Review and comparison of geometric distortion correction schemes in MR images used in stereotactic radiosurgery applications

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Abstract. In Stereotactic Radiosurgery (SRS), MR-images are widely used for target localization and delineation in order to take advantage of the superior soft tissue contrast they exhibit. However, spatial dose delivery accuracy may be deteriorated due to geometric distortions which are partly attributed to static magnetic field inhomogeneity and patient/object-induced chemical shift and susceptibility related artifacts, known as sequence-dependent distortions. Several post-imaging sequence-dependent distortion correction schemes have been proposed which mainly employ the reversal of read gradient polarity. The scope of this work is to review, evaluate and compare the efficacy of two proposed correction approaches. A specially designed phantom which incorporates 947 control points (CPs) for distortion detection was utilized. The phantom was MR scanned at 1.5T using the head coil and the clinically employed pulse sequence for SRS treatment planning. An additional scan was performed with identical imaging parameters except for reversal of read gradient polarity. In-house MATLAB routines were developed for implementation of the signal integration and average-image distortion correction techniques. The mean CP locations of the two MR scans were regarded as the reference CP distribution. Residual distortion was assessed by comparing the corrected CP locations with corresponding reference positions. Mean absolute distortion on frequency encoding direction was reduced from 0.34mm (original images) to 0.15mm and 0.14mm following application of signal integration and average-image methods, respectively. However, a maximum residual distortion of 0.7mm was still observed for both techniques. The signal integration method relies on the accuracy of edge detection and requires 3-4 hours of post-imaging computational time. The average-image technique is a more efficient (processing time of the order of seconds) and easier to implement method to improve geometric accuracy in such applications.

1. Introduction

In order to take advantage of the superior soft tissue contrast compared to computed tomography (CT), Magnetic Resonance (MR) images are being increasingly used in radiotherapy treatment planning [1], especially in stereotactic radiosurgery (SRS) applications for the management of a wide variety of brain lesions. Despite resulting in better and more efficient tumor delineation, MR images introduce additional spatial inaccuracies in dose delivery stemming from geometric distortion, an inherent
feature of MRI [2]. Although major contributors to distortion, such as static magnetic field inhomogeneity and gradient field nonlinearity are system-related and can be measured a priori and corrected for as a post-imaging process [3,4], other sources of geometric distortion are patient/object-induced (i.e., owing to susceptibility differences and chemical shift artifacts) and, therefore, cannot be predicted or accounted for using phantoms, as the set of magnetic properties depend on the imaging object and vary from patient to patient in addition to not being stable over time [2]. In non-EPI (echo planar imaging) 3D pulse sequences, both static field inhomogeneity and object-induced distortions mainly affect the frequency encoding direction of the image and are sequence-dependent, with effects being minimal or negligible on phase encoding directions [4].

Several sequence-dependent distortion correction schemes have been presented [4–7], all requiring an extra imaging step and, therefore, resulting in additional scanning time. An efficient approach for distortion correction in 3D imaging protocols is to take advantage of the fact that sequence-dependent distortions are polarity dependent [4], in the sense that distortion changes sign with the polarity of the frequency encoding gradient field. Correction methods that rely on read gradient reversal account for all sequence-dependent sources of distortion including static magnetic field inhomogeneity, susceptibility differences and chemical shift artifacts but disregard distortion induced by gradient nonlinearities (sequence-independent distortion [4]). In this work, a custom-made prototype head phantom was employed to investigate and compare the efficacy of two proposed sequence-dependent distortion correction schemes; the average-image method and the signal integration technique, both relying on read gradient reversal.

2. Materials and Methods

2.1. The phantom
A custom head-size MR phantom was implemented [8] to study the efficacy of distortion correction techniques. The phantom incorporates acrylic planes on which small 947 3mm-diameter holes are drilled, serving as control points (CPs) for distortion detection. The phantom is MR scanned filled with copper sulfate solution, yielding high-contrast with acrylic and facilitating CP localization. However, differences in magnetic susceptibility between the solution and acrylic also result in object-induced geometric distortions, which come in addition to existing static magnetic field inhomogeneity related spatial distortions [2,4].

2.2. Image acquisitions
The phantom was filled with copper sulfate solution and positioned in the center of the head coil of a Philips Multiva 1.5T scanner (Philips Medical Systems, The Netherlands). A clinically used for intracranial SRS treatment planning 3D T1-weighted Fast Field Echo (FFE) pulse sequence was employed using a reconstruction voxel size of 0.98 x 0.98 x 1 mm³, a bandwidth of 191 Hz/pixel and selecting the y-axis (i.e., Anterior-Posterior) as the frequency encoding direction. Since sequence-dependent distortion correction schemes are based on the reversal of read gradient polarity [4], an identical pulse sequence was added to the imaging protocol after reversing the read gradient direction (i.e., Posterior-Anterior), doubling the scanning time. The MR imaging protocol was repeated by applying the frequency encoding direction on the x-axis (i.e., Left-Right and Right-Left directions). 3D sequence-independent distortion correction routines, provided by the MR scanner vendor, were enabled and applied to all sequences acquired, thus, minimizing distortions due to gradient nonlinearities which are not corrected nor accounted for by the read gradient reversal method.

2.3. Image processing
Using in-house MATLAB (The MathWorks, Inc., Natick, MA) routines, forward and reversed polarity image scans were processed to determine the distortion of the original images. In particular, the mean CP location identified in the opposed polarity MR scans was considered as the reference (undistorted)
CP position. Distortion of the original image was determined by the spatial offset between CP location in the forward image scan and the corresponding reference location.

2.3.1. Signal integration correction method. The signal integration technique for sequence-dependent distortion correction is described in detail in the work of Morgan et al [5]. Briefly, the method relies on the fact that the integral of the image signal along the frequency encoding direction is not affected by spatial distortion and, therefore, matching points between the forward and the reversed polarity MR scans are identified at the locations where the two integrals equalize, yielding the distortion map. In addition, the original image is also corrected for the pixel intensity using the Jacobian for the transformation between the undistorted and distorted image spaces [9].

2.3.2. Average-image correction method. The average-image method was presented and evaluated for a series of multiple brain metastases cases [7]. It simply combines the forward and reversed polarity images into a new image in which pixel intensity is the average of the original images. Consequently, post-image processing is minimal (of the order of seconds) and straightforward. However, no signal intensity corrections can be applied since distortion maps cannot be determined.

Following implementation of both distortion correction methods, residual distortions were assessed as the CP offsets between the reference and the corresponding corrected CP locations.

3. Results and discussion
Detected distortion for the original images identified using all 947 CPs is presented in table 1. The mean distortion in the entire volume scanned is 0.3 mm irrespective of selection of frequency encoding direction, while the maximum distortion exceeds 1 mm. After applying the correction techniques, mean and median distortion reduces to practically zero. However, the maximum detected CP offset reaches 0.7 mm which could be attributed to non-systematic errors of the CP localization algorithm occurring in areas of reduced signal. Implementation of the signal integration method requires a few hours of post-imaging computational time while the average-image method is more efficient (processing time of the order of seconds).

| Frequency encoding direction | Image          | Range (mm)  | Mean ± 1std (mm) | Mean abs ± 1std (mm) | Median (mm) |
|-----------------------------|----------------|-------------|------------------|----------------------|-------------|
|                             | Original       | -0.16 – 1.07| 0.34 ± 0.23      | 0.34 ± 0.23          | 0.35        |
| y-axis (A-P)                | Average-Image  | -0.75 – 0.49| -0.02 ± 0.20     | 0.14 ± 0.15          | 0.00        |
|                             | Signal Integration | -0.59 – 0.77| 0.05 ± 0.22      | 0.15 ± 0.16          | 0.00        |
|                             | Original       | -0.12 – 1.06| 0.32 ± 0.25      | 0.33 ± 0.24          | 0.27        |
| x-axis (L-R)                | Average-Image  | -0.70 – 0.74| -0.05 ± 0.18     | 0.12 ± 0.14          | -0.01       |
|                             | Signal Integration | -0.56 – 0.68| -0.03 ± 0.18     | 0.13 ± 0.13          | 0.01        |

- Mean detected distortion ± 1 standard deviation
- Mean detected distortion magnitude ± 1 standard deviation

Figure 1 depicts the distortion magnitude and orientation for the original and corrected images with the frequency encoding gradient set on y-axis. For the original image (figure 1a), distortion is mainly directed towards the frequency encoding direction due to the combined effect of susceptibility differences and static magnetic field inhomogeneity. For both corrected images (figures 1b and 1c), distortion on y-axis is negligible while minimal residual distortion is observed at random orientations.

Figure 2 presents histograms of the identified distortion using all 947 CPs in all image series for direct comparison. While for the original images, distortion distribution is shifted towards the positive distortion direction, residual distortion identified in the corrected images is distributed around zero.
Figure 1. Distortion vectors (magnified by a factor of 15 to facilitate readability) for an indicative axial plane of the phantom corresponding to the original (a), as well as the average-image (b) and signal integration (c) corrected images with the read gradient direction set on the y-axis.

Figure 2. Histograms of the identified distortion for the original and corrected images with the read gradient direction set on y-axis (a) and x-axis (b).

4. Conclusion
Both sequence-dependent distortion correction techniques perform equally well, minimizing the mean and median residual distortions. However, the signal integration method requires a few hours of post-imaging computational time while the average-image method is simple and efficient.

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