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Original contribution

A feasible study on using multiplexed sensitivity-encoding to reduce geometric distortion in diffusion-weighted echo planar imaging

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ABSTRACT

Purpose: Single-shot (SS) echo planar imaging (EPI) is commonly used for diffusion-weighted imaging (DWI) in radiation treatment planning. The geometric distortion of single-shot diffusion-weighted EPI (SS-DWEPI) limits its utility in precision radiotherapy. We aimed to investigate the use of multiplexed sensitivity-encoding DWEPI (MUSE-DWEPI) to reduce the geometric distortions to allow its use in brain radiotherapy.
Methods: In a phantom study, phantom diameters measured using MUSE-DWEPI and SS-DWEPI were compared and the percentages of geometric distortion (%GD) were calculated. The shifting vectors of control points were also plotted and calculated. In a patient study, ten patients (six with post-surgery glioma, four with brain metastases) requiring MRI were enrolled, and the image distortion levels in SS-DWEPI and MUSE-DWEPI were compared using T2 Periodically Rotated Overlapping Parallel Lines with Enhanced Reconstruction (T2 PROPELLER) images as the reference. The tumor targets and four brain regions were delineated based on T2 PROPELLER, SS-DWEPI, and MUSE-DWEPI, using the Dice similarity coefficient (DSC) and the Hausdorff distance (HD) to quantify the level of geometric distortion.
Results: In the phantom study, the most prominent image distortion was along the phase encoding direction in terms of %GD fluctuated from 7%–8% for SS-DWEPI (b = 0 s/mm^2 and 1000 s/mm^2) and fluctuated from 2%–3% for MUSE-DWEPI (b = 0 s/mm^2 and 1000 s/mm^2) with different positions and b values. The mean relative displacement of all control points was 4.45 ± 3.44 mm for SS-DWEPI b_0 and 2.17 ± 1.9 mm for MUSE-DWEPI b_0 close to the isocenter. Increasing the distance away from the isocenter in the z direction, the distortion increased more in SS-DWEPI. For all brain regions and targets involved, higher DSC values and lower HDs were obtained using MUSE-DWEPI than with SS-DWEPI ($p < .01$). The mean improvement in HD when switching from SS-DWEPI to MUSE-DWEPI was 3.65 ± 1.31 mm.
Conclusions: MUSE-DWEPI is an improvement upon SS-DWEPI in geometric distortion reduction, which might be a promising application strategy for DWI in radiotherapy.

1. Introduction

A critical requirement for the use of magnetic resonance imaging (MRI) in radiation treatment planning is images with high geometric accuracy to allow the tumors and surrounding tissues to be precisely delineated in all three dimensions. Image deformation, unsharpness, and aliasing artifacts have adverse impacts on the accuracy of radiation treatment planning [1].

Diffusion-weighted imaging (DWI) is now a pillar of modern oncologic imaging which is of great value in monitoring tumor treatment response [2,3]. Single-shot (SS) echo planar imaging (EPI) is commonly used for DWI in both diagnostic and radiotherapy MR imaging owing to its efficiency and robustness [3]. Geometric distortion, a weakness of SS-EPI, limits its use in precision radiation treatment planning [4]. Being able to correct EPI image distortion would greatly enhance the utility of diffusion-weighted echo planar imaging (DWEPI).

The geometric distortion in SS-DWEPI arises from magnetization spin dephasing along the readout train length in SS acquisition due to B_0 field inhomogeneities, with consequent phase error accumulation resulting in voxel shifts that distort the image along the phase-encoding direction [5,6]. Several methods have been proposed to reduce image distortion in DWEPI by addressing the issue of B_0 field inhomogeneities.

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Retrospective correction of EPI images can be performed based on an estimate of the B_0 field map [7], but the effectiveness of this procedure is uncertain in regions close to air cavities [8]. Hacck et al. [9] proposed to use the voxel displacement maps from field maps of the resonance frequency for distortion correction. However, the corrected images may be blurred, and the apparent diffusion coefficient (ADC) may be biased. Prospective distortion correction with real time monitoring of the B_0 field map is feasible using an additional field probe [10]; however, its application in routine clinical use is still limited by the high demands of hardware.

In addition to correcting for B_0 field inhomogeneities, a key factor in reducing geometric distortion is shortening the echo train length (ETL) used in acquisition [11,12]. Parallel imaging exploits the spatially varying coil sensitivity weighting from different receiver coil and may effectively reduce the number of phase encoding steps while maintaining the image spatial resolution, hence reducing the level of geometric image distortion [13,14].

However, the acceleration factor in parallel imaging is fundamentally limited by the spatial distribution of the receiver coil elements, which deteriorates the signal-to-noise ratio (SNR). These features restrict the use of a high acceleration factor in SS-DWEPI for geometric distortion reduction, and in practice, only an acceleration factor of 2.0 is commonly used. Reduced field-of-view (rFOV) DWI uses a spatial-spectral 2D RF pulse to restrict image excitation, so that a portion of the overall target may be imaged without imposing aliasing artifacts [15]. In this way, the ETL may be proportionally reduced. However, full FOV imaging is preferable for radiation treatment planning.

Another approach to reduce the readout ETL is segmented acquisition, where the overall ETL in SS-DWEPI is segmented into several shots. There are two possibilities for segmenting the overall k-space EPI trajectory: readout-segmented EPI (rs-EPI) and phase encoding segmented EPI [16,17]. In rs-EPI, the overall trajectory is segmented along the readout directions and an additional navigator echo is used to correct the phase errors among the different readout segments retrospectively [18]. Warren et al. [19] demonstrated the use of rs-EPI to improve geometric fidelity in image-guided radiation therapy of pelvic tumors. However, an intrinsic limitation of rs-EPI is the fact that in cases of continuous motion, there may exist differences between the navigator acquisition and the data acquisition, which may lead to reconstruction failure [20].

Multi-shot acquisition with multiplexed sensitivity-encoding DWEPI (MUSE-DWEPI) segments the overall acquisition along the phase encoding direction. It is an interleaved DWEPI technique correcting nonlinear shot-to-shot phase variations without the need for navigator echoes [21]. MUSE was initially developed to provide high resolution DWI and diffusion tensor tracking. To the best of the authors' knowledge, MUSE-DWEPI has not been investigated for the purpose of radiotherapy. We evaluated the level of geometric distortion and ADC accuracy for use in brain radiotherapy, comparing MUSE-DWEPI with SS-DWEPI in phantoms and *in vivo*.

2. Materials and methods

2.1. MRI acquisition

All the MR scans in this study were performed on a 3.0 T scanner (Discovery MR750W®, GE Healthcare, Milwaukee WI, USA) equipped with a 32-channel head target array coil (32CH coil). The MR scan included routine sequences as well as SS-DWEPI, T2 with Periodically Rotated Overlapping Parallel Lines With Enhanced Reconstruction (T2 PROPELLER), and MUSE DWI. Identical FOVs of 25.6 cm × 25.6 cm were maintained among the three different acquisitions. The scan parameters are shown in Table 1. Both SS-DWEPI and T2 PROPELLER images were reconstructed automatically at the scanner console, whereas MUSE images were reconstructed offline on a separate workstation (PowerG® with AIMS® software, Magtron Inc., Jiangyin, China) using post-processing [21]. The time of reconstructing 50 slices with MUSE-DWEPI was about 4–5 min. ADC values were derived for both SS-DWEPI and MUSE-DWEPI using b-values of 0 s/mm^2 and 1000 s/mm^2. Diffusion gradients were concurrently applied in all three axes (x, y, z) such that the resulting diffusion direction was the vector summation of three axes.

The main workflow of MUSE reconstruction is illustrated as a three-step process in Fig. 1, illustrating a four-shot case. In step 1, four full-FOV images are reconstructed from each of the four shots separately using traditional sensitivity encoding (SENSE) reconstruction, which is susceptible to undesirable noise amplification and shot-to-shot phase inconsistencies [21]. In step 2, the motion-induced phase inconsistencies in between shots are estimated using the MUSE algorithm, and phase inconsistency corrections are performed for SENSE reconstructed images [21,22]. In step 3, the phase inconsistency-corrected images corresponding to different shots are combined to produce the final MUSE DWI image.

2.2. Phantom study

The small American College of Radiology (ACR) cylinder phantom of 100 mm inside length and 100 mm inside diameter was used in this study because of its similarity in size to the head coil. The phantom was kept in the scanner room for at least 24 h to reach thermal equilibrium, while the room was kept at a constant temperature of 20 °C during the whole study. Seven slices of 5 mm thickness and 3 mm gap as specified in the guidelines [23] were acquired in all three acquisitions with matching spatial locations. The geometric distortion calculation was performed on the b_0 and b_{1000} images (b = 0 s/mm^2 and 1000 s/mm^2) from SS-DWEPI and MUSE-DWEPI separately. The geometric distortion section was separately prescribed at 0.8 cm and 4.8 cm away from the isocenter in the z direction.

Diameters along four evenly spaced radial lines (0°, 90°, ± 45°) on the geometric distortion section were measured and the percentages of geometric distortion (%GD) were calculated according to the following equation [24]:

$$\%GD = \frac{actual\ dimension - measured\ dimension}{actual\ dimension} \times 100 \tag{2}$$

In the diffusion images (b_0 and b_{1000}), 38 control points on the geometric distortion section were used to analyze the vector of shifting, and the relative displacements were calculated between the DWI sequences and actual values. The overall distortion levels were quantitatively measured using the mean value with standard deviation (SD) and root-mean-square error (RMSE) of the relative displacements (Eq. (3)).

$$\boldsymbol{RMSE} = \sqrt{\frac{\sum_{i=1}^{N} \Delta \boldsymbol{d}_i^2}{\boldsymbol{N}}} \tag{3}$$

where i is the index of control points, N is the total number of control points, and Δd_i is the relative displacement of the ith control point.

The ADC maps of the image uniformity section were compared between SS-DWEPI and MUSE-DWEPI for structural uniformity. A 50 cm^2 circular region of interest (ROI) was used for calculating at the image center.

To ensure consistency and reproducibility of the results, the measurements of %GD, the displacements of control points, and ADC maps were performed using in-house programs developed using commercial software (Matlab®, Mathworks Inc., Natick MA, USA) [25].

2.3. Patient study

Approval of the study protocol was obtained from our Institutional Review Board prior to the study. Between August 2017 and December