Characterization and correction of time-varying eddy currents for diffusion MRI

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Funding information
Natural Sciences and Engineering Research Council of Canada (Grant No. RGPIN-2018-05448); Canada Research Chairs (Grant No. 950-231993); Canada First Research Excellence Fund to BrainsCAN; NSERC CGS M program; and Ontario Graduate Scholarship program

\textbf{Purpose:} To develop and test a method for reducing artifacts due to time-varying eddy currents in oscillating gradient spin-echo (OGSE) diffusion images.

\textbf{Methods:} An in-house algorithm (TVEDDY), that for the first time retrospectively models eddy current decay, was tested on pulsed gradient spin echo and OGSE brain images acquired at 7 T. Image pairs were acquired using opposite polarity diffusion gradients. A three-parameter exponential decay model (two amplitudes and a time constant) was used to characterize and correct eddy current distortions by minimizing the intensity difference between image pairs. Correction performance was compared with conventional correction methods by evaluating the mean squared error (MSE) between diffusion-weighted images acquired with opposite polarity diffusion gradients. As a ground-truth comparison, images were corrected using field dynamics up to third order in space, measured using a field monitoring system.

\textbf{Results:} Time-varying eddy currents were observed for OGSE, which introduced blurring that was not reduced using the traditional approach but was diminished considerably with TVEDDY and field monitoring–informed model-based reconstruction. No MSE difference was observed between the conventional approach and TVEDDY for pulsed gradient spin echo, but for OGSE TVEDDY resulted in significantly lower MSE than the conventional approach. The field-monitoring reconstruction had the lowest MSE for both pulsed gradient spin echo and OGSE.

\textbf{Conclusion:} This work establishes that it is possible to estimate time-varying eddy currents from the actual diffusion data, which provides substantial image-quality improvements for gradient-intensive diffusion MRI acquisitions like OGSE.

\textbf{KEYWORDS}
brain, diffusion MRI, eddy currents, field monitoring, OGSE, validation

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Magnetic Resonance in Medicine

RESEARCH ARTICLE

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1 | INTRODUCTION

The diagnostic capabilities of diffusion MRI (dMRI) are largely dependent on the gradients used to encode the behavior of water molecules within biological tissue. Advancements in gradient hardware have the potential to broaden the clinical role of dMRI; however, eddy currents induced by these strong gradients cause image artifacts that degrade diagnostic efficacy.1

Traditionally, dMRI data are acquired using pulsed gradient spin echo (PGSE), in which two gradient lobes with long duration are applied successively.2 The time between the two gradient lobes and their duration determines the effective diffusion time, which characterizes how long water molecules are allowed to probe their local environment.3 In PGSE, the effective diffusion time is inherently long, which limits sensitivity to varying microstructural length scales. In contrast, oscillating gradient spin echo (OGSE) has been shown to enable short diffusion times through the use of successive short diffusion-weighting periods, with diffusion time scaling inversely with OGSE waveform frequency.4 Variation of the apparent diffusion coefficient with OGSE frequency has been observed in the healthy human brain.5,6,22 and OGSE has been used in conjunction with PGSE to provide additional insight into acute ischemic stroke.7 However, the strong oscillating gradients required for successful OGSE acquisitions necessitate the use of high-amplitude gradients with multiple slew s. These gradients ultimately induce strong eddy currents that degrade the quality of OGSE data to a greater extent than PGSE.5,9 Accordingly, a robust technique is needed to correct the induced eddy currents that may be unique to this diffusion-weighting method.

There are a variety of precautionary measures taken to compensate for the effects of eddy currents, including shielding of magnetic field coils and pre-emphasis of magnetic field gradients. Despite these precautions, eddy current distortions remain a significant burden to dMRI.10 Field monitoring (FM) technology11–13 is a robust solution in research settings but has seen little clinical adoption due to high costs and hardware requirements. A clinically feasible solution that greatly reduces higher-order eddy currents is the use of bipolar diffusion gradients,14 but this limits the choice of diffusion gradients. Postprocessing eddy current–correction techniques offer affordable, convenient implementation within routine clinical workflow, with the potential to handle various gradient waveforms. A prominent postprocessing method for correcting diffusion gradient–induced eddy currents is the “eddy” algorithm in FMRIB Software Library (FSL).15 The FSL eddy and other similar packages correct for eddy currents that are assumed to remain constant with time, which leads to simple affine distortions that are relatively straight-forward to detect and correct. Although generally effective for PGSE, the applicability of this assumption has not been verified for advanced forms of dMRI, such as OGSE, which have multiple gradient ramps that run the full span from the negative maximum to positive maximum gradient level.

In this work, we use dynamic FM to characterize eddy currents induced by in vivo PGSE and OGSE acquisitions. We then use this characterization to validate a new automatic method that uses the actual diffusion-weighted data to estimate the parameters for an eddy current correction model that, contrary to existing automatic approaches, includes a finite time-constant of eddy current decay.16 Improved image quality and diffusion dispersion mapping6 are shown when using this technique compared with FSL eddy.

2 | METHODS

2.1 | Magnetic resonance acquisition

Two healthy male subjects were scanned on a Siemens MAGNETOM 7T Plus head-only system (80 mT/m gradient strength and 350 T/m/s slew rate). This study was approved by the institutional review board at Western University, and informed consent was obtained before scanning. Diffusion MRI data were acquired in a single scan using an in-house sequence consisting of standard PGSE (effective diffusion time = 52.8 ms) and cosine modulated trapezoidal OGSE (40 Hz). The peak slew rate used was 310 T/m/s. Signal readout was performed using singleshot EPI with an echo train length of 112, an echo spacing of 0.52 ms, a dwell time of 2.1 μs, and a pixel bandwidth of 2125 Hz/Px. The PGSE and OGSE diffusion-sequence diagrams are shown in Figure 1. The remaining parameters were b = 750 s/mm², four-direction tetrahedral encoding, and b = 0 acquisitions with six averages each, full Fourier encoding, TE = 124 ms, TR = 11 seconds, FOV = 224 × 224 mm², 2-mm isotropic in-plane resolution, 40 slices (2 mm), and scan time = 9.5 minutes. The sequence was prepared similarly to the diffusion dispersion–sequence optimization parameters that have been previously determined,6 except here half of the averages were acquired with negative diffusion gradient polarity. Eddy current compensation was provided by the scanner vendor and calibrated during their normal maintenance.

2.2 | Eddy current characterization and FM model-based reconstruction

An FM system consisting of 16 19F NMR field probes connected to an acquisition system (Skope, Zurich
Switzerland) was used to monitor the evolution of field dynamics during the EPI readout period for the same diffusion dispersion protocol performed on the healthy patients. The probes were spherically distributed on a scaffold in verified, fixed positions, and the scaffold was positioned near isocenter in the magnet’s frame of reference.17 Excitation and signal reading of the $^{19}$F probes was handled by inserting a transistor-transistor logic pulse in the MR sequence before the initialization of the readout gradients. During the EPI readouts, the signal amplitude from all 16 field probes did not fall below 10% of the initial probe signal amplitudes.

The dynamic spatially varying phase data, which are described with third-order spherical harmonics, were computed from the FM raw data using vendor-provided MATLAB (MathWorks, Natick, MA) software.12 Ignoring the effects of $B_0$ inhomogeneity, the signal measured under the influence of the gradients and diffusion gradient–induced eddy currents is given by

$$S(t_n) = \sum_m \rho(r_m) \exp \left( -i \left[ \sum_l \left( k_{l,\text{traj}}(t_n) + k_{l,\text{eddy}}(t_n) \right) b_l(r_m) \right] \right).$$

An overview of the reconstruction pipeline is shown in Figure 2A, where the eddy current correction step that occurs after N/2 Nyquist ghost correction20 and ramp sampling regridding along the readout direction is either the FM correction method described here or TVEDDY (subsequently). For the FM approach, the acquired data were corrected for diffusion-gradient eddy currents using an iterative conjugate gradient solution for $\rho(r_m)$.21 The details of this reconstruction are specified in Supporting Information S1. All receiver channels were considered separately and were combined using a SENSE1
reconstruction after eddy current correction. Receiver combination was followed by principle component analysis denoising applied to the complex data, removal of the phase, export to the NIFTI data standard, Gibbs ringing reduction using the Kellner method, and application of FSL eddy to correct motion between different diffusion weighted acquisitions, which is not corrected by Equation 1.

2.3 Data driven correction: TVEDDY

Eddy currents lead to the k-space trajectory deviating from the intended trajectory, which primarily manifests as distortions and blurring along the PE direction due to its slow k-space traversal. Eddy current fields are generally spatially and temporally varying in nature, but a linear approximation of the spatial variation typically accounts for most of the eddy current–induced phase. As such, eddy current fields produced by diffusion gradients can be accurately defined by a spatially invariant term \( B_0 \) and linear field gradient terms \( G_x, G_y, \) and \( G_z \), all of which are generally time-varying. Each term is often modeled in time as a sum of exponentially decaying functions with discrete time constants. The FSL eddy and similar data-driven approaches assume that there is a single, infinite time constant.

Our approach, which will be referred to as time-varying eddy (TVEDDY) because it considers the time-varying nature of eddy currents, assumes that the eddy current fields approximated by the \( B_0 \) and linear gradient eddy currents are each described by

\[
k_{\text{L, eddy}} = A_{r,j} \exp \left\{ -\frac{t_n}{\tau_j} \right\} + A_{\text{inf},j},
\]

FIGURE 2  (A) Reconstruction pipeline. Either the time-varying model (TVEDDY), time-constant model (TCEDDY), or field-monitoring (FM) eddy current modeling was used in the highlighted eddy current correction step. FMRII Software Library (FSL) eddy is applied in the final step of the reconstruction to correct for motion. (B) The TVEDDY algorithm. Eddy current–induced artifacts are corrected starting with k-space data that were acquired with opposite polarity diffusion gradients. The eddy current parameters \( A_{r,j}, A_{\text{inf},j} \) and \( \tau_j \) which are corrected using phase shifts and regridding (net operator “F”), and translations (T) and rotations (R), are iteratively adjusted until the mean squared error (MSE) has converged, where \( j = \{0,1,2,3\} \) corresponds to the zeroth and first-order eddy currents along x, y, and z, respectively (Equation 2). The data sets shown here were synthetically distorted with large eddy current and motion parameters to illustrate the correction process. The sign of G indicates diffusion gradient polarity. Abbreviation: PCA, principal component analysis.
where \( A_{r,l} \) and \( A_{\inf,l} \) are eddy current amplitudes; \( \tau_l \) is a finite time-constant; and \( l = \{0, 1, 2, 3\} \) corresponds to \( b_l \) (\( r_m \) = \{x, y, z\}, respectively. It will be assumed henceforth that x, y, and z correspond to the readout, PE, and slice-select axes, respectively. The values of \( A_{\inf,l} \) and \( A_{r,l} \) will produce image artifacts that appear as bulk distortions (e.g., shearing, stretching) and spatially dependent blurring, respectively. For TVEDDY, spherical harmonic terms of second order and higher are ignored, and it is presumed that the polarity of \( A_{\inf,l} \) and \( A_{r,l} \) will be equal in magnitude but opposite in direction if gradient polarity is reversed.\(^{27}\) Accordingly, to improve the ability of our approach to use the data to determine a time constant and two separate eddy current amplitudes, TVEDDY currently requires that for each diffusion direction, two images are acquired with antipodal diffusion gradients. The optimization procedure aims to determine the eddy current model parameters that minimize the mean squared error (MSE) between images acquired with opposite diffusion gradient polarity through multistart gradient descent (MATLAB). Every iteration, the parameters \( A_{r,l} \) \( A_{\inf,l} \) and \( \tau_l \) are used to estimate \( \rho(r_m) \) from the raw data using a combination of complex phase shifts and regridding\(^{28}\) (Figure 2B; algorithm details found in Supporting Information S1). Notably, a noniterative estimation based on the fast Fourier transform is possible, because only the zeroth and first-order terms are considered for TVEDDY, and because the k-space position changes from the diffusion gradient eddy currents are not typically large enough to introduce violations of Nyquist criteria. The eddy current parameters are applied to all slices simultaneously (i.e., all slices were assumed to have the same \( A_{r,l} \) \( A_{\inf,l} \) and \( \tau_l \)). The lower threshold of \( \tau_l \) was 1.6 ms, as smaller time constants decay to zero within a few PE lines and the upper threshold of \( \tau_l \) was heuristically determined to be 25 ms, as larger time constants become close to approximating an infinite time-constant (which is already modeled by \( A_{\inf,l} \)). Ten randomly generated sets of eddy current parameters were used as starting points in the optimization.

During a scan, the subject may move between the acquisition of volumes with opposite polarity diffusion gradients. To overcome this limitation, rigid body motion parameters were jointly determined within the optimization to improve the accuracy of the eddy current parameter estimations. The PE shifts were omitted from the motion model to avoid redundancy with the \( A_{\inf,0} \) term, which also creates bulk PE shifts. Finally, our implementation of TVEDDY allows for manually setting \( A_{r,l} \) to zero, which results in a time-constant correction similar to FSL eddy; for simplicity, this will be referred to as time-constant eddy (TCEDDY). Source code and an example implementation of TVEDDY are available from https://doi.org/10.17605/OSF.IO/4XTF3.

The primary purpose of TVEDDY is to regrid trajectory errors that occur due to time-varying eddy currents, and it does not compensate for subject motion. To correct for motion, FSL eddy was applied after our custom image reconstruction that contains the TVEDDY algorithm. Also, to reduce the number of interpolations applied to the data, the final iteration of TVEDDY does not apply the motion parameters. Accordingly, our custom reconstruction uses TVEDDY to correct the in-plane eddy current effects without any interslice interpolation, and all interslice interpolation to correct for motion is applied as a postprocessing step through FSL eddy (version 6.0.1). Default FSL eddy settings were used with the exception of morder, an intravolume motion correction setting, which was set to four for all corrections.

### 2.4 The TVEDDY validation

#### 2.4.1 Validation 1: algorithm convergence

The optimization solved by TVEDDY is generally nonconvex, and to assess its ability to find the global minimum, 80 pairs of images were synthetically generated from one of the PGSE diffusion-weighted 2D slices acquired in vivo. Pulsed gradient-echo spin echo was chosen for this purpose because time-varying eddy current effects were small for PGSE. Each pair of images was generated assuming eddy currents with opposite polarity and no motion according to Equation 1, where the eddy current parameters \( A_{r,l} \) \( A_{\inf,l} \) and \( \tau_l \) (Equation 2) were randomly generated.

After these distortions with known parameters were applied using the TVEDDY forward model (Equation 1 with \( k_{\text{eddy}} \) defined by Equation 2), noise with a SNR of 10 was added to each image. Eddy current correction was performed on a single pair of oppositely distorted slices at a time, and the recovered parameters were correlated against the known parameters to determine whether TVEDDY could accurately correct distorted data after the addition of noise.

#### 2.4.2 Validation 2: model accuracy

Here, synthetic data were generated from a volume of in vivo PGSE slices. Instead of using Equation 2, eddy current–distorted raw data were generated on a slice-by-slice basis using Equation 1, in which the zeroth and first-order (i.e., \( l = \{0,1,2,3\} \)) \( k_{\text{eddy}} \) data were acquired from an OGSE acquisition using FM, which does not necessarily fit the exponential assumption used in Equation

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2. The volume of simulated images was then processed using TCEDDY and TVEDDY, and eddy model parameters were extracted for comparison of the zeroth and first-order eddy current deviations predicted by the two models to the ground truth. Accordingly, this simulation tests the accuracy of the exponential assumption used by TVEDDY. Before comparison of TVEDDY to the ground-truth eddy current data, global background phase shifts common in both diffusion polarities were removed from the FM data. Given that TCEDDY and TVEDDY only capture inverse eddy current effects, any time-varying field terms that are consistent between the two acquisitions will not be accounted for. Accordingly, to separate the common background phase effects from eddy current induced phase profiles in opposite polarity scans, the profiles were decomposed as follows:

\[
k_{l,\text{eddy}}[\text{pos}] = k_{l}[\text{BG}] (t_n) + k_{l}[\text{EC}] (t_n)
\]

\[
k_{l,\text{eddy}}[\text{neg}] = k_{l}[\text{BG}] (t_n) - k_{l}[\text{EC}] (t_n)
\]

where \(k_{l,\text{eddy}}[\text{pos}]\) and \(k_{l,\text{eddy}}[\text{neg}]\) represent the total FM k-coefficient deviations for positive and negative diffusion gradient directions; and \(k_{l}[\text{BG}]\) and \(k_{l}[\text{EC}]\) are the predicted distortions resulting from background and eddy current effects, respectively. A zeroth-order decomposition example is shown in Figure 3.

2.5 | Technique comparisons

Reconstructions that used no correction (neither the custom eddy current–correction step in Figure 2A nor FSL eddy), FSL eddy only, TCEDDY, TVEDDY, and FM were compared by computing the MSE in pixel intensity between images acquired with opposing diffusion gradient polarities. For the TCEDDY, TVEDDY and FM cases, FSL eddy was still applied afterward to correct for subject motion; accordingly, any comparisons investigated the incremental gain of combining these eddy current–correction approaches with FSL eddy. Applying FSL eddy ensures that the motion-correction process is consistent for all techniques, reducing the ambiguity in what MSE differences may be attributed to. For each approach and subject, the MSE was measured for all volume pairs with opposing polarity diffusion gradients, normalized by the mean MSE of each subject’s PGSE volumes without correction, and averaged over all subjects, diffusion directions, and averages (12 total pairs per subject, each for PGSE and OGSE). A mask made using FSL BET was used to exclude voxels outside the brain. Paired t-tests were used to determine statistical significance between the MSE measurements of different techniques (\(N = 24\)).

The diffusion dispersion (\(\Delta D\)) is a dMRI parameter that represents the difference in the apparent diffusion coefficient between PGSE and OGSE as follows:

![Figure 3](image-url)
\[
\Delta D = -\frac{1}{N_\omega} \sum_{i=1}^{N_\omega} \ln \left( \frac{S_{\omega i}}{b_{\omega i}} \right) + \frac{1}{N_\omega} \sum_{i=1}^{N_\omega} \ln \left( \frac{S_{0,0,i}}{b_{0,0,i}} \right),
\]

where \( N_\omega \) and \( N_{0,0} \) are the number of OGSE and PGSE acquisitions, respectively; \( S_{\omega i} \) and \( S_{0,0,i} \) are the direction-dependent diffusion-weighted signals at the frequencies \( \omega > 0 \) and \( \omega = 0 \), respectively; and \( b_{\omega i} \) and \( b_{0,0,i} \) are the direction-dependent b-values at frequencies \( \omega > 0 \) and \( \omega = 0 \), respectively.\(^6\) The \( \Delta D \) maps were computed after acquired OGSE and PGSE data were corrected with FSL eddy, TCEDDY, TVEDDY, and FM.

3 | RESULTS

Generally, PGSE eddy current–induced phase increased approximately linearly over the course of the readout, whereas OGSE exhibited nonlinear time dependence (Figure 4).

3.1 | TVEDDY validation 1: algorithm convergence

Synthetically applied eddy current parameters were accurately recovered using the algorithm’s parameter determination capabilities, as indicated in Figure 5 by the good agreement with the expected linear slope of 1 between ground truth and predicted parameters. The slope of the regression line for \( \tau \), \( A_\tau \), and \( A_{inf} \) is 0.689, 0.939, and 0.915, respectively, which indicates a tendency to underestimate the larger values of \( \tau \). Bland-Altman analysis revealed little overall bias.

3.2 | TVEDDY validation 2: model accuracy

For uniform and relatively linear phase accrual from eddy currents, as typically exhibited by PGSE, the time-constant correction manages to apply phase shifts that reflect the ground-truth phase error well (Figure 6). However, the temporally varying OGSE eddy currents are not well characterized by the time-constant model TCEDDY. The corrections applied using TVEDDY match well with the linear applied corrections for PGSE, while also providing a closer approximation to the more complex eddy currents produced by OGSE, compared with TCEDDY.

Figure 7A shows example reconstructed images without correction and the associated difference image after correction with FSL eddy, TVEDDY, and FM eddy current modeling. In the original OGSE image, blurring is most noticeable near contrast changes, such as near the ventricles and the cortical ribbon. The difference images show how this blurring persists after correction with FSL eddy but is greatly reduced using the time-varying eddy current model, which is qualitatively comparable to the FM correction.

In Figure 7B, histograms made with each correction technique have the same area, yet the rightward shift in the FSL eddy histogram reveals more voxels with large errors.

Decreasing MSE values are observed with increasing model complexity (Figure 8). Statistically significant changes in MSE (\( p < 0.01 \)) were observed between each neighboring correction technique in Figure 8, except between TCEDDY and TVEDDY, and TVEDDY and up to first-order FM correction for PGSE volumes.

The \( \Delta D \) maps made using TVEDDY and FM correction show recovered signal voids and improved signal homogeneity of problematic regions in maps made using FSL eddy only, highlighted by yellow circles in both slices of each subject (Figure 9). Notably, the nature of the time-varying eddy currents depends on the diffusion gradient direction, which can result in deleterious signal voids when combining the different directions to compute \( \Delta D \), and small \( \tau \) can create signal pile-up in k-space, which results in ringing artifacts that can be observed near the frontal lobe.

4 | DISCUSSION

In this work we have revealed that advanced diffusion-weighting approaches may introduce eddy currents that are not compensated well by standard vendor precompensation, introducing a new eddy current–correction technique that models diffusion gradient–induced eddy current decay. Current retrospective techniques do not account for this decaying behavior, which results in residual image blurring along the PE direction.

The high \( R^2 \) values and low SEM of the regression line slopes in Figure 5A, and the average difference between applied and recovered parameters being less than or equal to 0.02 in Figure 5B, indicate a consistent relationship between recovered and applied parameters. The \( A_\tau \) and \( A_{inf} \) regression line slopes close to unity in Figure 5A, suggesting that TVEDDY can accurately recover a wide range of eddy current amplitudes. The analysis of recovered \( \tau \) parameters reveals an underestimation between recovered and applied \( \tau \), which tends to increase with increasing \( \tau \). This finding is likely because at large \( \tau \) the blurring becomes very subtle and difficult.
to detect. Our current prototype version of TVEDDY correction of a pair of volumes acquired with opposite polarity diffusion gradients requires up to 1 hour. Thus, a 2D version of TVEDDY that ignores $k_2$ was used to assess convergence, which enabled a wide range of eddy current parameters to be tested efficiently. Assessing convergence in 3D may result in slightly different performance; however, TVEDDY’s ability to capture eddy current behavior in a volume of data (Figure 6) suggests that it can still converge when $k_2$ is included to correct multiple slices.

Looking at the FM zeroth-order and first-order $k$-coefficient deviations produced by eddy currents, the PGSE sequences generally exhibited linear behavior that could be modeled with a single constant term. Conversely, the OGSE scans displayed more complex eddy current evolutions that are not accurately characterized by linear phase accrual. Figure 6 shows how TCEDDY can accurately model the most basic linear phase accruals but performs more poorly with any offsets that deviate from linearity. Because TCEDDY is very similar in approach and performance to FSL eddy, it can be inferred that FSL eddy also struggles with correcting distortions from eddy currents with decay constants on the order of the readout duration. Improved blurring correction can be seen with TVEDDY, where including an additional eddy current term with a finite time-constant provides additional degrees of freedom to correct for time-varying eddy currents. Overall, this results in correction profiles that more accurately capture the behavior of applied distortions, which resulted in less blurring/artifacts in reconstructed OGSE images (Figure 7) and $\Delta D$ maps (Figure 9), as well as reduced MSE between images acquired with opposite diffusion gradient polarity (Figure 8).

As described in section 2, the displayed FM eddy current–induced profiles in Figure 6 are not results monitored directly by the system, as the original data were decomposed into two time-varying functions, $k_{[BG]} (t_n)$ and $k_{[EC]} (t_n)$, with $k_{[BG]} (t_n)$ being the component that reversed polarity with opposite diffusion gradient polarity (Figure 3). We suspect that the background effects arise from temperature increases from the high duty-cycle OGSE diffusion gradients. The temporal zeroth and first-order phase accrual originating from the background effects, $k_{[BG]} (t_n)$, were approximately linear (Figure 3), primarily affecting alignment of images depending on the diffusion gradient direction, while contributing very little to blurring. Because FSL eddy focuses on correcting linear phase variations, this residual misalignment between nonparallel diffusion directions was likely corrected by FSL in the last step of Figure 2A. Accounting for nonlinear background effects provides almost no improvement to image quality, as seen in correction of
PGSE volumes using FM1 relative to TVEDDY, confirming its small contribution.

An unexpected result was the oscillatory eddy current phase accumulation for OGSE (Figure 4). This suggests that there may be an interaction between residual mechanical vibration of the gradient hardware and eddy current fields. Although TVEDDY provides improved correction by accounting for eddy current decay, it does not capture this oscillating behavior. Accurately capturing this behavior would improve correction quality, as seen by the reduced MSE for FM1 recon compared with TVEDDY in OGSE volumes (Figure 8), but the required degrees of freedom for modeling would be high, and a robust fitting procedure would likely not be possible. Instead, TVEDDY could be extended to include multiple decaying terms; however, this would increase computation time and may reduce the reliability of correction due to overfitting. Although FM can fully account for this oscillating behavior, it requires dedicated FM hardware. Alternatively, gradient-impulse response–function approaches may be an effective method to account for the eddy currents, but they require detailed knowledge of the diffusion gradient waveforms, which TVEDDY does not. Additionally, the eddy currents produced by OGSE show some evidence of nonlinearity (Supporting Information S2), which would violate linearity assumptions required for gradient-impulse response function–based corrections. Notably, this oscillatory eddy current behavior may be vendor-specific or hardware-specific.

Other strategies for eddy current reduction in OGSE include sinusoid (or near-sinusoid) oscillating gradient ramps and the exclusion of particular encoding frequencies in the OGSE waveform that are responsible for artifacts. However, these techniques come with costs that we could potentially avoid using our method. For example, the exclusion of select frequencies likely requires time-consuming calibration, and having to choose a lower than optimal frequency would decrease OGSE contrast. The use of non-trapezoidal gradient lobes reduces efficiency in generating desired b-values. Notably, these strategies can be used in combination with our method if necessary. In addition, while this work demonstrates the utility of TVEDDY for OGSE, it could likely be applied to other advanced diffusion MRI approaches like b-tensor encoding, where it would still be possible to acquire data with all gradient channels applied with opposite polarity.

The TVEDDY method was developed to correct time-varying eddy current distortions that lead to blurring and ringing. This requires pairs of images to be acquired with opposing diffusion gradient polarity, which will generally come at no cost to scan time as typical OGSE scans already require many signal averages. The FSL eddy was used to correct remaining misalignment between

**Figure 5** (A) The TVEDDY’s ability to recover known eddy current parameters. The linear regression line with slope $m$ and standard error SE are shown in comparison with the expected true line with a slope of 1. (B) A Bland-Altman plot in which the difference between a recovered and applied parameter pair is plotted against the mean of that pair. The solid lines in each plot show the average difference between recovered and applied parameters, while the dotted lines represent the interval in which 95% of differences are included. In both panels, $\tau$ is in units of milliseconds and amplitudes are approximately in units of voxels, where $A_{\text{inf}} = 1$ indicates a distortion that causes a shift of one voxel and $A_\tau = 1$ indicates blurring on the order of one voxel.
FIGURE 6  Ground-truth FM zeroth and first-order eddy current terms (i.e., $k_{EC}$ in Equation 3) compared with the output of TCEDDY and TVEDDY for one of the four diffusion directions acquired. The entire acquisition window is displayed (58.2 ms), and the spin echo at the center of the readout is marked. Both approaches perform a good correction for PGSE, but only TVEDDY can at least partially correct for the time-varying eddy currents observed for OGSE. Similar results were observed for the other three directions. The linear terms were multiplied by FOV/2 such that all of the terms displayed the net phase accrued by a voxel at the edge of the FOV. The labeled $x$ units are shared by all plots.
diffusion directions that occur from subject motion and the background eddy current terms (Figure 3). However, it is likely possible to combine the behaviors of TVEDDY and FSL eddy into a single, unified framework. The efficacy of TVEDDY was assessed by determining both quantitatively and qualitatively whether TVEDDY in conjunction with FSL eddy provides any additional benefit to using FSL eddy alone.

Given the opposing polarity of distortions in data acquired with opposite polarity diffusion gradients, volume pairs that are not corrupted by eddy current distortions should have very low MSE (Figure 8). For both PGSE and OGSE volumes, full third-order FM correction resulted in the most improved correction, while FSL eddy alone performed the worst out of the tested techniques. In OGSE volumes, TVEDDY outperformed TCEDDY for all volumes acquired. In PGSE volumes, there was no significant difference between correction with TCEDDY, TVEDDY and FM1, which is consistent with the conventional assumption that modeling eddy current decay is not necessary for PGSE, as the linear phase shifts are adequately handled using a static eddy current model. The reduced MSE in OGSE volumes corrected with TVEDDY and FSL eddy versus FSL eddy alone demonstrates the benefit of accounting for eddy current decay when correcting OGSE data. It is also notable that the MSE of PGSE volumes is typically higher than that of OGSE volumes. The PGSE gradients tend to induce bulk shifts in the image domain, which inflates MSE before correction. Additionally, OGSE gradients are intrinsically velocity-compensated, which ensures that there is minimal change in CSF between OGSE volumes. The substantial movement of CSF between acquisition of PGSE volumes leads to higher MSE than a corresponding OGSE volume pair once eddy...
currents have been accounted for. The reduced OGSE MSE can also be attributed to the self-cancellation properties exhibited by OGSE waveforms. Reduced second- and third-order eddy current effects were observed for OGSE relative to PGSE, consistent with the work presented by Chan et al for bipolar diffusion gradients. As a result, full third-order FM correction provides most of the correction improvement for PGSE volumes, while OGSE correction quality is improved primarily using FM1 (Figure 8).

The ΔD maps illustrate the impact of eddy current correction on a parametric map that can be computed using combinations of OGSE and PGSE scans. Notably, ΔD is vulnerable to eddy current distortions and blurring, which present differently between the PGSE and OGSE scans; this is problematic without a correction like TVEDDY, given that only the OGSE scans have blurring and ringing due to time-varying eddy currents.

A volume-based correction was used, as it provides a good balance between speed and accuracy. The assumption that eddy current parameters do not change across an imaging volume is generally not valid for the first few slices of each TR while the system achieves a steady state, and potentially when the imaging slab size is very large. Otherwise, this assumption provides an accurate definition of eddy currents for an imaging volume. Although it is possible to use a slice-based correction approach in which each slice is treated separately, this drastically increases the computation time, and our preliminary investigations (not shown) suggest that it does not provide substantial improvements in image correction.

The images displayed in this work exhibit distortions from B₀ inhomogeneity. Using parallel imaging would reduce these distortions, but as described by Arbabi et al, the low contrast between PGSE and OGSE volumes necessitates maximizing SNR and makes ΔD maps particularly sensitive to residual aliasing artifacts that can manifest through parallel imaging. It is likely possible to combine the acquired data with the FM trajectory in a B₀ map-informed model-based reconstruction to drastically reduce these distortions, but TVEDDY is designed for use in situations in which such hardware is not available; thus, it is more relevant to validate it in these realistic scanning conditions. Additionally, the implementation of GRAPPA or even partial Fourier may create stronger blurring artifacts due to more data being collected during the most rapidly varying period of the eddy currents. However, the performance of the algorithm would need to be validated for these cases.

Although rigid body motion between paired acquisitions is addressed using our technique, cardiac pulsation may have reduced the performance of TVEDDY by inducing signal changes that are not equal in magnitude and opposite in direction between image pairs. Cardiac gating could be used to mitigate the effects of pulsation on correction quality.

The FSL topup could also be implemented in a post-processing pipeline but would require an additional non-diffusion-weighted acquisition with reversed phase encoding. This reversed phase-encoding scan is distinct from TVEDDY’s reversed diffusion gradient requirement. As a result, an additional scan with PE gradient reversal (and zero amplitude diffusion...
gradients) can be performed, and the two correction methods can be applied to the same data to correct both diffusion gradient and image-encoding gradient artifacts.

5 CONCLUSIONS

This work presented and validated a new computational method for correcting diffusion gradient eddy currents that
outperforms FSL eddy for OGSE acquisitions. The capacity to correct distortions induced by advanced dMRI techniques without the substantial cost of an FM system will promote the development and clinical application of advanced dMRI.

DATA AVAILABILITY STATEMENT
The TVEDDY source code along with the necessary files required to correct provided example data are available at https://doi.org/10.17605/OSF.IO/4XTF3.

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**SUPPORTING INFORMATION**
Additional supporting information may be found in the online version of the article at the publisher’s website.

**SUPPORTING INFORMATION S1** Outlines the implementation of the signal model for the field-probe-informed and TVEDDY reconstructions

**SUPPORTING INFORMATION S2** Outlines how the eddy currents produced by the OGSE diffusion gradients exhibit evidence of non-linearity

How to cite this article: Valsamis JJ, Dubovan PI, Baron CA. Characterization and correction of time-varying eddy currents for diffusion MRI. *Magn Reson Med.* 2022;87:2209–2223. doi:10.1002/mrm.29124