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Dynamic Correction of Artifacts due to Susceptibility Effects and Time-Varying Eddy Currents in Diffusion Tensor Imaging

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Abstract

In diffusion tensor imaging (DTI), spatial and temporal variations of the static magnetic field (B_0) caused by susceptibility effects and time-varying eddy currents result in severe distortions, blurring, and misregistration artifacts, which in turn lead to errors in DTI metrics and in fiber tractography. Various correction methods have been proposed, but typically assume that the eddy current-induced magnetic field can be modeled as a constant or a single exponential decay within the DTI readout window. Here, we show that its temporal dependence is more complex because of the interaction of multiple eddy currents with different time constants, but that it remains very consistent over time. As such, we propose a novel dynamic B_0 mapping and off-resonance correction method that measures the exact spatial, temporal, and diffusion-weighting direction dependence of the susceptibility- and eddy current-induced magnetic fields to effectively and efficiently correct for artifacts caused by both susceptibility effects and time-varying eddy currents, thereby resulting in a high spatial fidelity and accuracy.

Keywords

dynamic; artifact correction; susceptibility; eddy currents; diffusion tensor imaging; magnetic field mapping

1. Introduction

Diffusion tensor imaging (DTI) (Basser et al., 1994) has become a valuable neuroimaging technique for assessing white matter connectivity and integrity noninvasively in both basic neuroscience research (Mori and Zhang, 2006) and clinical applications, such as the investigation of stroke, multiple sclerosis, epilepsy, and neurodegenerative diseases (Ciccarelli et al., 2008). It is, however, typically performed with fast imaging sequences, such as echo-planar imaging (EPI), which are vulnerable to spatial and temporal variations of the static magnetic field (B_0) due to susceptibility differences at air/tissue interfaces, $B_0^{\text{susc}}(\mathbf{x})$, and eddy currents induced by the strong diffusion-weighting (DW) gradients, $B_0^{\text{eddy}}(\mathbf{x}, t, d)$. These off-resonance effects vary with space ($\mathbf{x}$), time (t), and/or DW direction (d), resulting in severe geometric distortions, image blurring, and misregistration

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among different diffusion-weighted images, respectively. All of these artifacts in turn lead to errors in the derivation of the diffusion tensor and consequently in DTI metrics, such as mean diffusivity or fractional anisotropy (FA) maps, as well as in fiber tractography.

A number of correction methods have been developed to address these issues, involving the acquisition of B_0 maps with the same DW gradients as the DTI scan (Chen et al., 2006; Truong et al., 2008), the acquisition of two DTI datasets with reversed DW directions (Bodammer et al., 2004) or reversed phase-encoding directions (Embleton et al., 2010; Gallichan et al., 2010), and the coregistration of DTI images with various post-processing algorithms (Rohde et al., 2004; Ardekani and Sinha, 2005; Zhuang et al., 2006). However, reversed gradient methods typically require a twice longer scan time, which is particularly impractical in pediatric or patient populations, whereas post-processing methods require the coregistration of images with highly variable contrasts. Furthermore, all of these methods assume that B_0^{eddy} remains constant within the DTI readout window and can therefore not correct for artifacts (e.g., blurring) caused by time-varying eddy currents. Such methods include the affine registration algorithms available in major post-processing packages such as FSL (<http://www.fmrib.ox.ac.uk/fsl>) or AIR (<http://bishopw.loni.ucla.edu/AIR5>), which are used by most DTI end users to correct for eddy currents.

In reality, B_0^{eddy} varies over time and is generally modeled as a sum of exponential decays with different time constants, ranging from a few microseconds to hundreds of seconds (Bernstein et al., 2004). The twice refocused spin-echo method (Reese et al., 2003) further assumes that it can be characterized by a single exponential decay, which is known not to be the case (Frank et al., 2010). As such, this method can only compensate for the dominant eddy currents, moreover at the cost of a longer echo time (TE) and hence lower signal-to-noise ratio (SNR), which is particularly undesirable in DTI. A modified version that can compensate for eddy currents with two different time constants has been proposed (Finsterbusch, 2010), but requires an even longer TE. Finally, one reversed gradient method (Shen et al., 2004) may be able to correct for the higher-order distortions caused by the time-varying component of B_0^{eddy} , but only provided that they are small in comparison to the linear distortions caused by the static component of B_0^{eddy} , which is not necessarily true, as will be shown in the Results section.

In this study, we show that B_0^{eddy} varies substantially within the DTI readout window and cannot be accurately modeled as a constant or a single exponential decay. We thus propose a novel dynamic B_0 mapping and off-resonance correction method that measures the exact spatial, temporal, and DW direction dependence of B_0^{susc} and B_0^{eddy} to effectively and efficiently correct for artifacts caused by both susceptibility effects and time-varying eddy currents.

2. Methods

2.1 Dynamic B_0^{eddy} mapping

The first step consists in acquiring a series of dynamic B_0^{eddy} maps for a range of time points spanning the DTI readout window and for each DW direction. To this end, a train of diffusion-weighted asymmetric spin-echo images are acquired with a multiecho spin-echo pulse sequence at different time points $\tau_m = \tau_1, \dots, \tau_M$ spanning the readout duration T_{acq} of the DTI scan (Fig. 1). The resulting phase images are then unwrapped along the time dimension and, for each time point τ_m and DW direction d , fitted with:

$$\phi(\mathbf{x}, t, d) = \phi_0(\mathbf{x}, \tau_m, d) + \gamma B_0^{\text{eddy}}(\mathbf{x}, \tau_m, d) t \quad [1]$$

where ϕ_0 is a constant, γ is the gyromagnetic ratio, and t is a 5-point moving window $\{\tau_{m-4}, \tau_{m-2}, \tau_m, \tau_{m+2}, \tau_{m+4}\}$, to derive a time series of B_0^{eddy} maps. The phase unwrapping and fitting are performed separately on odd and even echoes, which are acquired with readout gradients of opposite polarity, to avoid errors due to off-resonance effects (as shown in different colors in Fig. 1B). Although dynamic B_0^{eddy} maps could also be generated from successive pairs of phase images, we found that a 5-point moving window provides a higher SNR with no significant loss in temporal resolution, because B_0^{eddy} varies relatively slowly with respect to the echo spacing.

Since eddy currents depend on the DW gradients, but not on the subject being imaged, this dynamic B_0^{eddy} mapping is performed with the same DW gradients as the DTI scan, but only once on a phantom. The resulting B_0^{eddy} maps can then be repeatedly used to correct for artifacts caused by time-varying eddy currents in all previous and subsequent DTI scans acquired with the same protocol, without requiring any additional scan time. Furthermore, since B_0^{eddy} varies slowly in space, the B_0^{eddy} mapping can be performed at a lower spatial resolution than that of the DTI scan, which can be traded for a higher SNR and a higher temporal resolution (because of the reduced echo spacing) to accurately measure the temporal dependence of B_0^{eddy} within T_{acq} (Fig. 1).

To remove any common B_0 inhomogeneities not caused by the eddy currents, such as those due to imperfect shimming, non-diffusion-weighted B_0 maps of the phantom are also acquired and are subtracted from the diffusion-weighted B_0 maps for all time points and DW directions. To improve the SNR, the resulting B_0^{eddy} maps are then fitted with a third-order polynomial function in space within the phantom, which was previously shown to be an adequate model (Truong et al., 2008), and extrapolated outside the phantom. However, in contrast to existing eddy current correction methods, the proposed method does not make any assumptions regarding the temporal dependence of B_0^{eddy} , so that it can effectively correct for artifacts caused by any time-varying eddy currents.

2.2 Static B_0^{susc} mapping

The second step consists in acquiring a static B_0^{susc} map by using the same pulse sequence and post-processing method as for the dynamic B_0^{eddy} mapping, but by simultaneously fitting all odd or even echoes with:

$$\phi(\mathbf{x}, t) = \phi_0(\mathbf{x}) + \gamma B_0^{\text{susc}}(\mathbf{x}) t \quad [2]$$

(instead of using a moving window) and by averaging the two resulting B_0^{susc} maps. In contrast to eddy currents, since susceptibility effects depend on the subject being imaged, but not on the DW gradients, this B_0^{susc} mapping is performed *in vivo*, but without diffusion-weighting. In addition, since B_0^{susc} can vary rapidly in space, but does not vary in time, the B_0^{susc} mapping is performed at the same spatial resolution as that of the DTI scan, but with fewer echoes than the dynamic B_0^{eddy} mapping.

2.3 Dynamic off-resonance correction

The third step consists in using the static B_0^{susc} map and dynamic B_0^{eddy} maps to correct for distortions and blurring artifacts in the DTI images caused by both susceptibility effects and time-varying eddy currents (Fig. 2). Specifically, for each time point $t_n = t_1, \dots, t_N$ corresponding to the acquisition of a k_y line in k-space (where y denotes the phase-encoding direction) and for each DW direction d , the uncorrected DTI image is multiplied by $\exp[-i\phi(\mathbf{x}, t_n, d)]$, where

$$\phi(\mathbf{x}, t_n, d) = \gamma \int_{TE}^{t_n} [B_0^{\text{susc}}(\mathbf{x}) + B_0^{\text{eddy}}(\mathbf{x}, t, d)] dt. \quad [3]$$

Each of these N images is Fourier transformed to k-space and the n^{th} k_y line (acquired at time t_n) is extracted from the n^{th} k-space to form a new k-space, which is inverse Fourier transformed to yield the corrected image. If partial Fourier imaging is used, the partial Fourier reconstruction is performed at the last step.

Note that the acquired B_0^{susc} and B_0^{eddy} maps require further processing before they can be used in Eq. [3]. First, the B_0^{eddy} maps are interpolated in space and downsampled in time (from $\{\tau_m = \tau_1, \dots, \tau_M\}$ to $\{t_n = t_1, \dots, t_N\}$) to match the spatial resolution and echo spacing of the DTI scan. Second, the B_0^{susc} and B_0^{eddy} maps, which are initially in undistorted coordinates, need to be transformed into distorted coordinates, so that they are coregistered with the uncorrected DTI images. In practice, we found it sufficient to perform this transformation only on the B_0^{susc} map, because the B_0^{eddy} maps vary slowly in space so that the difference is negligible. To this end, each undistorted asymmetric spin-echo image acquired for the B_0^{susc} mapping is distorted by using the same procedure as above, except that Eq. [3] is replaced by

$$\phi(\mathbf{x}, t_n) = -\gamma B_0^{\text{susc}}(\mathbf{x})(t_n - TE) \quad [4]$$

where $B_0^{\text{susc}}(\mathbf{x})$ is the undistorted B_0^{susc} map. The resulting distorted asymmetric spin-echo images are then used to generate a distorted B_0^{susc} map as described in section 2.2.

2.4 Experiments

We studied three healthy volunteers, who provided written informed consent as approved by our Institutional Review Board, on a 3 T Excite MRI scanner (GE Healthcare, Milwaukee, WI) equipped with an eight-channel phased-array head coil and a gradient system with 40 mT/m maximum amplitude and 150 T/m/s slew rate. High-order shimming was applied to minimize the global B_0 inhomogeneity. Axial DTI images of the brain were acquired with a single-shot spin-echo EPI pulse sequence and the following parameters: repetition time (TR) = 5 s, TE = 73 ms (minimum), field-of-view = 24×24 cm, matrix size = 96×96 (interpolated to 128×128 by zero padding), slice thickness = 2.5 mm, number of slices = 20, frequency-encoding direction = right/left, partial Fourier encoding = 5/8, echo spacing = 944 μ s, and $T_{\text{acq}} = 56$ ms.

Static B_0^{susc} mapping was performed with TR = 2 s, TE = 32 ms, and number of echoes = 24, whereas dynamic B_0^{eddy} mapping was performed with TR = 2.5 s, TE = 55 ms, matrix size = 48×48, slice thickness = 5 mm, number of slices = 10, number of echoes = 96, and echo spacing = 624 μ s on a 20-cm diameter spherical gel phantom. For the DTI scan and B_0^{eddy} mapping, DW gradients were applied with the following parameters: amplitude = 39.4 mT/m, duration (δ) = 22 ms, separation (Δ) = 26.7 ms, b -factor = 1000 s/mm², and 15 DW directions. To assess the temporal stability of the eddy currents, the B_0^{eddy} mapping was repeated four times over a period of six months. For anatomical reference, high-resolution T_2 -weighted images of the brain were also acquired with a fast spin-echo pulse sequence and TR = 3 s, TE = 86 ms, matrix size = 256×256, and slice thickness = 2.5 mm.

For comparison with the proposed dynamic off-resonance correction, a static off-resonance correction was also performed by using the same $B_0^{\text{susc}}(\mathbf{x})$ map, but time-averaged $B_0^{\text{eddy}}(\mathbf{x}, d)$ maps instead of the dynamic $B_0^{\text{eddy}}(\mathbf{x}, t, d)$ maps. For each of the three DTI datasets obtained with no correction, static correction, and dynamic correction, FA maps were

computed and color-coded according to the direction of the first eigenvector. In addition, maps of the total FA error were computed as $(FA_{\text{uncorrected}} - FA_{\text{dynamic}})/FA_{\text{dynamic}}$ and maps of the residual FA error after static correction were computed as $(FA_{\text{static}} - FA_{\text{dynamic}})/FA_{\text{dynamic}}$, where $FA_{\text{uncorrected}}$, FA_{static} , and FA_{dynamic} are the FA values with no correction, static correction, and dynamic correction, respectively. All image reconstruction and analysis were performed in Matlab (The MathWorks, Natick, MA).

3. Results and Discussion

Representative B_0^{eddy} maps at different time points within the DTI readout window (Fig. 3A) as well as dynamic B_0^{eddy} time courses in a given voxel (Fig. 3B) for different DW directions show that B_0^{eddy} varies substantially with space, time, and DW direction. In particular, these results demonstrate that B_0^{eddy} cannot simply be modeled as a constant within the DTI readout window, as assumed in the vast majority of existing eddy current correction methods. Furthermore, the zero-crossing clearly shows that it cannot be accurately characterized by a single exponential decay either, as assumed in the twice-refocused spin-echo method, because of the interaction of multiple eddy currents with different amplitudes and time constants. The range of variation of B_0^{eddy} within the DTI readout window is 2 to 3 orders of magnitude larger than its temporal average for all DW directions.

A representative uncorrected diffusion-weighted image shows severe distortions and blurring artifacts caused by susceptibility effects and time-varying eddy currents, particularly near the anterior horns of the lateral ventricles and in the frontal lobes (Fig. 4A, arrows). These artifacts are only partially reduced with the static off-resonance correction (Fig. 4B), whereas they are completely eliminated with the proposed dynamic off-resonance correction (Fig. 4C).

Similarly, uncorrected FA maps in three representative slices show severe susceptibility-induced distortions, particularly near the genu of the corpus callosum (Fig. 5A, dashed lines), as well as eddy current-induced FA errors, most prominently at the anterior and posterior edges of the brain (arrows). The static off-resonance correction can only correct for the susceptibility-induced distortions as well as the eddy current-induced FA errors at the posterior edge of the brain, but not at the anterior edge (Fig. 5B, arrows). In contrast, the proposed dynamic off-resonance correction can effectively correct for all artifacts (Fig. 5C).

Although the susceptibility- and eddy current-induced artifacts may initially appear fairly localized, a representative map of the total FA error reveals that they are actually extensive throughout the whole brain (Fig. 6A), with $(75.9 \pm 1.4)\%$ of the voxels in the brain having an absolute FA error larger than 10% (Fig. 6C, solid line). Similarly, a map of the residual FA error shows that even after static off-resonance correction, residual eddy current-induced artifacts are still widespread (Fig. 6B), with $(26.2 \pm 1.9)\%$ of the voxels in the brain having an absolute FA error larger than 10% (Fig. 6C, dashed line).

Dynamic B_0^{eddy} mapping repeated four times over a period of six months yielded virtually identical B_0^{eddy} maps (Fig. 3B, error bars), whereas dynamic off-resonance correction of the same DTI data with these four sets of B_0^{eddy} maps resulted in virtually identical FA maps (e.g., $FA = 0.744 \pm 0.004$ in the genu of the corpus callosum). These results demonstrate that the eddy currents induced by the DW gradients remain very stable, even over a long time period, so that the B_0^{eddy} mapping can be performed very infrequently (i.e., no more than once or twice a year).

Furthermore, although in this study the B_0^{eddy} maps were acquired with the same DW gradients as the DTI scan, it is in principle possible to use linear combinations of these

B_0^{eddy} maps for the correction of DTI data acquired with other DW gradient amplitudes and directions as well, provided that the eddy currents induced by gradients applied along the x , y , and z axes can be modeled as a linear system (Zhuang et al., 2006), which can be verified in future studies.

Since the focus of this study was the correction of artifacts caused by time-varying eddy currents, the static B_0^{susc} mapping was performed with the same multiecho spin-echo sequence as the dynamic B_0^{eddy} mapping (but without diffusion-weighting) for convenience. However, more efficient B_0^{susc} mapping methods, such as the acquisition of two asymmetric spin-echo EPI images at different TEs, can also be used to further reduce the scan time.

In contrast to existing eddy current correction methods, the proposed method can effectively correct for artifacts caused by any time-varying eddy currents. Furthermore, it does not require any additional scan time as compared to static B_0 mapping methods, is much more efficient than reversed gradient methods, and does not suffer from the SNR penalty of twice-refocused spin-echo methods. In this study, we used a single-shot EPI pulse sequence because it is the most commonly used for DTI. However, the proposed method is also compatible with multishot acquisitions, other imaging sequences (e.g., spiral or radial imaging), as well as parallel imaging techniques.

This method will be beneficial in longitudinal studies of the same subjects, because different head positions with respect to the B_0 field or the gradient axes will lead to different susceptibility- and eddy current-induced artifacts, resulting in different DTI metrics, if not corrected for. It will also be particularly beneficial in multicenter studies, which may use different scanner models and/or manufacturers with potentially very different eddy currents, even though the scan parameters are the same. Finally, it will benefit both basic neuroscience research and clinical applications such as presurgical planning, for which a high spatial fidelity is especially critical.

4. Conclusions

The results of this study demonstrate that B_0^{eddy} remains very consistent over time, but varies substantially within the DTI readout window and cannot be accurately modeled as a constant or a single exponential decay, as assumed in nearly all existing eddy current correction methods. The proposed dynamic B_0 mapping and off-resonance correction method can measure the exact spatial, temporal, and DW direction dependence of B_0^{susc} and B_0^{eddy} to effectively and efficiently correct for the severe distortions, blurring, and misregistration artifacts caused by susceptibility effects and time-varying eddy currents, thereby leading to a high spatial fidelity and accuracy in the resulting DTI metrics.

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4. Highlights

- Susceptibility effects and time-varying eddy currents cause severe artifacts in DTI
- Eddy current-induced magnetic fields vary substantially within the readout window
- They cannot be modeled as a constant or a monoexponential decay, as usually assumed
- A novel dynamic off-resonance correction method is proposed to address these issues
- This method can effectively and efficiently correct for both types of artifacts

Fig. 1. DTI (A) and dynamic B_0^{eddy} mapping (B) pulse sequence diagrams (RF: radiofrequency excitation; G_z , G_y , G_x : slice-select, phase-encoding, and readout gradients; acq: data acquisition). For clarity, only a limited number of echoes are shown in both (A) and (B). In reality, the width of the moving window is much smaller in comparison to the DTI readout duration T_{acq} . Odd and even echoes are shown in different colors in (B).

Fig. 2. Schematic diagram of the dynamic off-resonance correction method (FT: Fourier transform, PF: partial Fourier reconstruction (optional), FT^{-1} : inverse Fourier transform).

Fig. 3.
 (A) Representative B_0^{eddy} maps at different time points within the DTI readout window for 4 out of 15 DW directions. (B) Dynamic B_0^{eddy} time courses in the voxel marked as + in (A) for all DW directions (mean $\pm$ standard error of the mean for four repeated measurements performed over six months).

Fig. 4. Representative diffusion-weighted images with no correction (A), static correction (B), and dynamic correction (C). Contour lines generated from an undistorted fast spin-echo image are overlaid on one hemisphere. The arrows show severe distortions and blurring artifacts caused by susceptibility effects and time-varying eddy currents.

Fig. 5. Color-coded FA maps in three representative slices with no correction (A), static correction (B), and dynamic correction (C) (red: right-left, green: anterior-posterior, blue: superior-inferior). The dashed lines and arrows highlight regions with severe susceptibility-induced distortions and eddy current-induced FA errors, respectively. The middle slice is the same as the one shown in Fig. 4.

Fig. 6. Maps of the total FA error (A) and residual FA error after static off-resonance correction (B) in the same slice as the middle slice of Fig. 5. (C) Percentage of voxels in the brain having an absolute FA error exceeding a given value (mean $\pm$ standard error of the mean across subjects).