

GEOMETRIC DISTORTION CORRECTION IN EPI BY PHASE LABELING USING SENSITIVITY ENCODING (PLUS)

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ABSTRACT

Geometric distortion induced by magnetic field inhomogeneity is an intrinsic problem for echo-planar imaging (EPI). Most of the existing correction techniques require separate field map scans in addition to the measurement scans. The effectiveness of these methods requires consistent patient position between these scans, thus inter-scan patient motions degrade the correction results. We propose a new method for geometric distortion correction in EPI that derives the field maps from the measurements themselves, without the need of additional field map scans. This method takes advantage of parallel imaging and k-space trajectory modification to generate multiple images for field mapping from a single measurement needed. The derived field maps are retrospectively applied to the measurements. The proposed method has been successfully demonstrated on combined k-space data that simulate the data from the modified k-space trajectories. These data were acquired from a phantom and a human brain.

Index Terms—EPI artifact, field mapping, geometric distortion correction, magnetic field inhomogeneity, parallel imaging

1. INTRODUCTION

Echo-planar imaging (EPI) is a fast MRI data acquisition scheme that has been widely used in structural and functional imaging. However, EPI is sensitive to magnetic field inhomogeneity caused by imperfect shimming and tissue susceptibility differences. This magnetic field inhomogeneity creates local magnetic field gradients superimposed on the applied spatial encoding gradients, causing incorrect spatial encoding/decoding, i.e., geometric distortion. For stereotactic or image-guided applications, image geometric accuracy is crucial. In functional MRI (fMRI), diffusion tensor imaging (DTI), and DTI-based tractography, severely distorted images are difficult to register to anatomical images, leading to incorrect spatial mapping of structures of interest (i.e., activation loci and/or white matter fibers). Furthermore, severity of such

geometric distortion may be different from one subject to another, reducing the power of group analysis.

Many published distortion correction algorithms generally involve two steps: first acquiring field inhomogeneity information and second applying the information to correct the distortion. To obtain the field inhomogeneity information, single or multiple scans are performed with varying scan parameters, such as echo time, direction or polarity of the phase/frequency encoding, location of the k-space data, number of echoes, and number of encoding directions. Many methods are neither practical because of their lengthy reference scans nor sufficiently robust for a high degree of distortion. Moreover, field inhomogeneity information is generally derived from a fixed position of a patient in the magnet and is valid only for that position. Therefore, the scans for field inhomogeneity information have to be acquired immediately before or after the measurements to reduce patient position inconsistency. For lengthy or repeated measurements, such as fMRI or dynamic contrast agent studies, patient motions during or between the scans could invalidate the obtained field maps. It is also impractical to sacrifice the temporal resolution by inserting reference scans in between fMRI measurements or dynamic scans.

In this paper, we proposed a method dubbed phase labeling using sensitivity encoding (PLUS), that utilizes the measurement data themselves (without acquiring separate field map scans) to correct the geometric distortion. PLUS embodies the reference and measurement scans into one scan, which further develops from our previous work [1, 2]. It utilizes the phase labeling for additional coordinate encoding (PLACE) [3, 4] to map the field inhomogeneity, and utilizes the simulated phase evolution rewinding (SPHERE) [5] to apply the field maps to correct the distortion. PLUS incorporates a parallel imaging technique [6] using a phased array coil and its spatial sensitivity to produce the multiple images needed for field mapping from a single measurement scan. The k-space trajectories are modified such that the images after parallel image reconstruction contain the information of the local phase shifts. These phase shifts are afterward applied to correct the distortion of the measurement image. The measurement image can be reconstructed conventionally to preserve the signal-to-noise ratio (SNR) or reconstructed using parallel imaging to reduce the ghosting artifacts. The proposed

method has been successfully demonstrated on combined k-space data that simulate the data from the modified k-space trajectories. These data are acquired from a phantom and a human brain.

2. MATERIALS AND METHODS

To create a field map, PLACE requires extra scans that have different pre-phase-encoding areas compared to the measurement scan. In order to unify these scans, PLUS skips lines on each of those k-space data sets and acquires the remaining lines in a time multiplex fashion. An example of k-space data unification is shown in Fig. 1. In this figure, three k-space data sets with -1, 0, and +1 additional gradient blips (in the upper panel and from left to right, respectively) are combined into one k-space data set (in the lower panel). Note that the k-space trajectory is presented by the thin dotted lines. The first lines (the lowest lines) on the k-space trajectories (in the upper panel) are acquired first and at the same time relative to the pulse sequences but these lines are in different positions in k-space. Each line in the combined k-space data set (in the lower panel) has to be selected from one of these k-space data sets (in the upper panel). Each k-space location can be selected only once. In addition, because of the time multiplexing, each line has to be selected from a different time (i.e., different line numbers on the k-space trajectory). For these reasons, only some combinations of the lines could be formed and the k-space trajectory has to be modified. As shown in Fig. 1, for every four lines, the third line of the combined k-space data is acquired before the second line. If repeated k-space locations are allowed, more combinations could be generated. However, such repeated acquisition would reduce the efficiency of the algorithm. The combined k-space data in Fig. 1 is only one of many possibilities for unification. More k-space data sets could be applied; however, this demands higher degree of parallel imaging. The EPI sequence diagrams corresponding to Fig. 1 are demonstrated in Fig. 2.

After obtaining all the lines in the combined k-space data, these lines are sorted into groups according to their original number of additional blips. Each group represents k-space data with lower sampling rate in the phase-encoding direction. For the combination in Fig. 1, the lines are sorted into four groups, instead of three, by grouping one line from every other four lines. The Fourier transformation of these k-space data sets will be aliasing or folded. The missing lines on each data set are estimated, i.e., the images are unfolded, using a parallel image reconstruction [6]. Therefore, the number of k-space data sets used in the combination cannot exceed the number of phased array coils used in the measurement. The phase differences between pairs of the reconstructed images are calculated and averaged to create a phase-shift map as described in PLACE algorithm [1-4]. Finally, the SPHERE algorithm is applied

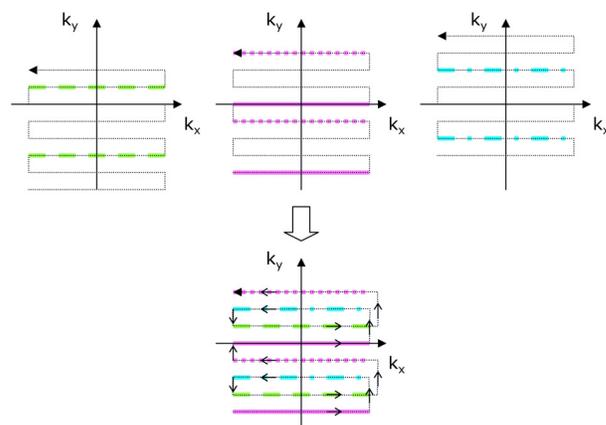


Fig. 1. Combining of k-space data from three scans (upper panel) with -1, 0, or +1 additional blips in the pre-phase-encoding gradient (from left to right, respectively) into a combined k-space data set (lower panel). The thick lines (dashed, solid, dotted, and dash-dot) indicate the lines that are taken from each of the k-space data. The lines in the combined k-space data are retrospectively sorted into four groups as indicated by their line types. Note that the thin dotted lines represent k-space trajectories. The thinner arrows (lower panel) indicate directions of the combined k-space trajectory.

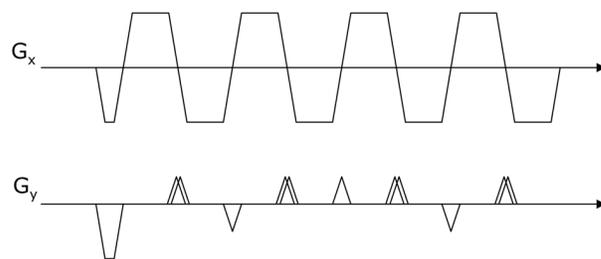


Fig. 2. The EPI sequence diagram corresponding to Fig. 1. Note that the blip gradients are double in some locations, as represented by overlapping triangles.

to correct the image distortion using the created phase-shift map. The phase-shift map can be applied to the image reconstructed conventionally from the combined k-space data or to the average of the images from parallel imaging reconstruction.

MRI experiments were performed on a phantom and a normal volunteer's brain using a 3T MR scanner with an eight-channel SENSE head coil (Achieva, Philips Medical Systems, Best, The Netherlands). A general gradient echo EPI sequence was used in both experiments. This sequence was intentionally created for fMRI studies in our lab. The scan parameters were as follows. TR was fixed at 2500 ms and TE was 35 ms. FOV was 320 mm with $2.5 \times 2.5 \text{ mm}^2$ in-plane resolution for the phantom and was 256 mm with $3 \times 3 \text{ mm}^2$ in-plane resolution for the brain. Slice thickness was 4 mm. Reference scans for parallel imaging were acquired at the beginning of the experiments with the following parameters. FOV is around 410 mm with an in-plane resolution of $3 \times 3 \text{ mm}^2$. Slice thickness and slice locations

were the same as in the measurements. The data were acquired using a pulse-sequence patch that acquired three dynamic scans, each of which contained an additional blip of -1, 0, or +1 blips. The combined k-space data sets were generated by rearranging the k-space lines from these dynamic scans as shown in Fig. 1. Raw data from individual coils were recorded. All image processing was performed offline using MATLAB (MathWorks, Inc., Natick, MA) on a PC with a 3.4-GHz Pentium-4 CPU and 3.5-GB RAM.

3. RESULTS

The k-space trajectory for the combined k-space data is slightly different from a conventional linear EPI scan, i.e., for every four lines, the scan order of the second line and the third line are switched (the third line is scanned before the second lines). After sorting the lines, four four-fold folded images are generated. Each k-space data set of these folded images contains lines scanned in the same readout gradient polarities, i.e., unidirectional readout. For example, the data set that keeps the first or the second lines contains all lines in the positive readout gradient polarity (i.e., scanning from left to right). The other data sets that keep the third or the fourth lines contains all lines in the negative readout gradient polarity (i.e., scanning from right to left). The folded images are then unfolded using parallel imaging.

The phase-shift maps generated using different pairs of the unfolded images are investigated. For convenience, let u_n be an unfolded image created using the data set that keeps the n^{th} line of the combined k-space data. We investigate the phase-shift maps using pairs of (1) u_1 and u_2 , (2) u_3 and u_4 , (3) u_1 and u_3 , and (4) u_2 and u_4 . Note that the numbers in these parentheses represent the method number. Methods 1 and 2 pair images that are scanned in the same direction (i.e., Method 1 pair images that are scanned from left to right, and Method 2 pair images that are scanned from right to left), while Methods 3 and 4 pair images that are scanned in the opposite readout directions. The results after applying these phase-shift maps are shown in Fig. 3. In general, Methods 1 and 2 provide better correction results than the others. As shown, Methods 1 and 2 almost remove the geometric distortion. However, some small areas around the center of the images are slightly deformed. Methods 3 and 4 generate unacceptable results because they introduce more artifacts, especially on the top half of the images, where the top left sides of the images are compressed and the top right sides are expanded.

We further combined Methods 1 and 2 using averaging. The averaged phase-shift map $\Delta\phi$ can be written as

$$\Delta\phi = \text{Arg}((RPR(u_1)RPR(u_2)^* + RPR(u_3)RPR(u_4)^*)/2), \quad (1)$$

where $RPR(\cdot)$ is a function to remove phase ramp from the input image, and $\text{Arg}(\cdot)$ is a function to determine the phase image from the input. This phase-shift map is then used to correct the distortion. The result using the averaged phase-shift map is shown in Fig. 4c. The T1-weighted image and

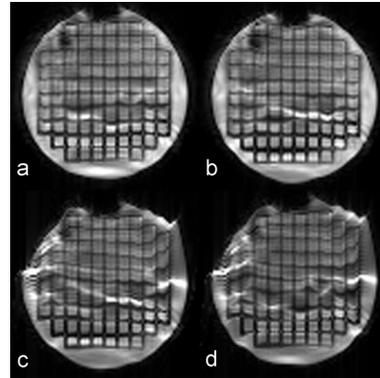


Fig. 3. Distortion correction results using phase-shift maps from Methods 1 to 4 from (a) to (d), respectively.

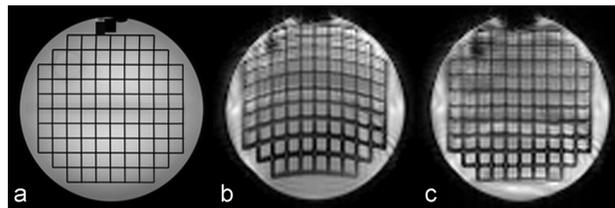


Fig. 4. Comparison of images from a phantom: (a) a T1-weighted image, (b) distorted image, and (c) a corrected image using averaged phase-shift maps from Method 1 and 2.

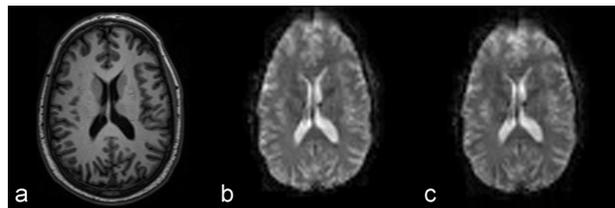


Fig. 5. Comparison of images from the brain of a volunteer: (a) a T1-weighted image, (b) distorted image, and (c) a corrected image using averaged phase-shift maps from Method 1 and 2.

the distorted image from the same slice are also shown in Figs. 4a and 4b, respectively. As expected, the results are improved by using the averaged phase-shift map, and the distortion is almost completely corrected. This combined method is also performed on other slices and on other data sets, and the results confirm this finding. The results from the brain of a volunteer are displayed in Fig. 5. A T1-weighted image, a distorted image, and a corrected image are shown in Figs. 5a to 5c, respectively. As can be seen, the elongated frontal part of the brain is corrected and matched with the T1-weighted image.

4. DISCUSSION

In this paper, we have introduced the basic concept of PLUS to correct the geometric distortion in EPI without using extra scans for field mapping. The technique modifies the EPI trajectory and utilizes parallel imaging to create multiple images that are later used to reconstruct phase-shift maps for distortion correction. Since the phase-shift maps

are derived from the measurements themselves, the proposed method eliminates the error caused by the inconsistency of patient positions between the field mapping scans and the measurements. Therefore, it enables the possibility to apply the correction in real time measurements.

We have demonstrated that the quality of the phase-shift maps can be improved by averaging a number of phase-shift maps. As shown in our previous work [1, 2], for regular EPI readout, the number of phase-shift maps should be at least four. However, in this study, if unidirectional readout is applied, only two phase-shift maps are needed.

Using unidirectional readout in EPI eliminates ghosting artifact and improves the quality of the phase-shift maps. To calculate phase shifts, the images have to possess the same readout direction. Otherwise, the phase shifts will be in error. This error is caused by the off-resonance phase accumulated along the readout. This off-resonance phase will cancel out if the readout is in the same direction. The averaging can be applied later, even though the images are derived from different readout directions.

The modified k-space trajectory used in PLUS induce ghosting artifacts on the measurements if using the conventional sum of the squares. Therefore, suppression of these ghosting artifacts [7, 8] is useful for improving the measurement accuracy. If the measurements are reconstructed using parallel imaging from unidirectional readout folded images, multiple unfolded ghost-free images can be generated and averaged to improve their signal to noise ratio. Nevertheless, these ghosting artifacts do not affect our phase error estimation (if applying unidirectional readout) and therefore the phase-shift maps are free from this distortion. In addition, the modified k-space trajectory requires stronger blip gradients. It may need different compensation to avoid eddy-current artifacts.

PLUS is not suitable for the sequence that already utilized parallel imaging with a reduction factor greater than one. This is because PLUS further applies parallel imaging with a high reduction factor. The product of these reduction factors might exceed the parallel imaging limit, i.e. the number of coils, or the noise level would be too high for phase-shift calculations.

Our results benefit from the coil sensitivity maps that are extracted from the reference scans with the same pulse sequence and the same slice locations as in the measurement scans. The accuracy of the phase-shift maps may be limited if the reference scans are performed with a different setup. In our study, the inconsistency of patient positions between the reference scan and the measurements could decrease the accuracy of the estimated sensitivity maps. To avoid the dependency of reference scans, self-calibration schemes [9-11] may be applied.

In this work, we tested the PLUS algorithm using combined k-space data instead of acquiring the data directly using the modified k-space trajectories. The results

demonstrated the effectiveness of the algorithm. Our future work will include implementation of a pulse sequence patch that acquires the data directly according to the modified k-space trajectory and evaluation of this patch in fMRI studies.

5. ACKNOWLEDGMENTS

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6. REFERENCES

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