**Purpose:** $B_1^+$ and $T_1$ corrections and dynamic multicoil shimming approaches were proposed to improve the fidelity of high-isotropic-resolution generalized slice-dithered enhanced resolution (gSlider) diffusion imaging.

**Methods:** An extended reconstruction incorporating $B_1^+$ inhomogeneity and $T_1$ recovery information was developed to mitigate slab-boundary artifacts in short-repetition time (TR) gSlider acquisitions. Slab-by-slab dynamic $B_0$ shimming using a multicoil integrated $\Delta B_0/Rx$ shim array and high in-plane acceleration ($R_{\text{inplane}} = 4$) achieved with virtual-coil GRAPPA were also incorporated into a 1-mm isotropic resolution gSlider acquisition/reconstruction framework to achieve a significant reduction in geometric distortion compared to single-shot echo planar imaging (EPI).

**Results:** The slab-boundary artifacts were alleviated by the proposed $B_1^+$ and $T_1$ corrections compared to the standard gSlider reconstruction pipeline for short-TR acquisitions. Dynamic shimming provided $>50\%$ reduction in geometric distortion compared to conventional global second-order shimming. One-millimeter isotropic resolution diffusion data show that the typically problematic temporal and frontal lobes of the brain can be imaged with high geometric fidelity using dynamic shimming.

**Conclusions:** The proposed $B_1^+$ and $T_1$ corrections and local-field control substantially improved the fidelity of high-isotropic-resolution diffusion imaging, with reduced slab-boundary artifacts and geometric distortion compared to conventional gSlider acquisition and reconstruction. This enabled high-fidelity whole-brain 1-mm isotropic diffusion imaging with 64 diffusion directions in 20 min using a 3T clinical scanner.

**KEY WORDS**

$B_1^+$ inhomogeneity, diffusion-weighted imaging, gSlider, shim array
High-resolution diffusion-weighted imaging (DWI) with echo-planar imaging (EPI) acquisition technique is a powerful tool for many neuroscientific and clinical applications. Single-shot EPI\(^1\) is one of the most commonly used methods in DWI because of its rapid encoding ability; however, its readout does not lend itself to high in-plane resolution imaging because of severe geometric distortion and \(T_2^*\)-related blurring artifacts, which are difficult to correct in post-processing, particularly where voxel pile-up occurs.\(^2\) Multishot EPI\(^3-5\) is a promising approach to improve the geometric fidelity of DWI and achieve high in-plane resolution with low distortion and \(T_2^*\) blurring. However, because of the prolonged scan time, shot-to-shot phase variations, and potential patient motion, multishot EPI continues to be a challenge in DWI. To mitigate these problems, previous studies combined multishot approaches with parallel imaging,\(^6,7\) sparse or low-rank models,\(^8-10\) joint reconstruction,\(^11-15\) and simultaneous multislice (SMS)\(^16-18\) to accelerate the acquisition and correct for shot-to-shot phase variations.

Another approach to mitigate distortion is to generate a compensating \(B_0\) magnetic field to counteract the off-resonance effects. Conventional scanners are equipped with first-order (the linear gradients) and static second-order spherical harmonic shim coils, which generate a spatial magnetic profile to compensate the \(B_0\) field over the target volume.\(^19,20\) Local multicoil (MC) shimming using small shim coils patterned around the imaged object has been introduced as a convenient way to provide higher-order \(B_0\) shimming without the need to modify the MRI scanner substantially.\(^21-24\)

Compared to second-order shimming, MC shim arrays have been shown to provide improved \(B_0\) homogeneity to improve EPI acquisitions for structural and functional MRI (fMRI) applications,\(^25,26\) and magnetic resonance spectroscopic imaging acquisition for spectroscopy studies.\(^27\) The ability of MC arrays to switch their shim currents rapidly without causing image artifacts allows the \(B_0\) shim to be optimized on a slice-by-slice basis, providing further gains in \(B_0\) homogeneity and mitigation of EPI geometric distortion.\(^21\)

Improving SNR efficiency is critical for achieving high-isotropic-resolution DWI. Three-dimensional multislab DWI has emerged as a promising strategy to enhance the SNR in such acquisitions.\(^28-32\) However, shot-to-shot phase variations and slab-boundary artifacts are key challenges for efficient sampling of whole-brain high-resolution DWI with this technique,\(^33\) where a number of effective techniques have been developed to mitigate these issues.\(^30,33,34\)

Another promising approach for high-SNR-efficiency, high-resolution DWI is the Generalized SLIce Dithered Enhanced Resolution (gSlider) method, which is a simultaneous multislab (SMSb) acquisition technique with self-navigated RF slab encoding, which has been demonstrated for motion-robust, high-resolution DWI.\(^35,36\) Here, the simultaneous multislabs are acquired together using the blipped-CAIPI acquisition scheme\(^37\) and separated through parallel imaging. With gSlider, a large number of slices can be acquired together per EPI-encoding (e.g. 10 simultaneous slices using gSlider RF encoding of \(5\times\) and multiband factor of 2\(\times\)) to achieve a short TR for high-resolution volumetric DWI and provide high-SNR efficiency. The encoding within each slab is then performed through sequential RF slab-encoding acquisitions and combined using super-resolution reconstruction to create high-resolution slices. The gSlider RF encodings are designed to provide high signal in each of the RF-encoded acquisitions for robust estimation and removal of shot-to-shot phase corruptions, removing the need for the additional lengthy navigator module per TR in conventional 3D multislab EPI acquisition. With such RF encoding, there is a small reduction in the orthogonality of the encoding bases (i.e. compare to \(k_z\) encoding), which translate to a minor reduction in the SNR gain from \(\sqrt{5}\) to \(\sqrt{4.6}\) for \(5\times\) encoding.\(^35\) The gSlider method is also robust to bulk motion using designed RF-encoding basis and spatially varying regularizations.\(^36\) With this approach, the effects of subject motion can be estimated and mitigated every TR/2 without a motion navigator. As such, the motion sensitivity timeframe is less than 2 s rather than multiple tens of seconds in 3D multislab EPI. Finally, with this approach, the ratio of slab thickness relative to slice resolution can be kept small (e.g. 5 for gSlider 5\(\times\)), allowing for sharp slab selective excitation with reduced slab boundary issues from partial excitation compared to typical 3D multislab DWI and removing the need for slab oversampling. This has enabled high-quality submillimeter gSlider DWI without slab boundary correction at TR ~4.5 s or above.\(^35,36,38\) However, at shorter TRs slab boundary artifacts can remain an issue, with partial \(T_1\) recovery in adjacent slabs causing striping artifacts. Therefore, a correction approach similar to nonlinear inversion for slab profile encoding\(^33\) is needed to achieve high-quality reconstruction.

In this work, we developed approaches to improve the fidelity of gSlider: first, by making it robust to slab-boundary artifacts, and second, by reducing geometric distortion from \(B_0\) inhomogeneity. To mitigate slab-boundary artifacts in short-TR acquisitions, dictionary-based \(B_1^+\) inhomogeneity and \(T_1\) recovery corrections were incorporated into gSlider reconstruction. To mitigate image distortions, a 32-channel integrated \(\Delta B_1/Rx\) array (AC/DC coil)\(^21\) was utilized for slab-by-slab shimming, to reduce \(B_0\) inhomogeneity by >50% as compared to static second-order shimming. This dual-purpose coil array provides both high spatial-order \(B_0\) field control as well as good parallel imaging capability. This is then combined with high in-plane acceleration (\(R_{\text{plane}} = 4\)) and virtual conjugate coil GRAPPA reconstruction,\(^11,39\) to achieve 8-fold
to 11-fold total geometric distortion reduction in single-shot gSlider-EPI. We demonstrate that the proposed method can achieve high-fidelity whole-brain 1-mm isotropic DWI with 64 diffusion directions in ~20 min on a clinical 3T scanner.

2 | METHODS

2.1 Pulse sequence design and slab-optimized dynamic shimming

Figure 1A shows the sequence diagram of gSlider, where two external triggers are added in each TR to enable slab-by-slab $B_0$ shimming with the AC/DC coil (Figure 1B). To prevent poor performance in whole-brain fat suppression from large out-of-slab $B_0$ inhomogeneity, the slab-by-slab shimming was turned off during fat saturation. In each TR, an additional $k_y$ blip was added to shift k-space and create more unique source points for improved virtual conjugate-coil GRAPPA reconstruction at high accelerations in DWI as outlined in Refs. 11 and 40.

To generate a sharp slab-encoding performance, the Shinnar-Le Roux algorithm$^{41}$ was used to design the gSlider excitation RF pulses. The time-bandwidth product of the 90° gSlider RF excitation pulses was 12, with a pulse duration of 11 ms. Figure 2A,B show the waveforms of five gSlider-encoding RF pulses that were used and their corresponding Bloch simulated slab profiles. After slab encoding, a standard Shinnar-Le Roux spin-echo (SE) refocusing pulse was applied without gSlider-encoding. The time-bandwidth product of refocusing pulse was 8 and the duration of the refocusing pulse was 7.3 ms. The blue lines in Figure 2B show the subslab profiles of gSlider-encoding after the SE refocusing pulse. To reduce the peak power of the RF pulses, a VERSE method$^{42}$ was applied to both gSlider excitation RF pulses and SE refocusing pulse.

To implement slab-optimized shimming, a low-resolution $B_0$ field map with conventional global shims applied was acquired using a vendor-provided two-TE gradient echo field mapping sequence. The field maps were registered to the thin-slab gSlider images and then masked using FMRIB...
Software Library (FSL) BET and phase unwrapped using FSL PRELUDE. The optimal DC shim currents in each channel of the AC/DC coil were then computed on a slabwise basis using a previously acquired calibration $B_0$ map basis set for the array. The details of the constrained optimization algorithm can be found in reference 21. For SMSb acquisition, the shims were jointly optimized over the two simultaneously acquired slabs. To match the geometric distortions between slab-collapsed EPI data and fully sampled reference data, the slab ordering of shimming and the currents used in the individual slab of reference data are the same as the corresponding slab group of the SMSb data. To prevent artifacts, the GRAPPA and SMS calibration scan data for each slab were shimmed with the same MC shim fields as the DWI acquisition.

2.2 $B_1^+$ and $T_1$ corrections for robust gSlider reconstruction

To eliminate shot-to-shot background phase variations in the acquired diffusion data, real-valued diffusion processing was applied. The gSlider reconstruction was then performed to obtain high slice-resolution data, using a forward model based on the Bloch simulated slab profiles of the gSlider encodings. Pseudoinverse with Tikhonov regularization was used:

$$X = (A^T A + \lambda I)^{-1} A^T b,$$

where $b$ (matrix size: $N_{slab} \times N_{rf-encoding}$) is the concatenation of acquired thin-slab data at a given in-plane spatial location, $X$ (matrix size: $N_{slice} \times 1$) is the corresponding super-resolution reconstruction, $A$ [matrix size: $(N_{slab} \times N_{rf-encoding}) \times (N_{slab} \times N_{rf-encoding})$] is the RF-encoding matrix that contains the subslab profiles simulated from the Bloch equations, and $\lambda$ is a Tikhonov regularization parameter. In our previous work, the same slab profiles ($M_{xy}$) of the RF encodings were used for the reconstruction at all spatial locations contained within the A encoding matrix. However, this does not account for potential spatial variations in $M_{xy}$ due to $B_1^+$ inhomogeneity. Furthermore, the initial longitude magnetization ($M_z$) in the Bloch simulation was set to 1, which ignored incomplete $T_1$ recovery. This can be particularly problematic in the adja-
cently partial excitation regions of a nonideal RF excitation after TR/2 slab-interleaved acquisition at short TRs. These imperfections could cause slab-boundary artifacts due to $B_1^+$ variations.

To mitigate slab-boundary artifacts, RF-encoding imperfections due to $B_1^+$ inhomogeneity and incomplete $T_1$ recovery were estimated and incorporated into the RF-encoding matrix $A$ of the gSlider reconstruction in Equation (1). Figure 3A shows the flowchart of $B_1^+$ inhomogeneity correction, where RF-encoding profiles at a range of discretized $B_1^+$ values ([0.70:0.05:1.30], ±30% $B_1^+$ variations) are simulated by using Bloch equation, which enabled the creation of a dictionary of RF-encoding matrices with different $B_1^+$ variations. The proposed $B_1^+$ correction method is a 3D “voxel-by-voxel” correction scheme based on a 3D discretized $B_1^+$ map.

FIGURE 3  (A) Slab profiles ($M_{xy}$) of an RF-encoded gSlider pulse with $B_1^+$ inhomogeneity and (B) longitude magnetizations ($M_z$) of the adjacent slab before and after repetition time (TR/2) recovery. (C) The flowchart of $B_1^+$ correction using the prescan $B_1^+$ maps.
For each spatial location, the corresponding RF-encoding matrix from the dictionary was selected on the basis of the corresponding $B_1^+$ value (shown in Figure 3C) and thereby spatially varying $B_1^+$ inhomogeneity is corrected by voxel-by-voxel correction.

For incomplete $T_1$ recovery, nonideal slab profiles of RF encodings can cause partial excitations in adjacent slabs that are not fully recovered in slab-interleaved acquisitions with short TRs. This effect was also modeled by adding partial recovery initial longitudinal magnetizations $M_0$ into the Bloch simulation of the RF encodings (assuming average $T_1 = 1000$ ms in the brain), thereby incorporating them into the encoding matrix. Figure 3B shows the partial $M_0$ recovery from adjacent slabs excitations before and after TR/2 longitudinal relaxation at various $B_1^+$ excitation levels.

2.3 | Data acquisition

All in vivo measurements were performed on a 3T scanner (MAGNETOM Prisma, Siemens Healthineers, Erlangen, Germany) with a custom 32-channel AC/DC receiver array with added $B_0$ shim capability. To assess the improvements provided by slab-optimized shimming, gSlider-EPI with five slab encodings and the corresponding $B_0$ field maps were acquired. The imaging parameters for gSlider data were field of view $220 \times 220 \times 170$ mm$^3$, 34 thin slabs (5-mm slab encoding), TR/TE = 5100/77 ms, and echo spacing = 0.93 ms. To accentuate changes in geometric distortion, data were acquired using both anterior-to-posterior and posterior-to-anterior phase encodings at different in-plane accelerations ($R_{\text{inplane}} = 1$ and $R_{\text{inplane}} = 4$), with and without slab-optimized shimming. The $B_0$ field maps were acquired using two-echo gradient echo with 2.5-mm slice thickness and 100% gap. The slice resolution including gap matched the 5-mm gSlider slab encoding. TR/TE = 5100/77 ms, and echo spacing = 0.93 ms. To validate the shim performance of SMSb imaging, the same gSlider data were acquired with a multiband (MB) factor of 2 and compared with non-SMSb gSlider data. A matching $T_2$ turbo spin-echo ($T_2$-TSE) data was also acquired as a distortion-free reference.

Whole-brain 1-mm isotropic resolution diffusion imaging data were also acquired with gSlider-EPI and dynamic MC shimming. The protocol used was field of view $220 \times 220 \times 170$ mm$^3$, $R_{\text{inplane}} \times MB \times gSlider = 4 \times 2 \times 5$, 34 thin-slabs (5-mm slab encoding), $b = 1000$ s/mm$^2$ with 64 diffusion directions and four interleaved $b = 0$ s/mm$^2$, TR/TE = 3500/86 ms. The total acquisition time is ~20 min.

To correct the $B_1^+$ effects in gSlider data, a field of view-matched $B_1^+$ map was obtained by using a Turbo-FLASH scan with preconditioning RF pulses. The in-plane resolution is $3.4 \times 3.4$ mm$^2$ with 2.5-mm slice thickness and 100% gap. The slice resolution including gap matched the 5-mm gSlider slab encoding.

2.4 | Reconstruction and postprocessing

To enable higher in-plane acceleration compared to conventional slice-GRAPPA and in-plane-GRAPPA reconstruction and further reduce the geometric distortions, virtual conjugate-coil GRAPPA with phase matching was used to achieve high-fidelity reconstruction for high acceleration factors. To achieve faster GRAPPA reconstruction, singular value decomposition coil compression was applied to compress the 32 channel coils to 20 channels. After GRAPPA reconstruction, the five RF-encoded volumes of each diffusion direction were then combined to create thin-slice data, using gSlider reconstruction with and without the proposed modified RF-encoding matrix. The RF-encoding matrix was generated using the SLR RF-pulse design and Bloch simulation toolbox (https://vuiis.vumc.org/~grissowa/software.html). The virtual conjugate-coil GRAPPA and gSlider reconstruction algorithms were implemented in MATLAB R2014a (MathWorks, Inc., Natick, MA). The reconstructed data were then corrected for motion and eddy-current distortion using the “eddy_correct” function from the FMRIB Software Library (FSL, https://fsl.fmrib.ox.ac.uk/fsl/fslwiki/). The diffusion tensor model was fitted using FSL’s “dift” function to obtain the fractional anisotropy maps and the primary eigenvectors.

3 | RESULTS

Figure 4 shows the results of $B_1^+$ and $T_1$ corrections in gSlider reconstruction at TR of 3.5 s. The slab-boundary artifacts shown in the sagittal and coronal views of a diffusion-weighted volume are well mitigated by incorporating $T_1$ correction into the $B_1^+$ corrected processing compared to the standard gSlider reconstruction without corrections and with $B_1^+$ correction only, which demonstrates the utility of the proposed $B_1^+$ and $T_1$ corrections in gSlider reconstruction in a short-TR acquisition.

Figure 5 compares image distortion with and without dynamic slab-optimized MC $B_0$ shimming. The green arrows highlight the $B_0$ distortion that was alleviated, with slabby-slab shimming achieving >50% reduction of standard deviation (std) in $\Delta B_0$ across the slab when compared to baseline global second-order shimming. Dynamic MC shimming was then combined with in-plane acceleration ($R_{\text{inplane}} = 4$) to achieve an 8× to 11× total reduction in $\Delta B_0$ distortion (depending on the slab), yielding images with outlines (red outlines in Figure 5) closely matching that of the reference $T_2$-TSE images.

Figure 6 shows the $B_0$ field maps obtained from global second-order shim, MB-2, and MB-1 dynamic MC shimming. Compared to MB-1 slab-optimized shimming, the MB-2 case applies shims simultaneously to two distant slabs, providing similar standard deviations of $\Delta B_0$ variations to the MB-1
case, while the out-of-slab regions were unconstrained and allowed to have a poor shim.

Figure 7 shows the dynamic MC shimming results of two representative slabs for MB-1 and MB-2 with anterior-to-posterior and posterior-to-anterior phase-encoding directions. Compared to the reference images, the contours of both the MB-1 and MB-2 images closely match those of the $T_2$-TSE images, which demonstrates that the dynamic MC shimming of MB-2 achieved a similar performance to MB-1 slab-optimized dynamic MC shimming. This demonstrates that the 32-channel AC/DC coil has enough degrees of freedom to control the $B_0$ field in two spatially separated slices at the same time with minimal loss of performance.

Figure 8 shows the averaged DW images from 64 diffusion-encoding directions (Figure 8A) and directionally encoded color fractional anisotropy maps (Figure 8B) of the 1-mm isotropic gSlider diffusion data. High-quality results are shown in Figure 8C, with minimal geometric distortions in the typically problematic temporal and frontal lobes. The high-quality results depict the primary eigenvectors from DTI with high fidelity (Figure 8C) that are beneficial for mapping structural connectivity using diffusion tractography and mapping cortical diffusion patterns in these problematic regions.

4 | DISCUSSION

In this work, we developed synergistic approaches to improve gSlider acquisitions where i) reconstruction with $B_1^+$ and $T_1$ corrections effectively mitigate slab-boundary artifacts in short-TR acquisitions, and ii) dynamic MC $B_0$ shimming and high in-plane acceleration achieve an 8-fold to 11-fold reduction in $B_0$ distortion. The results demonstrate that the proposed corrections and local-field control can together achieve high-quality, high-fidelity DWI with 1-mm isotropic resolution in 20 minutes on a 3T clinical scanner.

To minimize the slab-boundary artifacts, there is a trade-off between the design of slab thickness of the RF refocusing pulse and the sensitivity of gSlider to $B_1^+$ inhomogeneity. Using a slightly thicker refocusing slab than the target slab thickness can improve the refocusing performance at the edges of the slab, since the refocusing slab’s transition bands are moved away from the target slab region. This is particularly useful in the presence of $B_1^+$ inhomogeneity, where the refocusing performance at the edge of the refocusing slab can degrade significantly (much more than in the excitation pulse). This thicker refocusing approach was employed in our previous work to provide robustness to $B_1^+$ inhomogeneity, where the gSlider acquisition was performed with a TR of ~4.5 s. However, the partial excitations in adjacent slabs induced by this broader refocusing pulse can cause striping artifacts for short-TR acquisitions due to the partial recovery. For this work, we employed matching excitation and refocusing slab thickness to avoid a large $T_1$ recovery issue while correcting for the increased $B_1^+$ inhomogeneity effect through a modified reconstruction. The proposed method corrects both $B_1^+$ inhomogeneity and partial $T_1$ recovery jointly by using a dictionary of RF-encoding matrices, which has been demonstrated to improve the fidelity of gSlider data for short-TR acquisitions.

In this work, a TR of 3.5 s was chosen for the acquisition to balance the trade-off between SNR efficiency versus spin-history and motion sensitivity issues. With our whole-brain 1-mm isotropic gSlider acquisition, the TR can be further shortened to 2.5 s to provide higher SNR efficiency and shorter total acquisition time. However, at such TR, with a shorter $T_1$ recovery period, slab-boundary artifacts and...
motion sensitivity will be more severe, requiring the development of a more advanced correction approach such as in reference 33 Employing a TR of 3.5 s markedly reduced these issues, while still achieving an SNR-efficient level that is at 80% to 95% of the optimal value for white matter and gray matter tissues.48

There are some limitations in the proposed $B_1^+$ and $T_1$ corrections. First, in the Bloch simulation and dictionary generation, the continuous $B_1^+$ variations were discretized and sorted into several discretized bins, which may not reflect the accurate slab profile at a given position. However, since the spatial $B_1^+$ values vary slowly and smoothly, the discretization of

**FIGURE 5** The comparisons of the $B_0$ field maps and multiband (MB-1) echo planar imaging (EPI) distortions with anterior-to-posterior (AP) and posterior-to-anterior (PA) phase encodings, with and without dynamic multicoil $B_0$ shimming. The shimming is then combined with $R_{\text{aplane}} = 4$ acceleration to achieve ~10-fold reduction in EPI distortion, putting the anterior-to-posterior and posterior-to-anterior images into closer alignment. The resulting low-distortion EPI slices resemble the distortion-free $T_2$-turbo spin-echo (TSE) reference slices (see red brain mask outline).
FIGURE 6  Compared to baseline second-order spherical harmonic global shimming, dynamic multicoil (MC) shimming for multiband (MB-2) acquisitions jointly shims two slabs at the same time, while out-of-slab regions are unconstrained and are allowed to have poor shims. The shim result of MB-2 is performance close to MB-1 slab-optimized shimming.

FIGURE 7  Comparison between multiband (MB-2) and MB-1. The MB-2 shimming performance closely resembles that of the MB-1 slab-optimized shimming. The red outlines show the contours of the $T_2$-turbo spin-echo (TSE) reference brain mask.
64-direction averaged DWI maps, $b=1000$ s/mm$^2$

**FIGURE 8** (A) Averaged diffusion-weighted images and (B) directionallyencoded color fractional anisotropy (FA) maps of the 1-mm isotropic diffusion-weighted data with 64 directions obtained in ~20 min using the proposed correction and dynamic shimming framework. (C) The primary eigenvectors from diffusion tensor imaging in typically problematic temporal and frontal lobes of the brain were more accurately depicted by synergistically combining dynamic $B_0$ shimming with parallel imaging acceleration. The primary eigenvectors were color-encoded (red: left-right, green: anterior-posterior, blue: superior-inferior) and overlaid on FA maps.

$B_1^+$ maps should not affect the $B_1^+$ correction unduly. Second, for $T_1$ correction, the $T_1$ value used in the study is 1000 ms, which is taken to be the average value across the gray-matter and white-matter tissues of the brain.$^{49}$ However, this assumption does not accurately reflect the complicated biochemical environment in the brain, leading to residual slab-boundary artifacts due to partial volume effects caused by high-$T_1$ compartments such as blood and cerebrospinal fluid. Nevertheless, these residual artifacts have minimal impact on diffusion-weighted images because the fluid is almost completely attenuated by the diffusion encodings.

Slab-by-slab dynamic shim updating was used for gSlider acquisition. With a 32-channel custom-built AC/DC coil, the $B_0$ inhomogeneity was reduced by $>50\%$, thereby significantly mitigating the geometric distortion in EPI slices. Combined with parallel imaging with high in-plane acceleration ($R_{\text{inplane}} = 4$), the geometric distortion was further reduced. Consequently, the resultant images closely matched the TSE reference across all brain regions. Furthermore, we extended our slab-by-slab shimming scheme to slab-group-by-slab-group shimming for SMSb acquisitions that shim two distant slabs simultaneously. Compared to postprocessing-based EPI distortion correction methods$^{50-52}$ that require additional data (e.g. reversed phase-encoding images) and processing time, our proposed dynamic MC shimming method only needs a fast low-resolution $B_0$ scan for computing the optimal...
shim currents and reducing the geometric distortion during the acquisition. Furthermore, for high-resolution DWI, the dynamic MC shimming method reduces geometric distortion at its source, whereas the postprocessing methods have limited ability to correct accurately for severe voxel pile-up.

The level of distortion reduction from our approach at 10× to 12× is similar to that of current state-of-the-art multishot EPI approaches, such as readout-segmented EPI. However, for high-resolution imaging, there will still be some remaining distortion even with such an approach. For example, for the 1-mm resolution case (the field of view is 220 mm and the echo spacing is 0.93 ms), the distortion level at large susceptibility regions at for example 50 Hz off-resonance would be 0.91 voxel with dynamic shim 2.8× reduction and in-plane 4× acceleration (the distorted voxels in this region can be calculated as 50 Hz/[1000/(220*0.93ms)*(2.8*4)] = 0.91 voxel). In applications where geometric fidelity is of high priority, our acquisition method can also be combined with the dual anterior-to-posterior and posterior-to-anterior acquisition scheme, along with postprocessing correction, such as the top-up approach. In such a case, the distortion level to be corrected would be much smaller with our method, allowing the top-up correction to work robustly.

Dynamic MC shimming not only is applicable to gSlider acquisitions, but can also be easily applied to single-shot and multishot diffusion acquisitions such as single-shot SMS-EPI, multishot EPI acquisitions, which can enable high-resolution, high-quality functional MRI and DWI with low geometric distortion and short acquisition times.

5 | CONCLUSION

We proposed $B_0^+$ and $T_1$ corrections with the dynamic MC $B_0$ shimming strategy to improve the fidelity of high-resolution gSlider-EPI acquisitions with low geometric distortion. In vivo studies demonstrated that the proposed methods enabled markedly improved image quality with reduced slab-boundary artifacts for short-TR acquisition. The more than 2× geometric distortion reduction can be achieved by dynamic multicoil shimming. Combined with parallel imaging with high in-plane acceleration ($R_{\text{inplane}} = 4$), the geometric distortion was further reduced; it which should improve the quality of diffusion data that can be applied to many clinical and neuroscience applications.

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ORCID

Congyu Liao https://orcid.org/0000-0003-2270-276X
Berkin Bilgic https://orcid.org/0000-0002-9080-7865
William A. Grissom https://orcid.org/0000-0002-3289-1827

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