion gradient. The parameters of these gradients for a given sequence are incorporated in the 3×3 symmetric matrix for each gradient direction, so called B-matrix [15]. It should be pointed out that the B-matrix also depends weakly on other factors like imaging gradients, eddy current effects and other background interference [15–17]. Therefore, it is virtually impossible to calculate the B-matrix analytically taking into account all of those effects and thus commercial systems usually provide its approximated form. Such a practice can lead to systematic errors and a decrease in accuracy. There are several ways of indicating the B-matrix more accurately, e.g. refocusing each diffusion gradient before the imaging gradient is turned on and refocusing each imaging gradient before the next diffusion gradient [18] or acquiring data twice in each direction; once with the given diffusion gradient and once with the opposite polarity [19,20]. However, applying one of these methods entails prolonged acquisition time. The imaging gradient effect can also be reduced by establishing the optimal diffusion gradient scheme [21].

Another important fact is that, in general case, due to the inhomogeneity of the gradients which affect the B-matrix, it is not spatially constant [22–24]. For diffusion imaging, the spatial variability of the B-matrix results in serious inaccuracies of the quantitative characterization of the diffusion. Most of the effects caused by gradient uniformities can be removed by means of an

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Table 1
Diffusion gradient components along readout (G_R), phase (G_P) and slice (G_S) encoding directions for each of the six diffusion gradient directions.

Index	G_R	G_P	G_S
1	1.00	0.00	0.00
2	0.45	−0.90	0.00
3	0.45	−0.28	0.85
4	0.45	0.72	−0.53
5	0.45	0.72	0.53
6	−0.45	0.28	0.85

approximation of the magnetic field using spherical harmonic expansion [22,25].

The BSD-DTI (B-matrix spatial distribution in DTI) is an alternative method of determining the exact form of the B-matrix. The actual values of the matrix in a particular region of interest (ROI) are derived experimentally with the use of an anisotropic phantom with a well-defined structure. Unlike previous methods, in this technique, knowledge of almost all imaging sequence parameters is not required. In order to proceed, the BSD calibration phantom is situated inside an MRI scanner and the standard DTI measurement is performed. Subsequently the phantom is mechanically rotated using a certain set of Euler angles and the measurement is repeated to complete the data set [23,26].

Previous studies [23,27,28] were performed under the assumption of the high uniformity of the applied phantom. It means that the diffusion tensor was assumed to be the same across its entire volume. We recently called this approach uniform BSD-DTI or just uBSD-DTI. This method is simpler to implement and less time-consuming, but on the other hand, it requires using phantoms of very high quality, which are expensive and difficult to manufacture.

Nevertheless, the BSD-DTI method is not limited only to such cases. If the spatial distribution of the diffusion tensor is well known in the coordinate system associated with the phantom, it can be derived in the laboratory coordinate system by a rotation transformation [29]. This approach allows us to use basically any anisotropic phantom with well-known diffusion properties. The only requirement is that the phantom parameters must be constant during the calibration.

In this study we have examined the impact of the B-matrix spatial variability and the anisotropic phantom imperfections on the accuracy of the uBSD-DTI and BSD-DTI approaches and compare the results with standard DTI.

2. Material and methods

The simulation process was divided into three stages. The first concerned establishing the simulated experiment conditions, the structure of the virtual anisotropic phantom, patterns of the B-matrix spatial distribution and the outcome of the diffusion tensor imaging under such assumptions. The second pertained to the BSD calibration and the third to the threefold calculation of the diffusion tensor. The workflow presenting the whole simulation process was depicted in Fig. 8.

2.1. The preparation stage

In order to simulate a realistic experiment, six diffusion gradient directions were taken from a clinical scanner (GE 3.0 T Discovery 750 MR) DTI sequence (Table 1). The standard B-matrix (constant in the entire imaging volume) for each diffusion gradient direction was derived accordingly to the formulas proposed by Mattiello [15] for the particular sequence parameters (Eqs. (1a) to (1f)). The simulated in-plane image size was 25 × 25 pixels with 25 slices without interleaves. The BDS calibration procedure does not depend on the applied imaging protocol, therefore the presented study remains valid for any other set of parameters.

$$b_{RR} = 3.51 + 136.89G_R + 2228.52G_R^2 \tag{1a}$$

$$b_{PP} = 3.19 + 130.63G_P + 2228.52G_P^2 \tag{1b}$$

$$b_{SS} = 1.76-58.50G_S + 2228.52G_S^2 \tag{1c}$$

$$b_{RP} = b_{PR} = 3.17 + 68.45G_P + 65.31G_R + 2228.52G_RG_P \tag{1d}$$

$$b_{RS} = b_{SR} = -1.84-29.25G_P + 65.31G_S + 2228.52G_PG_S \tag{1e}$$

$$b_{PS} = b_{SP} = -1.93-29.25G_R + 68.45G_S + 2228.52G_RG_S \tag{1f}$$

The set of six equations above forms a B-matrix:

$$B = \begin{bmatrix} b_{RR} & b_{RP} & b_{RS} \\ b_{PR} & b_{PP} & b_{PS} \\ b_{SR} & b_{SP} & b_{SS} \end{bmatrix} \tag{2}$$

The B-matrix Spatial Distribution calibration rests on the assumption that the B-matrix varies spatially, so for each gradient direction and each point in space, and thus for each voxel of the image, an individual B-matrix should be derived. In order to reproduce such conditions, the standard B-matrix was spatially altered with three different patterns. Each pattern was generated by the multiplication of the diffusion gradient along three orthogonal directions (R, P, S) with one of the pattern functions listed below:

$$p_1 = \sigma\cdot(R + P + S-36)/10 \tag{3a}$$

$$p_2 = \sigma\cdot(|R-12| + |P-12| + |S-12|-13.83)/4.05 \tag{3b}$$

$$p_3 = \begin{cases} \sigma\cdot(R + P + S-36)/10, & \text{for : } G_R, G_S \\ \sigma\cdot(36-R-P-S)/10, & \text{for : } G_P \end{cases} \tag{3c}$$

Fig. 1. Visualization of the pattern $p_1 = \sigma\cdot(R + P + S - 36)/10$ across three slices. From left to right: $R = 1$, $R = 13$, $R = 25$, respectively.