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Simultaneous and Inherent Correction of B_0 and Eddy-Current Induced Distortions in High-Resolution Diffusion MRI using Reversed Polarity Gradients and Multiplexed Sensitivity Encoding (RPG-MUSE)

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Abstract

In diffusion MRI (dMRI), static magnetic field (B_0) inhomogeneity and time varying gradient eddy currents induce spatial distortions in reconstructed images. These distortions are exacerbated when high spatial resolutions are used, and many field-mapping based correction techniques often only acquire maps of static B_0 distortion, which are not adequate for correcting eddy current induced image distortions. This report presents a novel technique, termed RPG-MUSE, for achieving distortion-free high-resolution diffusion MRI by integrating reversed polarity gradients (RPG) into the multi-shot echo planar imaging acquisition scheme used in multiplexed sensitivity encoding (MUSE). By alternating the phase encoding direction between shots in both baseline and diffusion-weighted acquisitions, maps of both static B_0 and eddy current induced field inhomogeneities can be inherently derived, without the need for additional data acquisition. Through both 2D and 3D encoded dMRI acquisitions, it is shown that an RPG-MUSE reconstruction can simultaneously achieve high spatial resolution, high spatial fidelity, and subsequently, high accuracy in diffusion metrics.

Keywords

Diffusion MRI; multi-shot EPI; MUSE; spatial distortion; eddy currents; reversed polarity gradients

Introduction:

Over the past two decades, diffusion MRI (dMRI) and diffusion tensor imaging (DTI) [Mori et al. 1999; Tuch et al., 2002; LeBihan et al., 1986; Basser et al., 1994; Pierpaoli et al., 1996] have been commonly used to investigate neuronal microstructures in the brain and map brain connectivity. However, the use of fast imaging methods such as echo-planar imaging (EPI) [Turner et al., 1990] and large diffusion-weighting gradients induce geometric distortions in dMRI. Arising from both static magnetic field (B_0) inhomogeneity and gradient eddy

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currents across diffusion directions, these distortions are exacerbated when high spatial resolutions are used and ultimately create inaccuracies in DTI metrics.

Recent developments in dMRI acquisition techniques have achieved high spatial resolutions through multiplexed sensitivity encoding (MUSE) [Chen et al., 2013], in which each $b=0$ (b_0) and diffusion-weighted image (DWI) volume (across all diffusion directions) is acquired with multiple interleaved excitations using multi-shot EPI (ms-EPI) [Holdsworth et al., 2009; Jeong et al., 2013]. Such an acquisition scheme presents a unique opportunity to dynamically correct spatial distortions by incorporating reversed polarity gradients (RPG) [Bowtell et al., 1994] into the multiple interleaved excitations. In many traditional field-mapping based correction techniques, a static B_0 distortion map is separately acquired at the beginning of a scan and applied to all subsequent images (e.g. dMRI volumes) [Schneider and Glover, 1991; Weisskoff and Davis, 1992; Jezzard and Balaban, 1995; Reber et al., 1998; Holland et al., 2011; Visser et al., 2012; Dymerska et al., 2018]. However, if diffusion weighting gradients are applied to the subsequent images, additional distortions can be induced through eddy currents in the diffusion gradient coils, and thus a static B_0 distortion map derived from non-diffusion weighted data can be inadequate in representing the true distortion behavior. In another common correction technique, two complete scans (with double the scan time) are acquired with RPG to estimate and correct distortions in each diffusion volume [Andersson et al., 2001; Andersson et al., 2003; Smith et al., 2004; Holland et al., 2011; Zahneisen et al., 2017]. By applying RPG to ms-EPI, however, there is no increase in scan time and dynamic distortion maps (B_0 +eddy currents) can be individually estimated for all baseline and diffusion volumes.

Thus, we present here a technique, termed RPG-MUSE, that incorporates RPG into ms-EPI to inherently achieve dMRI data with high resolution, high spatial fidelity, and accurate DTI metrics. In RPG-MUSE, the interleaved multi-shot acquisition scheme used in a conventional MUSE scan is exploited by alternating the phase encoding (PE) polarity across the odd and even shots of each diffusion-weighted volume. This technique offers several advantages: 1) the condition of the distortion correction problem is improved by acquiring each dMRI volume with both forward and reversed PE in RPG, 2) the number of required image volumes is exactly the same as that of a conventional MUSE ms-EPI scan, imposing no increase on scan time, 3) distortion maps (B_0) are individually estimated and tailored to correct all distortions (B_0 and eddy currents) in each dMRI volume, and 4) artifacts induced by shot-to-shot motion and distortion in RPG ms-EPI data are simultaneously and dynamically accounted for during image reconstruction. Furthermore, we extend RPG-MUSE into 3D ms-EPI, where additional gains can be achieved in SNR through a secondary phase encoding performed across the slice dimension of a thick slab [Frank et al., 2010; Engstrom and Skare, 2013; Engstrom et al., 2015; Frost et al., 2015; Chang et al., 2015a,b; Song et al., 2014; Bruce et al., 2017].

Theory:

RPG in 2D MUSE dMRI:

In an ms-EPI acquisition with N_C receiver coils, the spatial frequencies of a slice are subsampled by a factor of N_A along the PE dimension in each of N_S interleaved shots. In

such an acquisition, each of the N_S shots acquires N_C reduced FOV aliased coil images that can be un-aliased into full FOV images (one for each shot) using parallel imaging techniques [Roemer et al., 1990; Pruessmann et al., 1999; Griswold et al., 2002]. In the SENSE model [Pruessmann et al., 1999], a $N_C \times 1$ vector of aliased coil measurements, a_j , for a voxel j is ideally a linear combination of N_A true un-aliased voxel values (spaced FOV_{PE}/N_A apart) in vector v_j by

$$a_j = \begin{bmatrix} S_{j,1} & \cdots & S_{j,N_A} \end{bmatrix} \begin{bmatrix} v_{j,1} \\ \vdots \\ v_{j,N_A} \end{bmatrix} = \sum_{k=1}^{N_A} S_{j,k} v_{j,k}, \quad (1)$$

where $S_{j,k}$ is a $N_C \times 1$ vector of coil sensitivity weightings for voxel j in aliasing locations $k=[1, \dots, N_A]$. While most ms-EPI acquisitions are performed with $N_A=N_S$ to retain SNR, it is possible to acquire data with $N_A > N_S$, and thus a generalized approach is outlined here. The 2D-MUSE model in Chen et al., (2013) is formed by expanding Eq. (1) to incorporate acquisitions from N_S shots by

$$\begin{bmatrix} a_{j,1} \\ \vdots \\ a_{j,N_S} \end{bmatrix} = \begin{bmatrix} S_{j,1} \theta_{j,1,1} & \cdots & S_{j,N_A} \theta_{j,N_A,1} \\ \vdots & \cdots & \vdots \\ S_{j,1} \theta_{j,1,N_S} & \cdots & S_{j,N_A} \theta_{j,N_A,N_S} \end{bmatrix} \begin{bmatrix} v_{j,1} \\ \vdots \\ v_{j,N_A} \end{bmatrix}, \quad (2)$$

where

$$\theta_{j,k,s} = e^{i \frac{2\pi}{N_A} (k-1)(s-1)} e^{i \phi_{k,s}} \quad (3)$$

represents the Fourier offset between the interleaved shots as well as the motion induced phase aliased into voxel j , $\phi_{k,s}$, drawn from aliasing location k in a full FOV SENSE reconstructed image for shot $s=[1, \dots, N_S]$. Thus, the $N_C \times 1$ vector of aliased coil measurements for shot s in voxel j in Eq. (2) is a linear combination of true un-aliased voxel values in vector v_j weighted by coil sensitivities and shot phases

$$a_{j,s} = \sum_{k=1}^{N_A} S_{j,k} \theta_{j,k,s} v_{j,k}. \quad (4)$$

To best outline the RPG-MUSE model, the MUSE formulation in Eq. (2) is expanded to perform a MUSE reconstruction on all $p_y = PY/N_A$ voxels in a column of the acquired aliased coil images at once,

$$\begin{bmatrix} a_{1,1} \\ a_{1,2} \\ \vdots \\ a_{j,1} \\ a_{j,2} \\ \vdots \\ a_{py,1} \\ a_{py,2} \end{bmatrix} = \begin{bmatrix} S_{1,1}\theta_{1,1,1} & 0 & S_{1,2}\theta_{1,2,1} & & \\ S_{1,1}\theta_{1,1,2} & & S_{1,2}\theta_{1,2,2} & & \\ & \ddots & & \ddots & \\ & & S_{j,1}\theta_{j,1,1} & & S_{j,2}\theta_{j,2,1} \\ & & S_{j,1}\theta_{j,1,2} & & S_{j,2}\theta_{j,2,2} \\ & & & \ddots & \\ & & & & S_{py,1}\theta_{py,1,1} & & S_{py,2}\theta_{py,2,1} \\ 0 & & S_{py,1}\theta_{py,1,2} & 0 & & S_{py,2}\theta_{py,2,2} \end{bmatrix} \begin{bmatrix} v_1 \\ \vdots \\ v_{PY} \end{bmatrix}. \quad (5)$$

As each element $S_{j,k}\theta_{j,k,s}$ in Eqs. (4) and (5) represents an $N_C \times 1$ vector of phase adjusted coil sensitivities, the aliasing matrix for an entire column in Eq. (5) is therefore $pyN_C N_S \times PY$ in dimension. For simplicity in representation, the full column expansion in Eq. (5) represents Eqs. (2) and (4) with $N_S = N_A = 2$, but can be easily expanded to incorporate any combination of N_S and/or N_A . The MUSE model in Eqs. (2), (4) and (5) assumes that the motion induced phase term $\theta_{j,k,s}$ is the only term that varies between shots, and that an aliased voxel, $a_{j,s}$, from any shot s is a combination of the same true voxel values in v_j weighted by the same sensitivities, $S_{j,k}$, for all shots. When a region with compression/stretching is observed in shots acquired with forward PE (odd shots), the opposite behavior (stretching/compression) is observed in the same region of shots acquired with reversed PE (even shots). Thus, by inserting full FOV SENSE reconstructed images from all shots acquired with RPG into tools such as FSL's topup [Andersson et al., 2003; Smith et al., 2004], a combined map of both static B_0 inhomogeneity and eddy current induced distortions (if present), B , can be estimated for each acquired volume. With knowledge of the distortion characteristics in each voxel across the PE dimension, the accuracy of the aliasing model in Eq. (4) can be improved to reflect which true un-aliased voxels in vector v from Eq. (5) are aliased into voxel j in shot s by

$$a_{j,s} = \sum_{k=1}^{N_A} S_{(j,k)} - p\Delta B_{j,k} \theta_{j,k,s} v_{(j,k)} - p\Delta B_{j,k}, \quad (6)$$

where $B_{j,k}$ is the level of distortion in voxel j at location k , and $p = \pm 1$ for shots with forward/reversed PE directions respectively. In a two-shot example of an aliased voxel a_j , if shot $s=1$ observes stretching in location $k=1$ ($B_{j,1} > 0$) and compression in location $k=2$ ($B_{j,2} < 0$), then shot $s=2$ observes the opposite behavior in each location. For this example, the corresponding rows of the aliasing matrix in Eq. (5) change from

$$\begin{bmatrix} \vdots \\ a_{j,1} \\ a_{j,2} \\ \vdots \end{bmatrix} = \begin{bmatrix} \vdots & \vdots \\ S_{j,1}\theta_{j,1,1} & S_{j,2}\theta_{j,2,1} \\ S_{j,1}\theta_{j,1,2} & S_{j,2}\theta_{j,2,2} \\ \vdots & \vdots \end{bmatrix} \begin{bmatrix} v_1 \\ \vdots \\ v_{PY} \end{bmatrix}, \quad (7)$$

to incorporate the appropriate aliasing behavior in Eq. (6) by

$$\begin{bmatrix} \vdots \\ a_{j,1} \\ a_{j,2} \\ \vdots \end{bmatrix} = \begin{bmatrix} \vdots & \vdots & \vdots & \vdots \\ \dots S_{(j,1)-\Delta B_{j,1}}\theta_{j,1,1} \dots 0 \dots 0 \dots S_{(j,2)+\Delta B_{j,2}}\theta_{j,2,1} \dots \\ \dots 0 \dots S_{(j,1)+\Delta B_{j,1}}\theta_{j,1,2} \dots S_{(j,2)-\Delta B_{j,2}}\theta_{j,2,2} \dots 0 \dots \\ \vdots & \vdots & \vdots & \vdots \end{bmatrix} \begin{bmatrix} v_1 \\ \vdots \\ v_{PY} \end{bmatrix} \quad (8)$$

In Eq. (8), the voxel shift $B_{j,k}$ is used to describe which true voxel values in vector v are aliased into the acquired shot measurements $a_{j,s}$ by updating the column indices and coil sensitivities in the aliasing matrix of Eq. (7) accordingly. It is of note that the phase terms $\theta_{j,k,s}$ in Eq. (7) (drawn from SENSE reconstructed shot images) represent the motion characteristics associated with the acquired aliased voxels $a_{j,s}$, and therefore remain unchanged in Eq. (8), where they are simply applied to the updated sensitivities. For voxels near the edge of the FOV in which B is large, a wraparound condition is implemented in Eq. (8). As the compression of voxel measurements into a single voxel poses one of the greatest challenges in distortion correction, the RPG-MUSE aliasing matrix formulation in Eq. (6) represents a framework in which the aliasing of a compressed voxel a_j can be expanded to incorporate multiple sensitivities from more elaborate point spread functions.

RPG in 3D MUSE dMRI:

The RPG-MUSE framework in Eq. (6) can be extended to reconstruct 3D-ms-EPI data, where the imaging volume is acquired with both in-plane and through-plane phase encoding performed across thick (~ 10 mm) 3D slabs. With a total of N_Z slice-encoding planes (kz-planes), a 3D ms-EPI sequence encodes and acquires the same kz-plane in all slabs with ms-EPI before encoding and acquiring the next kz-plane. While such an acquisition offers a $N^{1/2}$ increase in SNR, the total acquisition time for a slab is $TR \cdot N_S \cdot N_Z$, which can be on the order of ~ 3 – 5 min per volume. As such, 3D ms-EPI data is prone to subject motion throughout the acquisition of both shots and kz-planes. The 3D-MUSE framework in Bruce et al., (2017) presents a means of accounting for the motion induced artifacts associated with 3D ms-EPI. In 3D-MUSE, a vector kz represents multi-shot coil measurements in the same voxel across all N_Z kz-planes in a slab,

$$kz = \Theta \circ \Gamma \circ \Omega Sz = Mz, \quad (9)$$

where z is a vector of true un-aliased voxel values in all slices that comprise the slab. To construct the aliasing matrix M in Eq. (9), Θ and Γ are matrices of motion induced phase variations between shots and kz-planes respectively, Ω is a 1D Fourier encoding matrix, S represents a block diagonal matrix of coil sensitivities from all slices in the slab, and $\circ$ represents a Hadamard (element-wise) product (a detailed derivation of the model in Eq. (9) can be found in Bruce et al., (2017)). If $N_Z=1$, the 3D-MUSE model in Eq. (9) reduces to the 2D-MUSE model in Eq. (2). While Eq. (9) represents the reconstruction of a single voxel in all kz-planes, this formulation can be easily expanded to reconstruct all voxels across both in-plane and through-plane PE dimensions at once. This is achieved by placing the aliasing matrix from Eq. (9) for each voxel j along the in-plane PE dimension, M_j , along the diagonal of a large block-diagonal matrix. When represented in this form, RPG can be incorporated into 3D-MUSE by applying the appropriate distortions, B , to the coil sensitivity component of the aliasing matrix in Eq. (9) through the same application used in Eq. (6). It is of note that RPG was only applied to the in-plane PE dimension in this report, but could ultimately be expanded to account for distortions along the through-plane PE dimension of thick slabs as well.

Methods:

RPG Acquisition Scheme:

Illustrated with two shots in Fig. 1a, each shot of a conventional ms-EPI scan acquires an array of spatial frequencies sub-sampled along the PE dimension (k_y). When the shots are interleaved together they form a complete sampling of k_y , and a full FOV image can be obtained through a simple 2D Fourier reconstruction. In RPG-MUSE, the ms-EPI shot acquisitions shown in Fig. 1b are acquired in the same interleaved fashion, only with even shots acquired with a reversed PE direction (from $+k_y$ to $-k_y$). When partial Fourier sampling is applied to RPG ms-EPI in Fig. 1c, spatial frequency measurements along $k_y < 0$ are partially sampled by odd shots and sampled in full by even shots, and vice versa for measurements along $k_y > 0$. The inter-shot phase errors induced by motion artifacts under

diffusion weighting gradients were corrected with the MUSE algorithm, as originally proposed in Chen et al., (2013).

Experiment:

A total of six dMRI data sets were acquired from healthy volunteers on a 3 T MR750 MRI scanner (General Electric, Waukesha, WI) using a 32-channel head coil (Magtron Inc., Jiangyin, China). Informed consent was obtained from all human subjects, in accordance with the Duke University Medical Center IRB. To investigate the degree to which geometric distortions are impacted by eddy currents in dMRI, the first four scans (see Table 1) were each acquired with one $b=0$ s/mm² volume followed by three $b=800$ s/mm² volumes in which diffusion gradients were individually applied along the three principal axes (G_x , G_y and G_z). In each of the final two scans (5 and 6), a DTI protocol was undertaken in which one $b=0$ s/mm² volume was acquired followed by 24 diffusion directions with $b=800$ s/mm². In all scans, RPG ms-EPI was used to acquire images over a 25.6 cm FOV in a 256×256 encoding matrix (65% partial Fourier sampling in ky), with forward PE (anterior/posterior) in all odd shots and reversed PE (posterior/anterior) in all even shots. Repetition times (TR) and echo times (TE) corresponding to each scan can be found in Table 1. Volumes acquired with 2D encoding (scans 1, 2, 5 and 6) were comprised of 96 axial slices (1 mm thick), while volumes acquired with 3D encoding (scans 3 and 4) were comprised of twelve 10 mm slabs with 12 kz-encoding planes, 20% oversampling along the z-dimension, and 20% overlap with adjacent slabs. Under these conditions, the same imaging volume was acquired at a 1.0 mm isotropic spatial resolution in all 2D and 3D scans. To account for bulk through-plane motion in 3D EPI, low resolution 2D navigator images were acquired in a 96×96 matrix at $k_z=0$ of each slab, immediately following the acquisition of each kz-plane in each shot.

Coil sensitivity profiles (free of distortion artifacts) were derived from a low-resolution calibration scan performed before each of the six high-resolution dMRI scans. To account for Nyquist ghosting artifacts, the phase shift between the odd and even lines in each shot was estimated from an integrated reference scan (with no phase encoding) acquired at the onset of each scan. Finally, to serve as a reference in evaluating the fidelity of each reconstruction, an anatomical T1 weighted volume was acquired.

Image Reconstruction and Processing:

The pipeline for reconstructing ms-EPI data with RPG-MUSE is presented in Fig. 2. First, the sub-sampled spatial frequencies from each shot are Fourier reconstructed into the reduced FOV aliased slice images (in each coil) shown in Fig. 2a. Using SENSE, the aliased shot images in Fig. 2a are then un-aliased into the full FOV magnitude images in Fig. 2b and (spatial filtered) phase maps in Fig. 2c. With shot-to-shot phase variations used to approximate inter-shot motion in MUSE, the phase terms for each shot s in Eq. (3), ϕ_s , are extracted directly from the low frequency phase maps for each shot in Fig. 2c. Representing acquisitions with both forward and reversed PE directions, the full FOV magnitude shot images in Fig. 2b are used to estimate the map of distortion characteristics, B , in Fig. 2d. In this report, B was derived for each dMRI volume through a nine-level iterative estimation using FSL's topup subroutine together with the standard b02b0.cnf configuration file

[<https://fsl.fmrib.ox.ac.uk/fsl/fslwiki/topup/>]. Finally, by inserting a column of values from the aliased coil images in Fig. 2a, the low frequency phase maps in Fig. 2c, and the B map in Fig. 2d into the RPG-MUSE model in Eq. (8), the resulting dMRI image in Fig. 2e is free of both motion and distortion artifacts.

To compare the effectiveness of static versus dynamic B maps, the RPG-MUSE reconstruction was performed for each dMRI volume twice. Modeled after the most commonly used field-mapping based correction techniques, the first reconstruction only used distortion maps estimated from the RPG shot images of the b_0 volumes, B_{b_0} , which represent static B_0 inhomogeneity. In the second reconstruction, each volume of a dMRI scan was reconstructed using dynamic distortion maps estimated from the RPG shot images specific to that volume, B_{DYN} , which represent both the B_0 inhomogeneity and the eddy current effects (in volumes with $b>0$).

For 3D-ms-EPI data sets, bulk through-plane motion artifacts were accounted for in each shot acquisition of each 3D slab using phase variations estimated from the additionally acquired low resolution navigator images. Individual slabs were reconstructed using the RPG adapted 3D-MUSE reconstruction before being stitched together by discarding the slices that overlapped with adjacent slabs. No additional slab stitching techniques such as PEN [Van et al., 2015] or slab profile normalization were utilized.

Upon observation, the processing pipeline presented in Fig. 2 requires each step to be performed sequentially. Although the SENSE and RPG-MUSE reconstructions are highly parallelizable and can be run with great efficiency using parallel computing techniques, the iterative estimation of B_{DYN} for each dMRI volume can be time consuming. However, based on the assumption that B_0 inhomogeneity remains constant throughout the scan, the estimation of B_{DYN} could be accelerated by treating B_{b_0} as an initial estimate.

To act as a baseline for comparisons with RPG-MUSE, data sets with forward PE and reversed PE were generated for scans 5 and 6 by separately combining the odd and even shots of each diffusion weighted volume through a MUSE reconstruction. Distortions in each volume of the forward/reversed PE data sets were corrected using the B_{b_0} maps of the corresponding scan. This resulted in four complete dMRI data sets for both scans 5 and 6: forward PE, reversed PE, and RPG-MUSE reconstructions using the static B_{b_0} map and dynamic B_{DYN} maps. It is important to note that data sets with forward/reversed PE are derived from half the number of shots as used in the RPG data sets, and thus a loss in SNR is expected. Finally, trace weighted images and fractional anisotropy (FA) maps were estimated using FSL [Woolrich et al., 2009; Smith et al., 2004; Jenkinson et al., 2012] for each data set associated with scans 5 and 6.

Results and Discussion:

Exploring eddy current effects in B distortion maps:

To assess the impact of eddy currents (from the diffusion-weighting gradients) on the distortion field, Fig. 3 presents maps of distortion along the PE dimension (anterior-posterior) between shots of RPG ms-EPI. In the left column, static B_0 distortions estimated

from b_0 data (B_{b0}) act as a baseline for comparison to data acquired with diffusion gradients applied left/right (B_{Gx} – second column), anterior/posterior (B_{Gy} – third column), and superior/inferior (B_{Gz} – fourth column). The B maps in Figs. 3a–3d were derived from 2D encoded data while the maps in Figs. 3e–3h were derived from data with 3D encoding. With phase encoding along ky , images acquired with forward PE will experience stretching when distortions are negative (blue) in the anterior half of the image and positive (red) in the posterior half, while images acquired with reversed PE will experience the opposite effects. As 2-shot ms-EPI has a prolonged TE, increased echo spacing and a sampling matrix twice as large than that of 4-shot, the distortions of 2-shot data (first row in each sub-figure of Fig. 3) are generally the most severe. When applying a diffusion gradient along the frequency encoding direction (left/right), perpendicular to PE, the B_{Gx} maps in Figs. 3b and 3f are very similar to those of B_{b0} , signifying minimal eddy current effects. By contrast, diffusion applied along the PE dimension results in strong negative eddy current induced distortions in the anterior region of the B_{Gy} maps in Figs. 3c and 3g, and similar behavior is noted when diffusion is applied in the through-plane dimension in the B_{Gz} maps in Figs. 3d and 3h. Thus, it is to be expected that dMRI volumes with gradients heavily weighted by either Gy or Gz will experience more severe distortions than would be characterized by the static B_{b0} map.

Advantages of RPG-MUSE in correcting both B_0 and eddy current induced distortions:

In recent DTI studies, it has become common practice to use multiple RPG b_0 volumes to estimate a static B_0 map used for removing distortions in dMRI data that is subsequently acquired with either forward or reversed PE. While this technique only marginally increases the overall scan time, the correction of distortions in data with a single PE dimension is often incomplete. Presented in Figs. 4a and 4b are images with forward and reversed PE (drawn from 2-shot RPG ms-EPI data) that have diffusion weighting applied along PE. When applying a map of static distortions, such as the B_{b0} map in Fig. 3a, the corrected forward PE image in Fig. 4d exhibits residual stretching (compared with an anatomical overlay) while signal pileup due to compression remains in the reversed PE image in Fig. 4e. By contrast, combining data from both forward and reversed PE data together with the corresponding B map in Fig. 4c, RPG-MUSE can completely remove distortions and signal pileup induced by both static B_0 inhomogeneities and dynamic eddy currents, as shown in the reconstructed image in Fig. 4f.

Although eddy current effects are specific to individual diffusion volumes [Jezzard et al., 1998; Bodammer et al., 2004], it remains common for distortions in dMRI data acquired with RPG to be corrected using maps of static B_0 inhomogeneity [Smith et al., 2004, Irfanoglu et al., 2015]. To further confirm the advantage of dynamic distortion correction in RPG-MUSE, Fig. 5 presents images of dMRI data with a gradient applied along the PE dimension reconstructed by RPG-MUSE using both static B_{b0} and dynamic B_{Gy} maps. Upon observation, images reconstructed using static B_{b0} maps (left column) exhibit both residual distortions in the anterior brain (green arrows) and blurring artifacts throughout the brain (yellow arrows), while images reconstructed with dynamic B_{Gy} maps (center column) exhibit high spatial fidelity and appear largely distortion free. This implies that the effects of eddy currents, shown through the difference between the two reconstructions

(right column), render static B_0 distortion maps inadequate for appropriately aligning the stretched and compressed regions of dMRI data acquired with RPG ms-EPI. Despite an increase in scan time (see table 1), the additional SNR achieved through 3D RPG ms-EPI acquisitions in Figs. 5c and 5d further enhances the improved fidelity of RPG-MUSE observed in the corresponding 2D RPG ms-EPI acquisitions in Figs. 5a and 5b.

Improved accuracy in DTI measures using RPG-MUSE:

To further demonstrate the advantages of RPG-MUSE, Fig. 6 presents trace weighted images and color FA maps estimated from 24-direction DTI data (scans 5 and 6) with forward PE (odd shots), reversed PE (even shots), an RPG-MUSE reconstruction assuming only static B_0 inhomogeneity in all DTI volumes (B_{b0}), and an RPG-MUSE reconstruction assuming both B_0 and dynamic eddy current induced distortions in each DTI volume (B_{DYN}). As distortion artifacts were corrected in each DTI volume individually, a trace weighted combination of all diffusion volumes provides an overall evaluation of the correction process. Despite the application of B_{b0} , the trace weighted images of 2-shot dMRI data with forward PE in Fig. 6a exhibit stretching in both anterior and posterior brain regions (purple arrows), and a section of gray matter along the anterior most edge appears to be missing. This behavior is matched by the corresponding FA map in Fig. 6a, where voxels with diffusion along anterior/posterior (green) are stretched to the anterior of the brain (yellow arrows). Conversely in 2-shot dMRI data with reversed PE in Fig. 6b, voxels along the anterior and posterior edges appear compressed, with apparent blurring and residual signal pileup in these regions. When only accounting for static B_0 inhomogeneity through RPG-MUSE using B_{b0} , the distortions noted in Figs. 6a and 6b are combined in Fig. 6c, where gray/white matter boundaries appear blurred throughout the brain in the trace weighted images, and anatomical accuracy in the FA map is lost in the anterior brain. The distortions in Figs. 6a–6c demonstrate that the static B_{b0} maps are inadequate for correcting the effects of eddy currents, resulting in a misalignment of voxels from one volume to the next. It is of note that susceptibility differences along air/tissue interfaces (such as the sinus near the anterior brain) result in inhomogeneities observed in the static B_0 map. As such, the blurring and distortion artifacts noted along tissue boundaries within the brain as well as along the posterior edge in Figs. 6a–6c are predominantly induced by eddy currents. When using maps of dynamic distortions, B_{DYN} , to account for both B_0 and eddy current effects in each DTI volume with RPG-MUSE, trace images and FA maps exhibit high spatial fidelity with distinct boundaries between gray and white matter. Although images with 4-shot ms-EPI in Figs. 6e–6g exhibit reduced distortions when compared with their 2-shot counterparts in Figs. 6a–6c, distortions are apparent along the anterior and posterior edges and blurring artifacts remain along gray/white matter boundaries. Similar to Fig. 6d, trace weighted images and FA maps in Fig. 6h exhibit high spatial fidelity with virtually no residual distortion artifacts when derived from an RPG-MUSE reconstruction of 4-shot data using dynamic B_{DYN} maps for each DTI volume. It is important to note that the anatomical specificity in Figs. 6d and 6h were achieved with no additional post-processing steps (e.g. image registration, bias-field correction, etc.) after RPG-MUSE, indicating that the reconstruction process alone removed almost all motion and distortion artifacts.

Conclusions:

In recent years, geometric distortions in dMRI data have been mitigated through field-map based approaches or multiple b_0 images acquired with RPG. The emergence of high-resolution dMRI using multi-shot EPI through MUSE offers a unique opportunity to fully adopt RPG to remove both B_0 and eddy current induced distortions. Using RPG-MUSE, there is no increase in acquisition time over that of a conventional MUSE scan, and the RPG sampling of dMRI data enables dynamic B maps to be estimated for each diffusion volume individually. Through 2 and 4-shot RPG ms-EPI data acquired with both 2D and 3D encoding, it has been demonstrated that the RPG-MUSE model simultaneously accounts for both inter-shot motion and distortion artifacts as part of the reconstruction. Being specific to each dMRI volume, the dynamic B_{DTN} maps used in RPG-MUSE automatically incorporate the effects of both B_0 inhomogeneity and eddy currents, which is of particular importance in the presence of either strong diffusion gradients and/or magnetic fields. The RPG-MUSE model presented here provides an expandable framework on which one can incorporate either additional corrections (such as through-plane distortions associated with 3D encoding) or more complex point spread functions that employ a weighted combination of neighboring voxels to better redistribute signal pileup in voxels with severe compression. To further account for eddy current effects, dynamic coil sensitivity profiles can be derived from full FOV low-resolution navigator images acquired following the echo train of each shot in each slice of an ms-EPI acquisition [Taviani et al., 2018]. The proposed framework can also be expanded to incorporate both 2D and 3D multi-band acquisitions to simultaneously reduce both distortions and data acquisition times. Through RPG-MUSE, minimal changes to existing acquisition techniques are required to achieve dMRI data with high spatial-resolution, high spatial fidelity, and improved accuracy in diffusion measures.

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Figure 1.
Sampling trajectory of 2-shot interleaved EPI with a) forward phase encoding, b) RPG, and
c) RPG with partial Fourier sampling.

Figure 2.

Processing pipeline for RPG-MUSE. In a) multi-coil aliased images from forward PE (odd shots) and reversed PE (even shots) are reconstructed with SENSE into full FOV b) magnitude and c) phase shot images. Together with the aliased coil images and phase maps, d) a B map estimated from magnitude images are inserted into the RPG-MUSE reconstruction to produce e) the final image.

Figure 3.

Maps of distortion along PE dimension estimated from 2D RPG ms-EPI diffusion data with a) no diffusion weighting ($b=0$), b) left/right DW gradient (G_x), c) anterior/posterior DW gradient (G_y), and d) inferior/superior DW gradient (G_z), with corresponding maps for 3D RPG ms-EPI data in e)-h). Blue areas indicate voxel shift toward anterior if PE=anterior/posterior and toward posterior if PE=posterior/anterior. Dashed lines in axial/Sagittal slice images indicate location of corresponding Sagittal/axial slice.

Figure 4.

With diffusion along the PE dimension (anterior/posterior), stretching in the anterior brain of a) forward PE images and compression in b) reversed PE images is reflected in c) the distortion map, ΔB . Applying a map derived from b_0 data, B_{b_0} , the corrected d) forward PE and e) reversed PE images exhibit residual distortions and signal pileup when compared with an f) RPG-MUSE reconstructed image. Points for comparison along the anterior edge of the brain and along tissue boundaries within the brain are highlighted by green and yellow arrows respectively.

Figure 5. RPG-MUSE reconstructed images with diffusion applied along anterior/posterior (G_y) acquired with 2 and 4-shot RPG ms-EPI data using 2D (a-b) and 3D (c-d) encoding exhibit residual eddy current induced distortions when reconstructed using a map of static B_0 distortions (B_{b0}) compared with a dynamic map of B_0 and eddy current induced distortions (B_{Gy}). Points for comparison along the anterior edge of the brain and along tissue boundaries within the brain are highlighted by green and yellow arrows respectively.

Figure 6.

Trace weighted images and FA maps derived from 24-direction diffusion data with 2-shot EPI (a-d) and 4-shot EPI (e-h). Maps of static B_0 inhomogeneity (B_{b0}) were used to correct distortions in forward PE (a, e) and reversed PE (b, f) data post-reconstruction. Distortions in RPG ms-EPI data were corrected during RPG-MUSE reconstruction by assuming only static B_{b0} distortions in all DTI volumes (c, g) and by assuming dynamic B_0 +eddy current induced distortions (B_{DYN}) in each DTI volume individually (d, h). Comparable points of interest are highlighted in trace weighted images and FA maps with pink and yellow arrows respectively.

Table 1:

Number of shots, TR, TE, encoding dimensions, scan time and dMRI protocols used in each of six scans.

Scan	No. shots	TR (s)	TE (ms)	Encoding	Scan time (min)	dMRI protocol
1	2	15	54	2D	2:00	$1 \times b=0, 3 \times b=800 \text{ s/mm}^2$ (G_x, G_y, G_z)
2	4	11	66		2:56	
3	2	3	70	3D	4:48	
4	4	3	57		9:36	
5	2	15	54	2D	12:30	$1 \times b=0, 24 \times b=800 \text{ s/mm}^2$ DTI directions
6	4	11	66		18:20	