Magnetic Resonance Imaging 33 (2015) 286–295

Contents lists available at ScienceDirect

Magnetic Resonance Imaging

journal homepage: www.mrijournal.com

Improving the accuracy of PGSE DTI experiments using the spatial distribution of b matrix ☆

Artur Tadeusz Krzyżak [a,*], Zbigniew Olejniczak [b]

[a] AGH University of Science and Technology, Kraków, Poland
[b] IFJ PAN, Kraków, Poland

ARTICLE INFO

Article history:
Received 27 February 2014
Accepted 21 October 2014

Keywords:
Magnetic resonance imaging (MRI)
Diffusion tensor imaging (DTI)
Diffusion anisotropy
Anisotropic diffusion phantom
Matrix b
Diffusion tensor
B-matrix spatial distribution
Cross-terms

ABSTRACT

A novel method for improving the accuracy of diffusion tensor imaging (DTI) is proposed. It takes into account the b matrix spatial variations, which can be easily determined using a simple anisotropic diffusion phantom. In opposite to standard numerical procedure of the b matrix calculation that requires the exact knowledge of amplitudes, shapes and time dependencies of diffusion gradients, the new method, which we call BSD-DTI (B-matrix spatial distribution in DTI), relies on direct measurements of its space-dependent components. The proposed technique was demonstrated on the Bruker Biospec 94/20USR system, using the spin echo diffusion sequence to image an isotropic water phantom and an anisotropic capillary phantom. The accuracy of the diffusion tensor determination was improved by an overall factor of about 8 for the isotropic water phantom.

1. Introduction

Diffusion tensor imaging (DTI) of water using nuclear magnetic resonance (NMR) methods has developed for nearly two decades. In principle, it is a version of the magnetic resonance imaging (MRI), in which additional magnetic field gradients are applied. They cause an extra dephasing of precessing magnetization that depends on translational mobility of water. A pioneering work of P. Basser et al. [1,2] has began the era of DTI applications to biological systems, allowing to study the diffusion of water in a quantitative way. Comparing to the unrestricted diffusion in an isotropic medium, the measurements in biological tissues provide important structural and diagnostic information, which is not accessible by any other method. This is particularly evident in studies of biological systems exhibiting by nature strong structural anisotropy, such as spinal cord [3–11] or brain [12–15]. The initial diffusion measurements performed *in vitro* [3–7], were consecutively extended to *in vivo* studies [8–15]. A relationship was found between changes in parameters characterizing diffusion and the changes caused by injury or disease in the spinal cord or brain. Another interesting and promising application of DTI is tractography [16–30], which provides a way to visualize the orientation of nerve fibers in the brain and spinal cord. The quantitative data obtained by DTI can also complement the information obtained by standard functional MRI [31–33].

These applications stimulated a continuous progress of the DTI methodology [34–51]. The pulse sequences were optimized by proper selection of amplitudes and directions of diffusion gradient vectors [34,39,41–48,52,53]. Appropriate quantitative parameters derived from the measurements were selected to describe the condition of the studied tissue [12,52–74]. A significant effort was put to eliminate the adverse effects of imaging gradients, their coupling with the diffusion gradients [12,47,75–78], and to reduce the eddy current effects [79].

Due to anisotropic nature of the biological tissue, the diffusion must be often described by the tensor. It is a 3 by 3 symmetric matrix, characterized by six independent parameters, which can be visualized as an ellipsoid [59]. The eigenvectors and eigenvalues of the matrix correspond to the orientation of the ellipsoid in space and the squared amplitudes of its hemiaxes, respectively. The ultimate goal of the DTI experiment is to precisely determine these parameters for every individual voxel of the MRI image. This requires that, apart from the reference image taken without any diffusion gradients, at least six images are measured by applying at least six linearly independent directions of diffusion gradients. Assuming that the shapes, amplitudes and directions of diffusion gradients are known precisely, the components of the diffusion tensor can be determined and then it can be diagonalized, providing the required information.

☆ The National Centre for Research and Development for grant PBS2/A2/16/2013.
* Corresponding author at: AGH University of Science and Technology, al. A. Mickiewicza 30, 30-059 Kraków, Poland. Tel. +48 502721555.
E-mail address: Artur.Krzyzak@clarus.com.pl (A.T. Krzyżak).

http://dx.doi.org/10.1016/j.mri.2014.10.007

The parameters characterizing the diffusion gradients for a given imaging sequence and for a given diffusion gradient direction are incorporated in so called b-matrix. In commercial MRI systems the b-matrices are provided. As such, they usually do not take into account the imaging gradient effects, their coupling with the diffusion gradients, or the eddy current effects. Moreover, the provided values of b matrix are constant over the entire volume. It should be pointed out that the above effects lead to systematic errors, so they cannot be reduced by signal averaging. Although some ingenious techniques have been devised to reduce these effects, they are either measurement-time consuming, or not feasible in some circumstances. This leads to limited accuracy of obtained data, which in view of growing applications of the DTI, is a serious limitation of the technique.

In this paper we propose an alternative method of determining the b-matrix by using an anisotropic phantom of known spatial distribution of the diffusion tensor [82]. It almost completely eliminates the need of knowing the parameters of the imaging sequence. Moreover, the method provides detailed information on spatial dependence of the b-matrix, voxel-by-voxel, in the entire ROI. Recently the method has been named BSD-DTI (B-matrix spatial distribution in DTI) [83], and this term will be used in the following.

We provide a theoretical basis for this method, and discuss its advantages and limitations. As a preliminary evidence of the superiority of the BSD-DTI over the standard procedure, an improved accuracy of the diffusion tensor determination for the homogeneous water phantom and the anisotropic capillary phantom is demonstrated.

1.1. Theory

The NMR signal attenuation due to diffusion in the pulse gradient spin echo (PGSE) sequence is described by the Stejskal–Tanner equation [80,81]:

$$\ln\left(\frac{S(\mathbf{b})}{S(\mathbf{bo})}\right) = -(\mathbf{b}\text{-}\mathbf{bo}) : \mathbf{D} = -\sum_{i,j=1}^{3}\left(b_{ij} - bo_{ij}\right)D_{ij} \tag{1}$$

where S(**b**), S(**bo**) are the signal intensities with and without diffusion-sensitizing gradients, respectively, b_{ij}—components of the diffusion gradient matrix **b**, D_{ij}—components of the diffusion tensor **D**, and the colon designates the generalized dot product.

In the generic PGSE imaging sequence, a particular time-dependent diffusion gradient vector can be expressed in the Cartesian system of coordinates, $G(t) = [G_1(t), G_2(t), G_3(t)]^T$, which can be associated with the x, y, z axes of the gradient system implemented in the MR scanner. These axes can be also related to phase, frequency, and slice encoding directions of the MRI sequence. The overall time dependence of the b matrix can be concisely expressed in terms of the vector G in the following way [59]:

$$\mathbf{b}(t) = \int_0^t \mathbf{k}(t')\mathbf{k}^{\mathbf{T}}(t')dt', \tag{2}$$

where

$$\mathbf{k}(t') = \gamma\int_0^{t'}\mathbf{G}(t'')dt'' - 2H(t'-\tau)\mathbf{k}(\tau), \tag{3}$$

$$\mathbf{k}(\tau) = \gamma\int_0^{\tau}\mathbf{G}(t'')dt'', \tag{4}$$

and $H(t)$—the Heaviside unit step function, TE—echo time, $\tau = TE/2$.

The symmetric 3×3 **b(t)** matrix is calculated for the maximum of signal intensity corresponding to $t = TE$ and contains six components: b_{xx}, b_{yy}, b_{zz}, b_{xy}, b_{xz}, b_{yz}, for each orientation of the diffusion gradient vector. In order to accurately calculate the b matrix components, we need to know the real amplitudes, shapes, and time dependencies of all magnetic field gradients that are present during the DTI sequence.

The b matrix can be split into several parts corresponding to various sources of magnetic field gradients. Typically, the largest contribution $\mathbf{b_d}$ comes from diffusion gradients. The second largest part $\mathbf{b_i}$ is generated by the imaging gradients. Much smaller components originate from the background gradients, $\mathbf{b_g}$, and from $\mathbf{b_n}$, which combines the contributions from eddy currents, nonlinear parts of imaging gradients, radiation damping, and other sources of distortion associated with the background gradients. And finally, in addition to these first order terms, there are also cross-terms due to interference between various gradients, which are proportional to their products. Of these, typically the largest contribution comes from the pairs of diffusion, and imaging gradients $\mathbf{b_{di}}$; $\mathbf{b_{dg}}$, $\mathbf{b_{dn}}$, and $\mathbf{b_{ig}}$, $\mathbf{b_{in}}$, $\mathbf{b_{gn}}$, are usually much smaller, but all may be space dependent.

The importance of cross terms has been pointed out in the literature [75–79]. The proposed methods of their elimination or reduction rely on optimization of the imaging sequence, by shortening the echo time TE [39], or increasing the diffusion gradients amplitudes.

The above discussion shows how difficult, if not impossible is to determine the b matrix using the formula [2] with satisfactory precision. There are some simple mathematical manipulations which can reduce the influence of spurious gradients to some extent [77–79]. Two of them are described below.

1. Dividing the signal S(**b**) by S(**bo**) eliminates the impact of $\mathbf{b_i}$ and other possible components of **b**, which are the same for both MR sequences, like $\mathbf{b_g}$ and $\mathbf{b_n}$. However, this may not be true, especially in microimaging, where strong diffusion gradients generate extra eddy currents. It follows from Eq. (1) that:

$$\begin{gathered}-(\mathbf{b}-\mathbf{bo})\mathbf{D} = \\ -\left(\left(\mathbf{b_i}+\mathbf{b_g}+\mathbf{b_n}+\mathbf{b_d}+\mathbf{b_{di}}+\mathbf{b_{dg}}+\mathbf{b_{dn}}+\mathbf{b_{ig}}+\mathbf{b_{in}}+\mathbf{b_{gn}}\right)-\left(\mathbf{b_i}+\mathbf{b_g}+\mathbf{b_n}+\mathbf{b_{ig}}+\mathbf{b_{in}}+\mathbf{b_{gn}}\right)\right):\mathbf{D} = \\ -\left(\mathbf{b_d}+\mathbf{b_{di}}+\mathbf{b_{dg}}+\mathbf{b_{dn}}\right):\mathbf{D}.\end{gathered} \tag{5}$$

   For the convenience we introduce the substitution:

$$\mathbf{b_d} + \mathbf{b_{di}} + \mathbf{b_{dg}} + \mathbf{b_{dn}} = \mathbf{b'}. \tag{5'}$$

2. Multiplying the signals S(**b**) and S^-(**b**) (signal intensity with the opposite diffusion gradients) eliminates the cross-terms associated with diffusion gradients. However, this procedure requires twice as much measurement time:

$$\begin{gathered}-\left(\left(\mathbf{b_i}+\mathbf{b_g}+\mathbf{b_n}+\mathbf{b_d}+\mathbf{b_{di}}+\mathbf{b_{dg}}+\mathbf{b_{dn}}+\mathbf{b_{ig}}+\mathbf{b_{in}}+\mathbf{b_{gn}}\right)\right. \\ \left.+\left(\mathbf{b_i}+\mathbf{b_g}+\mathbf{b_n}+\mathbf{b_d}-\mathbf{b_{di}}-\mathbf{b_{dg}}-\mathbf{b_{dn}}+\mathbf{b_{ig}}+\mathbf{b_{in}}+\mathbf{b_{gn}}\right)\right): \\ \mathbf{D} = -2\left(\mathbf{b_d}+\mathbf{b_i}+\mathbf{b_g}+\mathbf{b_n}+\mathbf{b_{ig}}+\mathbf{b_{in}}+\mathbf{b_{gn}}\right):\mathbf{D}.\end{gathered} \tag{6}$$

   In subsequent discussion we will use the substitution:

$$\mathbf{b_d} + \mathbf{b_i} + \mathbf{b_g} + \mathbf{b_n} + \mathbf{b_{ig}} + \mathbf{b_{in}} + \mathbf{b_{gn}} = \mathbf{b''}. \tag{6'}$$

   Combining both methods we obtain the dependence on the diffusion gradients only:

$$\mathrm{Ln}\left((S(\mathbf{b})S^-(\mathbf{b}))^{1/2}/S(\mathbf{bo})\right) = -(\mathbf{b''}-\mathbf{bo}) : \mathbf{D} = -\mathbf{b_d} : \mathbf{D}. \tag{7}$$

In the next section, a new method is described for determining the b matrix and its space dependence that does not require any knowledge of shapes, amplitudes and time dependencies of diffusion