Correction of Vibration Artifacts in DTI Using Phase-Encoding Reversal (COVIPER)

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Diffusion tensor imaging is widely used in research and clinical applications, but still suffers from substantial artifacts. Here, we focus on vibrations induced by strong diffusion gradients in diffusion tensor imaging, causing an echo shift in k-space and consequential signal-loss. We refined the model of vibration-induced echo shifts, showing that asymmetric k-space coverage in widely used Partial Fourier acquisitions results in locally differing signal loss in images acquired with reversed phase encoding direction (blip-up/blip-down). We implemented a correction of vibration artifacts in diffusion tensor imaging using phase-encoding reversal (COVIPER) by combining blip-up and blip-down images, each weighted by a function of its local tensor-fit error. COVIPER was validated against low vibration reference data, resulting in an error reduction of about 72% in fractional anisotropy maps. COVIPER can be combined with other corrections based on phase encoding reversal, providing a comprehensive correction for eddy currents, susceptibility-related distortions and vibration artifact reduction. Magn Reson Med 68:882–889, 2012. ©2011 Wiley Periodicals, Inc.

Key words: partial Fourier imaging; table vibration; blip-up blip-down; k-space echo shift

Diffusion tensor imaging (DTI) allows for noninvasive imaging of water diffusion (1–4), an important marker for brain anatomy and physiology. In addition to its wide spread use in neuroscience research (5–9), DTI has become an essential diagnostic tool in stroke and neurodegenerative disease due to its sensitivity to microstructural and physiological changes (10–16).

However, obtaining reliable DTI data and drawing meaningful inference from them are more challenging than for conventional imaging modalities such as T1- or T2-weighted images, because in diffusion MRI the contrast depends on the molecular water diffusion at the micrometer scale (17–20). To achieve the required diffusion sensitization strong imaging gradients need to be applied, which usually entails a high first-order gradient moment. The large first moment makes the DTI sequence sensitive to brain-tissue movements [e.g., due to human physiology (21,22) or instrumental vibration (23,24)], resulting in a substantial echo shift in k-space locally where strong movement occurs. If this echo shift exceeds the k-space acquisition window, severe signal loss occurs.

Instrumental vibration is a recently discovered source (23) of movement-related signal-loss in DTI affecting multiple research sites and different types of scanners (20,25–28). Strong diffusion gradients excite low-frequency mechanical resonances of the MR system (29). This leads to a patient table vibration, which can be transferred to the subject’s brain (24). Movement-related signal-loss in diffusion-weighted (DW) images can reduce the diagnostic value of DW imaging (23), bias the estimation of the diffusion tensor (21,23,25,27,30), reduce the comparability of different quantitative DTI indices and hinder emerging diffusion techniques, which increasingly rely on high diffusion weighting (31–33).

Retrospective correction methods have been established for well-known artifacts such as eddy currents (e.g., Refs. 34 and 35) and susceptibility-related distortion (e.g., Refs. 36 and 37). In contrast, vibration artifacts in DTI have been explored and reported in the literature only recently (20,23,25,27,28). The goal of this study was to retrospectively correct vibration-induced signal loss motivated by a similar well established approach for reducing susceptibility-induced signal loss due to echo shifts in gradient echo echo-planar imaging (38), which is based on the concept of acquiring two data sets with opposite phase encoding (PE) direction (39,40). Although the acquisition and combination of data with reversed PE direction has already been established for DTI to correct for susceptibility and eddy current induced distortion (37,41–44), it needs to be modified for reduction of vibration artifacts.

We have developed a simple method for correction of vibration artifacts in DTI using phase-encoding reversal (COVIPER) by combining two images with reversed PE direction, each weighted by a function of its local tensor-fit error. To test the proposed method, we compared the fractional anisotropy (FA) and tensor-fit error of COVIPER corrected images with images combined using the conventional arithmetic mean (37,41), and with reference images that contained negligible vibration artifacts.

MATERIALS AND METHODS

Theory

Motion during the diffusion-weighting period may shift the echo center towards the edge of k-space in PE direction (Fig. 1) and lead to loss of the DW signal, as has been demonstrated for linear rigid-body motion (45). For linear rigid-body motion the shift of the echo center ∆k
only depends on rotational but not on the translational movement (45,46):
\[
\Delta \mathbf{k} = \gamma m_1 \mathbf{G} \times \mathbf{\Omega} \tag{1}
\]
where \( \gamma \) is the gyromagnetic ratio, \( \mathbf{\Omega} \) is the angular velocity vector, \( \mathbf{G} \) a dimensionless unit vector in the diffusion gradient direction and \( m_1 \) is the first moment of the gradient. Note that the first order gradient moment depends on the gradient waveform and thus might vary between diffusion sequences (see e.g., Ref. 47).

Usually movement-induced signal-dropout is caused by shifting the echo center in PE direction (e.g., Refs. 23 and 45), because the \( k \)-space is often only partially covered in Partial Fourier imaging. Previous studies suggested that mechanical vibration of the scanner primarily causes nonrigid rotational movement of the brain tissue in the transverse plane, which is relatively slow, varying across the brain (24). Motivated by those measurements, we assumed that the spatial derivative of the rotational in-plane movement could be neglected. Giving this approximation and assuming that the PE direction is applied along the anterior-posterior (y) direction the shift in the \( k_y \)-direction can be described as follows:

\[
\Delta k_y(r) \propto -\hat{G}_x \Omega_{y}^{\text{eff}}(r) \tag{2}
\]
where \( \hat{G}_x \) is the \( x \) component of the diffusion gradient vector and \( \Omega_{y}^{\text{eff}}(r) \) the \( z \) component of the effective angular velocity vector at voxel position \( r \) due to local rotational in-plane movement (Fig. 1a). Note that \( \Omega_{y}^{\text{eff}}(r) \) in Eq. 2 is time-averaged, i.e., it has been integrated over the diffusion weighting time.

According to Eq. 2, the local effective angular velocity might be strong enough to shift the echo center out of the \( k \)-space acquisition window (Fig. 1c). Since the central echo carries the main signal power, severe signal dropout will occur in this case (38,48). The acquisition window can be shifted by reversing the PE direction as shown in Fig. 1c,d, to capture the shifted main echo. If two images with reversed PE direction (subsequently denoted as blip-up and blip-down data set) are acquired, the signal lost in one of these images may be recovered in the other if the shift \( \Delta k_y(r) \) does not exceed the nominal \( k \)-space window of both images, i.e.

\[
|\Delta k_y(r)| < \max(|k_{\text{min}}, k_{\text{max}}|) \tag{3}
\]
and (b) but using a longer TR (slice TR = 10.2 s, DTI2\textsuperscript{+} and DTI2\textsuperscript{-}). Due to the short TR, datasets (a) and (b) were prone to vibration artifacts. Increasing the TR in datasets (c) and (d) made the data less affected by vibration artifacts, because the damped low-frequency mechanical vibration of the patient table induced by the diffusion gradients fade with increasing TR (see Refs. 23 and 29). As a result, the measurements (c) and (d) were used as the reference dataset.

**Data Preprocessing**

All analysis steps were performed using SPM8 [http://www.fil.ion.ucl.ac.uk/spm; (55)] and in-house software written in MATLAB (version 7.11.0; Mathworks, Natick, MA). First, all four DTI datasets were corrected for motion and eddy current effects (35). Next, the dual gradient echo FLASH data were processed to create a voxel displacement map (50). Each of the four DTI datasets was corrected for susceptibility-induced geometric distortion using this displacement map (23,50). Finally, to reduce residual misalignments between the flip-up and flip-down datasets, the first non-DW image of each flip-down DTI dataset was registered to the corresponding image of the flip-up DTI dataset and the resulting 12-parameter affine transformation was applied to the other images of the flip-down dataset.

**COVIPER Method**

The COVIPER method is a two-step procedure. In the first step, the diffusion tensor (D) as well as the residual error of the tensor fit (e) were calculated for all four DTI datasets of each subject using the standard diffusion tensor formalism (56). In the second step, the apparent diffusion coefficients (56) of the flip-up (+) and flip-down (−) DTI data were combined using a local weighted-sum approach:

\[ \text{ADC}(r)^w = \frac{w^+(r)\text{ADC}(r)^+ + w^-(r)\text{ADC}(r)^-}{w^+(r) + w^-(r)}. \]

where the spatially varying weighting \(w[r]\) is given by a Lorentzian function of the error in the tensor fit:

\[ w^j(r) = \frac{1}{1 + (\chi^j(r))^2} \quad \text{for} \quad j \in \{+', '-', -\}, \]

where \(x(r)\) is the normalized maximum-norm of the tensor-fit error \(e' = [e_1'(r), ..., e_N'(r)]\) over all applied diffusion directions (\(N = \text{number of directions}\), i.e., \(x'(r) = \max_{k \in \{1,..,N\}} (e_k'(r)) / \overline{\sigma}\)). The corresponding normalization factor \(\overline{\sigma}\) is given by the spatially averaged root-mean-square of the tensor-fit error

\[ \overline{\sigma} = \frac{1}{N_{\text{ROI}}} \sum_{j=1}^{N_{\text{ROI}}} \left( \sum_{k=1}^{N} \sqrt{\chi^2_k(r)} \right)^2, \]

with \(N_{\text{ROI}}\) being the number of voxels in the brain.

**Assessment of COVIPER**

Four types of FA images were calculated from the diffusion tensor of the affected (DTI1\textsuperscript{+} and DTI1\textsuperscript{-}) and reference (DTI2\textsuperscript{+} and DTI2\textsuperscript{-}) datasets: two directly (FA\textsuperscript{1,2} and FA\textsuperscript{1,2}) and the other two using their combination (arithmetic mean of the images, FA\textsubscript{mean}\textsuperscript{1,2}, and weighted sum of the images, FA\textsubscript{w}\textsuperscript{1,2}). To remove residual anatomical differences due to movement, each FA image of the affected datasets (DTI1\textsuperscript{-}) was registered to the corresponding FA image of the reference dataset (DTI2\textsuperscript{-}). Next, the vibration-induced bias in FA (\(\Delta F_{\text{bias}}\)) and its corrections (\(\Delta F_{\text{bias}}^\text{mean}\) and \(\Delta F_{\text{bias}}^\text{w}\)) were assessed by means of a root-mean-square FA difference between affected and reference data:

\[ \Delta F_{\text{bias}} = \sqrt{\sum_{r \in \text{ROI}} (F_{\text{a}}(r) - F_{\text{a}}(r))^2 + (F_{\text{a}}(r) - F_{\text{a}}(r))^2}, \]

and

\[ \Delta F^i = \sqrt{\sum_{r \in \text{ROI}} (F_{\text{a}}(r) - F_{\text{a}}(r))^2} \quad \text{for} \quad i \in \{\text{mean}', 'w', \} \]

with ROI being a region-of-interest that was affected by vibration artifacts and calculated from the tensor-fit error.
map of the short-TR dataset (Table 1). Finally, to assess miscellaneous effects of the proposed correction method (\(DFA_{misc}\)), the root-mean-square FA difference between the arithmetic-mean and weighted-sum combination was calculated for the reference data with reduced vibration artifacts, i.e., no correction effect for vibration artifacts was expected:

\[
DFA_{misc} = \sqrt{\sum_{r \in ROI} \left( FA^W(r) - FA^\text{mean}(r) \right)^2}
\]

RESULTS

Figure 2a shows an example of the signal-dropout due to vibration in an axial slice of a diffusion-weighted image (top row) as well as the corresponding shift in k-space (middle row, amplitude of complex raw data) and the projection of the k-space signal along the dashed line (bottom row). Severe signal-loss occurs (red arrow) if the echo center is shifted toward the shorter k-space edge (\(k_{\text{min}}\) for the blip-down data). As hypothesized, this signal-loss can be recovered if the PE direction is reversed (blip-up image). Note that the blip-up image shows also rudimentary signal-loss artifacts, however, in another region of the brain (yellow arrow) meaning that a partition of the echo is shifted towards \(k_{\text{max}}\) that is, the shorter k-space edge for the blip-up data. b: The reference data were acquired as in (a) but using a longer TR = 170 ms to reduce the vibration-induced echo shift effect. As expected, the diffusion-weighted images (top row) show no signal-dropout. Note that the geometrical distortions (due to susceptibility and eddy current effects) in blip-up and blip-down images (top row in a and b) have not been corrected to avoid interpolation artifacts. [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

Figure 3 shows the FA and the root-mean-square of the error of the tensor fit, and Fig. 4 shows the quantified bias in FA. For dataset DTI\(_1\) the vibration-induced bias in FA was visible in at least one dataset (blip-up or blip-down) of each subject (Fig. 3a,b, arrows), while the extent of the bias varied between individuals (\(DFA_{\text{bias}} = 0.38\); \(DFA_{\text{bias}} = 0.29\); \(DFA_{\text{bias}} = 0.39\); Fig. 4). The artifact manifested itself in different regions for the blip-up (Fig. 3a, yellow arrows) relative to the blip-down data (Fig. 3b, red arrows). Averaged over subjects the standard arithmetic mean combination of blip-up and blip-down data reduced the vibration-induced bias in FA by 49% (from \(DFA_{\text{bias}} = 0.35\) to \(DFA_{\text{bias}} = 0.18\), Fig. 4). In contrast, the proposed COVIPER correction based on weighted-sum combination of blip-up and blip-down data reduced the vibration-induced bias in FA by 72% (\(DFA^W = 0.1\)) and the resulting maps showed better correspondence to the reference FA maps (Fig. 3d,e). The contribution of miscellaneous effects that could not be attributed to vibration effects in the COVIPER method was about 6% (Fig. 4). Note that the arithmetic-mean combination affects the tensor estimate for all regions showing vibration artifacts in the original blip-up or blip-down data, i.e., the union
of affected regions in the blip-up and blip-down data. As a result the arithmetic-mean combination may show more widely spread artifacts (see regions highlighted by yellow and red arrows in Fig. 3a–c), than any of the two original datasets (although usually at a lower level).

**DISCUSSION**

We developed and validated a refined model and a correction of vibration artifacts in DTI using phase-encoding reversal (COVIPER). We acquired two DTI data sets with
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opposite PE direction (blip-up and blip-down data) and averaged them weighted by a function of the residual error of their tensor fit. Comparison with reference data containing negligible vibration artifacts showed that COVIPER efficiently reduced bias in FA due to vibration-induced signal loss. In contrast, the widely applied arithmetic mean of blip-up and blip-down data yields a smaller reduction in bias in FA and may even show more widely spread artifacts than any of the two original blip-up or blip-down datasets (though usually at a lower level).

The refined theoretical model for vibration-induced artifacts in DTI is based on the quantitative model of k-space shifts due to rigid body rotations (45,46) and the qualitative description of non-rigid body brain-tissue movement due to vibration (23,24). For axial echo-planar imaging acquisition with an anterior-posterior PE direction, the model predicts that non-rigid rotational in-plane movement of the brain-tissue and the strong diffusion gradients in x direction lead to the shift of the echo center (Eq. 2) in the PE direction. One consequence of this model is that signal-loss in Partial Fourier imaging might be recovered by inverting the PE direction (Fig. 1). The successful correction of vibration artifacts using COVIPER supports this refined model (Figs. 3 and 4).

Mechanical vibration, eddy currents and susceptibility differences are primary sources of artifacts in DTI (e.g., Ref. 20). To enable cutting-edge neuroscience applications and facilitate clinical diagnosis, it is essential to minimize these image artifacts. It has been shown that eddy currents and susceptibility-induced artifacts can be retrospectively corrected using an arithmetic-mean combination of blip-up and blip-down DTI data (e.g., Refs. 23,37,42,44, and 57). We showed that the arithmetic-mean combination should not be used in the presence of the vibration artifact. This is because the vibration artifact manifests itself in different regions for blip-up and blip-down data, when Partial Fourier imaging is used, as is common in DTI for reducing the echo time. The arithmetic-mean combination affects the tensor estimate for all regions showing vibration artifacts in the original data, i.e., the union of affected regions in blip-up and blip-down images, resulting in FA bias in previously unaffected regions (Fig. 3). Here, we proposed to reduce the vibration artifact using a weighted-average that reduces the contribution of the data affected by the vibration artifact as estimated from the error in the tensor fit. The weighted-average combination showed the smallest bias in FA and the best correspondence to reference DTI data, with an average reduction of 72% in the residual error in FA (Fig. 4).

In this study, we used zero filling for Partial Fourier image reconstruction. However, the signal-dropout artifact due to vibration might be further distributed across space if other Partial Fourier image reconstruction methods such as homodyne reconstruction were used (45).

The performance of the suggested correction method will depend on the extent of the shift of the echo center; if the shift is too large, i.e., the shift is in the same order of magnitude as the longest extent of the k-space window (Fig. 1), no signal will be recovered. To overcome this potential problem (which we did not observe), the analogy between echo shifts due to susceptibility artifacts (38,48) and vibration artifacts could be further exploited by applying other established techniques for reducing signal loss in gradient echo echo-planar imaging such as y-shim gradient prepulses (38) or appropriate slice tilts (49,58).

To save scan time, cardiac triggering was not used in this study. However, we expect the cardiac pulsation artifacts (see e.g., Refs. 21 and 30) to be minor compared to the vibration artifacts in the short TR data (DTI1), because most of the residual error of the tensor fit could be explained by the vibration artifact (Fig. 3). However, the COVIPER method can also be combined with cardiac triggered data. However, it must be noted that the first slice after each trigger pulse is acquired with an increased slice TR (i.e., due to waiting for the trigger and avoiding systole). Because one of the remedies recommended to avoid vibrations induced artifacts is expanded TR (23) cardiac triggering is expected to be partially beneficial in this respect.

One might argue that the residual error of the estimated diffusion tensor fit, which itself is affected by the vibration artifact, is not the optimal choice for the weighting process (i.e., to assess data quality) because there is a certain circularity in the reasoning. Despite the problem of circularity, previous studies successfully used the residual error of the tensor fit to assess and correct perturbed DTI data. It was used, e.g., to detect the improvement of data quality after correcting for eddy currents (35,59), or to down-weight regions with poor data quality to estimate local perturbation fields (60), or to detect regions affected by the vibration artifact (23).
We used a field map based correction (50) for susceptibility-induced distortions, to geometrically match blip-up and blip-down data. Although it has been shown that the field map based correction can be applied to DTI data (25,37), the correction may be finessed by methods directly exploiting the information in the PE reversed images (23,37,42,44,57).

CONCLUSION

We have presented a new correction method for vibration artifacts that uses two DTI data sets with reversed phase-encoding direction (blip-up and blip-down) to recover vibration-induced signal-loss in Partial Fourier phase-encoding direction (blip-up and blip-down) to PE reversal. The combination of these different facts (23,25,27) as well as with those for eddy currents and susceptibility-induced distortion (e.g., Refs. 37,41, and 44), which are also based on DTI data acquired with PE reversal. The combination of these different approaches is an important future direction of research towards providing DTI with minimal artifacts.

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