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Journal
International Journal of Radiation Oncology • Biology • Physics, 90(1)

ISSN
0360-3016

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Publication Date
2014-09-01

DOI
10.1016/j.ijrobp.2014.05.500

Peer reviewed

Author Manuscript
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Published in final edited form as:

Pract Radiat Oncol. 2016 ; 6(6): e319–e328. doi:10.1016/j.prro.2016.05.008.

Distortion Inherent to Magnetic Resonance Imaging (MRI) Can Lead to Geometric Miss in Radiosurgery Planning

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Abstract

Purpose—Anatomic distortion is present in all MRI data due to non-linearity of gradient fields and measures up to several millimeters. We evaluate the potential for uncorrected MRI to lead to geometric miss of the target volume in stereotactic radiosurgery (SRS).

Methods and Materials—Twenty-eight SRS cases were studied retrospectively. MRIs were corrected for gradient non-linearity distortion in three dimensions, and gross tumor volumes (GTVs) were contoured. The manufacturer-specified distortion field was then re-applied to GTV masks to allow measurement of GTV displacement in uncorrected images. The uncorrected GTV was used for SRS planning, and the dose received by the true (corrected) GTV was measured.

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A preliminary version of these results was presented at the American Society for Radiation Oncology 2014 Annual Meeting in San Francisco, CA.

Conflict of interest notification

Nathan S. White has no conflicts of interest.
Gwe-Ya Kim has no conflicts of interest.
Carrie R. McDonald has no conflicts of interest.
Nikdokht Farid has no conflicts of interest.
Hauke Bartsch has no conflicts of interest.
Joshua Kuperman has no conflicts of interest.
Roshan Karunamuni has no conflicts of interest.
Deborah Marshall has no conflicts of interest.
Parag Sanghvi has no conflicts of interest.
Daniel R. Simpson has no conflicts of interest.
Arno J. Mundt has no conflicts of interest.

Results—Median displacement of the GTV due to gradient distortion was 1.2 mm (interquartile range 0.1 mm – 2.3 mm), with a minimum of 0 mm and a maximum of 3.9 mm. Eight of the twenty-eight cases met *a priori* criteria for "geometric miss."

Conclusions—While MRI distortion is often subtle on visual inspection, there is a significant clinical impact of this distortion on SRS planning. Distortion-corrected MRI should uniformly be used for intracranial radiosurgery planning because uncorrected MRI can lead to potential geometric miss.

Introduction

Stereotactic radiosurgery (SRS) depends on highly accurate image guidance. For intracranial SRS, planning is generally based on magnetic resonance imaging (MR imaging, or MRI) because of its superiority to computed tomography (CT) in soft tissue contrast and sensitivity for delineating radiation targets and normal brain tissue [1–3]. However, MRI is subject to anatomic distortion from multiple sources, including static-field inhomogeneity, eddy currents, and gradient field non-linearity [4–6]. If uncorrected, this anatomic distortion might lead to missed targets and unnecessary treatment of normal brain tissue [7]. While MR physicists have long known of this problem and have developed methods to correct it, image processing is typically the purview of diagnostic radiology departments, where the primary concern is generally *qualitative tissue contrast*, not millimeter-level anatomic accuracy. As such, radiation oncologists and neurosurgeons performing radiosurgery are often using MR images for radiation planning that are uncorrected or only partially corrected (e.g., in only 2 dimensions). Small distortion effects due to gradient field non-linearity on modern scanners may go unnoticed on visual inspection and yet might still be meaningful in the accuracy of precision radiosurgery.

Gradient magnetic fields are required for spatial localization in MRI. Generating images from raw MR data assumes that the gradient fields are linear, but there is some geometric distortion of every MRI due to gradient non-linearity, regardless of the MRI sequence. This non-linearity is more severe on newer scanners with faster and more powerful gradient coils at the expense of intentionally compromising linearity [8, 9]. The magnitude of gradient non-linearity distortion has generally been reported up to 4–6 mm—with some reports of greater than 10 mm—and often represents the largest source of uncorrected distortion [6, 10–12].

As gradient non-linearity is part of the design specifications of MR systems, manufacturers can calculate how the images are expected to be distorted [13, 14]. Moreover, the distortion due to gradient non-linearity can be corrected computationally in three dimensions using the parameters known to manufacturers [13–16]. Nevertheless, these corrections are not universally applied in the clinical setting [16, 17], perhaps because the problem is not widely understood by clinicians or because the effects are assumed to be too small to be clinically meaningful. For neuroradiologists reading the high resolution MRI studies to evaluate the evolution of a lesion over time, slight anatomic distortions may not be clinically relevant. However, radiation oncologists can accept the images they are provided for SRS planning without realizing that diagnostic radiologists may choose to allow some geometric inaccuracy in the interest of preserved image sharpness (i.e., avoiding correction of images to avoid interpolation steps that blur the images). A standard procedure for comprehensive correction of MRI distortion has not been established, though the American Association of Physicists in Medicine (AAPM) has begun efforts to produce a guideline [18].

In this study, we seek to explicitly evaluate the clinical impact of gradient non-linearity distortion in intracranial radiosurgery planning. Previous studies have presented methods for correction of measured distortion using MRI phantoms, establishing a solid theoretical framework for understanding the problem and its potential solution [13, 16]. To study the effect of this distortion on accuracy of intracranial SRS, MR images of patients with brain targets treated with SRS are compared before and after correction for distortion. We hypothesized that displacement of targets due to gradient non-linearity distortion on modern MR systems may be sufficient to cause geometric miss in SRS. To our knowledge, this is the first study to assess the effect of gradient non-linearity on target coverage in patients undergoing SRS.

Methods and Materials

Patients and imaging

We performed a retrospective analysis of 28 consecutive intracranial radiosurgery cases treated in our department from 2011–2014. To meet inclusion criteria, we included cases for which radiosurgery planning was performed using our institution’s preferred high-resolution brain tumor MRI protocol on the same scanner. Twenty-eight such cases were identified. All cases had high-resolution planning MR images acquired on a 3T Signa Excite scanner (GE Healthcare, Milwaukee, Wisconsin). The current preferred imaging protocol includes pre- and post-gadolinium 3D volumetric, T1-weighted, inversion recovery spoiled gradient-echo sequences (TE, 2.8 ms; TR, 6.5 ms; TI, 450 ms; flip angle, 8 degrees; FOV, 24 cm; 0.93 × 0.93 × 1.2 mm; sagittal). All images underwent standard processing, including intensity normalization and correction for B0 inhomogeneity distortion as described previously [19]. Of note, patient-induced susceptibility effects (including from fat/water chemical shift) can be minimized by using a larger readout bandwidth [20]. A previous study reported sub-millimeter susceptibility distortion with a bandwidth of 180 Hz/pixel [21]; for images in the present work, readout bandwidth was 244 Hz/pixel.

MR images were corrected for gradient non-linearity distortion in three dimensions based on specifications provided by the scanner manufacturer in the form of spherical harmonics [13, 22]. The correction represents a post-processing step that can be applied as desired to raw images, which remain available for review (e.g., by a radiologist). The algorithm and software utilized for these corrections has been formally validated previously [13]. It does not depend on the MR sequence used and can therefore be applied to all images acquired on a given MR system. This method has been used widely in the neuroimaging literature, including large, multi-institutional neuroimaging cooperative studies [23] and is freely available (www.birncommunity.org/tools-catalog/gradient-non-linearity-distortion-correction). This 3D algorithm was chosen instead of using the distortion correction feature available on our GE MR system because the latter only corrects in two dimensions.

Target definition and distortion measurement

A single attending radiation oncologist contoured the gross tumor volume (GTV) on the corrected high-resolution (0.93 × 0.93 × 1.2 mm) T1 post-contrast MRI. For cases with multiple brain metastases, a single lesion was selected for simplicity of analysis. The manufacturer-specified distortion field was then re-applied to GTV masks using the inverse of the correction algorithm. This step allowed direct comparison of corrected and uncorrected GTVs on the same MR image, as well as measurements of the changes due to distortion. Direct comparison in MR space also avoided the introduction of additional variability through a second contouring step [24]. Displacement of the center of mass of the uncorrected GTV, relative to its corrected position, was measured for each lesion in millimeters. Dice similarity coefficients (0–1) were used to measure the degree of overlap of corrected and uncorrected GTVs.

Gradient non-linearity distortion has been reported to increase with distance from the center of the magnetic field [9, 12, 13, 22, 25]. To investigate this effect in our own data, we calculated the Euclidean distance (in 3D) of the center of mass of the true, corrected GTV from the magnet isocenter. We then correlated this distance from isocenter with the measured displacement of each GTV center of mass when distortion was not corrected. A correlation coefficient is reported, along with a p-value for difference from zero correlation.

Some MR systems come equipped with software to correct gradient non-linearity distortion in 2 dimensions (2D) only, leaving the through-plane distortion effects uncorrected. Such a system is available on the MR system used for this study. To assess whether 2D-only correction would be adequate for SRS treatment planning, we performed an additional analysis of one-dimensional displacements of the treatment targets. 2D-only algorithms correct the in-plane distortion, so their efficacy for a particular target depends on the slice acquisition plane. SRS planning requires accuracy in all 3 dimensions, and slice acquisition could be done in any plane chosen by the radiologist. Any of the standard planes (coronal, sagittal, or axial) potentially could have been used for the patients in this study, so the through-plane distortion was calculated for each of these options. The maximum of the 3 displacements (i.e., corresponding to the worst acquisition plane for that target, in terms of distortion) was taken for each target and compared to the corresponding displacement without 2D or 3D distortion correction.

Radiosurgery Re-planning

To assess the clinical significance of anatomic distortion, an SRS plan was generated using each uncorrected GTV, and these plans were evaluated for coverage of the corrected GTV to demonstrate the dose distribution that would be achieved if the gradient non-linearity distortion had not been corrected. Analyses were performed in corrected MRI space, as described above, to isolate the distortion effect and avoid confounding with registration effects. CT data from each patient was used only to calculate absorbed dose. Co-registration was achieved with a validated automated algorithm [26] and verified by visual inspection by a radiation oncologist and radiologist. The corrected MR images best approximate the anatomical accuracy of CT, so these images were used for the co-registration. The same registration (i.e., transformation matrix) was applied to both the corrected and uncorrected

MR data to maintain the relative positions of the two GTVs and to allow study of the effect of MR distortion independent of registration factors. Plans were generated using volumetric modulated arc therapy (Eclipse software system, Varian Medical Systems, Palo Alto, CA). Radiosurgery prescription dose was determined by size of the GTV per routine clinical practice at our institution, and all analyses compared relative dose. No margin was included for setup error, as this will vary across institutions and would only exacerbate any distortion effect by increasing the volumes being distorted.

Analyses of geometric miss in original patient position

To assess whether distortion was sufficient to cause a potential "geometric miss" of the target volume, the dose received by the true (corrected) GTV was measured for plans where the uncorrected GTV was the planning target. In our institution's clinical practice, plans are considered inadequate if greater than 2% of the planning target receives less than 100% of the prescription dose (i.e., V100 less than 98%). A recent cooperative group SRS clinical trial deems a plan a Major Deviation from protocol if any portion of the target receives less than 90% of the prescribed dose [27]. In this analysis we chose to highlight particularly inadequate plans by defining criteria for a plan leading to a "geometric miss". The true GTV was considered to be missed in this study if more than 10% of the volume received less than 90% of the prescription dose (i.e., V90 < 90%). Additionally, if the under-dosed (<90% of prescription) portion of the true GTV was greater than 125 mm^3 (equivalent to a cube with 5 mm sides), the distortion effect was also deemed to have caused a geometric miss. In our practice, we frequently treat tumors of this volume, so undertreating 125 mm^3 of tumor tissue is considered clearly unacceptable. Both of these criteria represent a gross violation beyond the Major Deviation criterion in the clinical trial.

Analyses of geometric miss in alternate patient positions

Patient position in the MR scanner is typically influenced by patient comfort and user variability by MR technologist. However, positioning may have a non-intuitive impact on the magnitude of distortion because the gradient non-linearity effect is a property of the precise position in the scanner. To evaluate whether reasonable changes in position might increase distortion effects, we repeated the above analysis, this time applying the distortion field to the contoured GTV after assuming one of 15 probable alternate positions. Included alternate positions were a rotation of 10 degrees in any plane; 10 cm farther or nearer into the bore; 2.5 cm to the left or right on the table; 2.5 cm up off the table (e.g., an extra pad under the head); and either 20 or 30 degrees rotation in the sagittal plane, which we have found in clinical observations to be the most variable direction of motion in patient positioning. All analysis metrics described above were repeated for the "worst case" alternate position, meaning the alternate position for each patient that gave the largest displacement of the center of mass. Potential geometric misses using the alternate, uncorrected GTV were also again determined using the same criteria as for the original, uncorrected GTV.

Results

Twenty-eight intracranial radiosurgery cases were identified for inclusion in this study (Table 1). The median target volume was 995.7 mm^3 (interquartile range 261.6 – 4703.1

mm^3), with a corresponding median equivalent sphere diameter of 6.2 mm (interquartile range 3.9 – 10.4 mm).

Distortion from gradient non-linearity caused a maximum displacement of targets of up to 3.9 mm (Table 2). The median displacement was 1.2 mm. While twenty-five percent of cases had displacement less than 0.1 mm, the worst twenty-five percent of cases had displacement greater than 2.3 mm. Visual examples of how contours are affected are demonstrated in Figure 1. When small shifts in patient MRI position were allowed, distortion for the least favorable alternate positions produced a median displacement of 1.9 mm, and the worst twenty-five percent of cases had displacement greater than 3.4 mm. Allowing for reasonable alternate positions, the greatest displacement measured in this study was 7.8 mm (Table 2).

Displacement of the target (in 3D) was significantly correlated with its Euclidean distance (in 3D) from the center of the magnetic field of the MR system. The correlation coefficient was $r = 0.79$ ($p < 10^{-6}$). A scatter plot showing the results is shown in Figure 2.

Correction of distortion in 2 dimensions alone only partially reduced the maximum displacement (Table 2). The median reduction with 2D-only correction was 0.05 mm. Only 2 (of 28) targets had a reduction in maximum potential distortion greater than 0.5 mm. The target subject to the most displacement with no distortion correction (3.9 mm) was still displaced by 3.8 mm when only 2D correction was applied.

Comparing the volumes from uncorrected and corrected MRI, respectively, the median Dice similarity coefficient was 0.86 for MRI with the patient in his or her original position (Table 2). The first and third quartiles for Dice similarity were 0.79 and 1.00, respectively. The most concerning case by this measure had a Dice coefficient of 0.15. Among least favorable alternate MRI patient positions, the median displacement was 0.79, with first and third quartiles of 0.60 and 0.92, respectively. The most concerning case had a Dice coefficient of 0.00, indicating there was no overlap between the corrected and uncorrected volumes.

Table 3 shows the number of cases (out of 28 total) that were deemed inadequate by various criteria. Twenty of the twenty-eight true targets in this study failed to meet the criterion for SRS plan adequacy in our routine practice, namely that at least 98% of the planning target receive the full radiation dose as prescribed. A more lenient standard was applied to define geometric misses. As described above, a true target was considered “missed” in the plan if more than 10% or more than 125 mm^3 of its volume received less than 90% of the intended dose. By the 10% criterion, four of the cases represented a geometric miss. An additional four cases met the 125 mm^3 criterion, bringing the total number of geometric misses to eight of twenty-eight cases. When alternate MRI positions were explored, it was found that ten cases met the 10% criterion for geometric miss, and ten cases met the absolute volume criterion (Table 3). Some of these cases failed by both criteria, so the total number of cases at risk for geometric miss increased to thirteen of twenty-eight with only slight shifts in patient position at time of MRI acquisition.

Radiosurgery plans generated using the uncorrected MRI target gave excellent coverage of the uncorrected target used in the plan, but were often inadequate for coverage of the true target. All radiosurgery plans met the optimization criterion of at least 98% of the uncorrected (planning) target receiving at least 100% of the prescription dose (Online Resource 1). However, the percent of the true target receiving the full prescription dose was a median of 94.5%, with interquartile range of 84.4% – 98.0%. One case (the smallest target volume in the study, at 11 mm^3) had only 1.8% of the true target volume receiving the prescription dose. In the analysis with alternate MRI patient positions, the median percent of true target receiving the prescription dose was 86.0%, with interquartile range 71.5% – 95.0%. The case with the smallest target volume had the true target completely outside the 100% isodose line.

Discussion

Distortion from gradient non-linearity in MRI can be sufficient to cause geometric miss of brain targets in intracranial stereotactic radiosurgery. Our findings demonstrate that using uncorrected MRI to delineate brain targets may yield contours that are displaced from the actual target position by several millimeters. This finding is consistent with the degree of distortion measured in previous MR studies [6, 10–12]. For highly conformal plans, such as those used in SRS, our findings suggest that those several millimeters are enough to result in considerable under-dosing of the true target or tumor. In the absence of distortion correction for gradient non-linearity, these shifts may go undetected by physicians and physicists, and resulting geometric misses might be interpreted as treatment failures because SRS planning software will report that the (displaced) target was well covered by adequate radiation doses.

In the absence of distortion correction, 8 of 28 lesions studied would be "missed" (i.e., undertreated) using the fairly liberal criteria that more than 10% or 125 mm^3 (equivalent to a cube with 5 mm sides) of the true tumor received less than 90% of the prescription dose. In a recent cooperative group clinical trial for SRS, any portion of the target volume receiving less than 90% of the prescription dose was considered a Major Deviation from protocol; the geometric misses highlighted in the present study would therefore represent gross violations beyond the minimum threshold for Major Deviation [27]. When slight adjustments to the patient's head position in the MRI system were explored, an additional 5 lesions were found to be at risk of frank geometric miss by these criteria (i.e., the lesion would have been missed in at least one alternate head position), bringing the total number of potentially missed lesions to 13 of 28. Moreover, the maximum potential displacement of a tumor was measured at nearly 8 mm. Patient positioning is often not tightly controlled when obtaining diagnostic MRI, and the present results illustrate the significant impact even slight position changes can have on the magnitude of distortion at the site of the target or tumor. There is no *a priori* ideal position for the patient's head for purposes of minimizing distortion near the target, especially since the location of the treatment target is often not known before the MRI is obtained. The safest way to avoid missing targets with SRS is to correct for gradient non-linearity distortion.

These results are the first to demonstrate underdosing of brain lesions with linear accelerator-based SRS due to the isolated effect of gradient non-linearity distortion. A 2010 study using cobalt-based radiosurgery on a spherical phantom predicted that MRI distortion could lead to underdosing of SRS targets compared to plans using CT [28]. Similarly, a 2014 study, also using cobalt-based SRS, found that uncorrected distortion due to B0 field inhomogeneities can lead to underdosed tumors [29], but it did not assess the effect of gradient non-linearity, which is often the dominant source of distortion in standard MRI [12, 16]. This is in contrast to some older studies that measured MRI distortion and concluded the magnitude of the distortion was not problematic for radiotherapy planning [30]).

The AAPM has an active Task Group (number 117) that is discussing quality control procedures for spatial accuracy for MRI-guided SRS, but a formal guideline is not currently available. A previous AAPM report in 2005 warned that users should verify whether the stereotactic software they employ is able to detect and correct MRI distortion, as not all commercial platforms have these capabilities [31]. For acceptance testing of MRI facilities, a 2010 AAPM report recommends the measurement of the diameter of a phantom on MRI should be within 2% of the true diameter [32]. The American College of Radiology (ACR) MR Accreditation Phantom has a diameter of 190 mm, so 2% implies expected errors up to 3.8 mm in either direction, which is similar to the distortions causing geometric miss in our study. For its own accreditation, the ACR limits errors to within ± 2 mm [33], but this is still only for a simple measurement of a phantom diameter.

Even when an MR system can demonstrate compliance with the phantom measurements described by the AAPM and ACR, meaningful distortion may still be present. First, the diameter of a phantom is hardly comprehensive, as the pattern of distortion in MRI is neither homogeneous nor linear. Second, as AAPM Task Group 1 notes, the phantoms used for the measurements typically cannot be rotated vertically within the phased-array head coils used for modern brain imaging, so only one special case of phantom position is being tested [32]. Due to the position and location dependent nature of MRI distortion, it is unlikely a phantom-based quality assurance program can be devised to give geometric accuracy superior to prospective correction of the distortion in images from individual patients.

While manufacturer-supplied software is available in some cases to correct for gradient non-linearity, 3D correction is a relatively new offering by the major vendors of MR systems. Further, to our knowledge, these algorithms have so far all been limited to MR data acquired in 3D, rather than the 2D slice acquisition more common in diagnostic radiology. Therefore, 3D correction by vendor-supplied algorithms is really only available to those who have recently purchased it and are careful to ensure MR images used for SRS planning were acquired with a 3D pulse sequence. This is an excellent option that will hopefully become more prevalent with adoption of the newest technology.

Vendor-supplied 2D-only gradient non-linearity correction is available on many systems, including the one used for the present study. 2D-only correction is helpful for qualitative assessment in diagnostic radiology, but it is an intuitively problematic approach in radiotherapy, where 3D accuracy is important. We have shown here that 2D-only correction may be inadequate for the prevention of meaningful shifts in SRS targets because the maximum displacement can remain large. This is consistent with other reports in the literature. For example, a recent study using an MR phantom found that 3D correction was superior to 2D-only correction. The authors also made the important observation that 3D correction actually also improves the in-plane slice accuracy over 2D-only algorithms because the 2D algorithms depend on the estimated 3D distance from isocenter to calculate distortion effect [25]. Another phantom study found that 2D-only correction merely reduced maximal distortion in their experimental conditions from 3.4 mm to 3.2 mm, numbers that are quite similar to those reported in the present work [9].

One alternative to correcting for distortion could be to increase the treatment margin to account for it. The primary disadvantage to this approach is the increased planning target volume, which for most brain metastases may be normal tissue that might otherwise be spared. It is also difficult to predict the appropriate margin given the differences in magnitude of distortion between different cases. In our original analysis, the largest displacement of a GTV center of mass was close to 4 mm. For example, to add a symmetric margin of 4 mm to a spherical tumor with 1 cm diameter implies a 6-fold increase in target volume. The situation is even worse when we consider alternate patient positioning at time of MRI scan, where an 8 mm margin might be required, implying a nearly 18-fold increase in target volume for a 1 cm diameter tumor.

Other options are similarly unsatisfying. When comparing CT and MRI, if distortion can be appreciated by the radiation oncologist or physicist, attempts can be made to compensate with manual adjustments to improve registration locally near the tumor. Unfortunately, this would likely prove unreliable and difficult to verify for quality control. Moreover, the effect of these manual adjustments on avoidance structures or additional metastases would be unpredictable. Finally, one might attempt to predict which patients are at risk of meaningful distortion of targets (or organs at risk) in order to refer these patients to a center using distortion correction. Unfortunately, this type of prediction is difficult, as the degree of distortion is dependent on the position of the various structures within the particular MRI gradient fields and only indirectly on patient-specific characteristics such as anatomic location of a tumor. The best predictor of gradient non-linearity effect is likely the distance from isocenter of the target or organ of interest. This method is probably only feasible on MR systems using isocentric acquisition (such that the center of the static B0 field and the gradient field coincide), with only a single body region imaged in the sequence. In these conditions, if the distance from magnet isocenter is deemed small, then the risk of large gradient non-linearity distortion is likely also small. However, if there are multiple structures of interest (e.g., organs at risk, additional targets), then this process would need to be performed for each structure, taking the maximal distance from isocenter in each case. This is a potentially tedious, though feasible, approach to identifying patients who require corrected MRI.

Intuitively, it might be expected that the risk of missing a target completely is highest when the target volume is small. However, the risk of missing tumor tissue is really a function of displacement more than anything else. A small tumor may be located near the magnet isocenter and experience virtually no distortion, therefore incurring a low risk of distortion-related geometric miss, whereas a large tumor that extends to the periphery of the brain may be subject to a relatively large displacement. Where the target size does seem to play a role is in which criterion for geometric miss it is more likely to violate. Targets in this study where >10% of the volume was underdosed were more likely to be smaller (median volume 213 mm^3, compared to a median of 996 mm^3 across all patients). Conversely, targets where the absolute volume of underdosed GTV was >125 mm^3—not surprisingly—tended to be larger (median volume 5858 mm^3). For example, one patient had a target whose volume was 5348 mm^3, and though 95% of the target received at least 90% of the prescription dose without distortion correction, the underdosed portion was still 267 mm^3—larger than the entire GTV for 7 of the other patients in the study. Accepting this plan would be arguably equivalent to adequately treating a large metastasis but completely underdosing a smaller satellite lesion. Minimizing displacement due to distortion should be the primary goal to ensure adequate target coverage; correcting for the distortion itself is the most direct approach to that end.

The distortion correction method used in this study has been extensively validated in previous work. Technical aspects of the method are described elsewhere in detail, along with the theoretical background and a demonstration of the impact on geometric accuracy [22]. The method was further validated on MR systems from four large, academic medical centers using both phantom and human data [13]. Using phantom data and MR devices from two major vendors, the authors showed that measurement of the diameter of a 220 mm phantom in uncorrected images could be off by more than 15 mm, and 2D-only correction (applied by default by the scanner software for some of the devices in the study) only reduced maximal error to approximately 13 mm. With 3D correction, the phantom diameter measurement was accurate to within 1 mm. The gradient non-linearity distortion method used in the present paper has been adopted as the standard for multi-institution cooperative imaging initiatives, including Morphometry Biomedical Informatics Research Network (BIRN: www.birncommunity.org/tools-catalog/gradient-non-linearity-distortion-correction) and the Alzheimer's Disease Neuroimaging Initiative [23]. The method has also been adopted for vendor-supplied distortion correction features provided with MR systems [9, 12, 25].

Limitations of this study include that the cohort represents a sample from a single institution. Therefore, the results cannot be used to make extrapolations to predict the prevalence of the problem in the general population, though the degree of distortion measured here is consistent with previous studies [6, 10, 11]. To avoid differences in the magnitude of distortion across different scanners, our data all come from one scanner—though it is a modern MR system from a popular vendor, and gradient non-linearity is known to affect all MRI [4, 5, 8, 9, 13]. Of note, distortion from gradient non-linearity is only one source of geometric distortion in MRI. Other sources, such as static field inhomogeneity and eddy currents are also known to affect image accuracy, and these are generally dependent on the sequence used and the nature of the tissue interfaces imaged, whereas gradient non-linearity is a function of the MR system itself, regardless of the sequence chosen [4, 5]. Efforts to address sequence-specific distortion have had some success though residual distortion is often dominated by effects of gradient non-linearity [16, 19, 29, 34, 35]. The present investigation focuses on the clinical significance of gradient non-linearity effects because correction of this source of inaccuracy has not been widely adopted in clinical practice [16, 17]. Despite these limitations, our data demonstrate that gradient non-linearity distortion in MRI can have a meaningful impact on SRS accuracy.

This study was limited to intracranial SRS planning, chosen because of the tight margins and requirement of high precision in these treatment plans. However, the principles of distortion in MRI apply to radiation targets and normal structures anywhere in the body, as gradient non-linearity will affect all tissues. In fact, MR images of other body sites (e.g., liver or pancreas) are vulnerable to greater distortion effects because the size of the body implies some tissues will be farther from the center of the MR bore. Further investigation is necessary to adequately describe the clinical significance of MRI distortion on stereotactic and non-stereotactic treatments for any targets using MRI, whenever the treatments require precise anatomic accuracy.

Conclusions

Gradient non-linearity causes distortion in MRI that can displace intracranial targets by nearly 4 mm, with the potential for displacement of up to 8 mm. This distortion is sufficient in some cases to cause geometric miss of SRS targets. Correction of MR images for gradient non-linearity distortion should be implemented as a routine step for accurate SRS planning. Each radiation oncology center should review its own process to ensure anatomic accuracy of MRI for stereotactic radiosurgery, including discussion with radiology departments, where applicable.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

Acknowledgments

This project was partially supported by the following grants: National Institutes of Health Grant KL2RR031978, for years 1 & 2 of CTSA funding, KL2TR00099 during year 3 and beyond of CTSA funding (JHG); and UL1TR000100 (JHG); American Cancer Society Pilot Award ACS-IRG #70-002 (JHG); a research grant from Varian Medical Systems and UC San Diego (JHG; TMS); and a UC San Diego Translational and Clinical Pilot Project Award (CRM; JHG). The content is solely the responsibility of the authors and does not necessarily represent the official views of the NIH, the American Cancer Society, UC San Diego, or Varian Medical Systems. The authors would like to thank Kelly Leyden for assistance in organizing and maintaining the imaging database.

Tyler M. Seibert is a Co-Investigator on a grant funded by Varian Medical Systems and UC San Diego.

Vitali Moiseenko reports prior honorarium and travel fees from Varian Medical Systems for a talk outside the submitted work.

Dominic Holland has a patent US8160319 (April 17, 2012) issued.

Anders M. Dale reports grants from National Institutes of Health (NIH); grants, non-financial support and other from General Electric Healthcare (GEHC); equity interest in CorTechs Labs, Inc.; service on the scientific advisory boards for CorTechs Labs, Inc. and Human Longevity, Inc. In addition, Dr. Dale has two patents (US20120280686 and US8160319) licensed to GEHC. Jona A. Hattangadi-Gluth is the Principal Investigator on a grant funded by Varian Medical Systems and UC San Diego.

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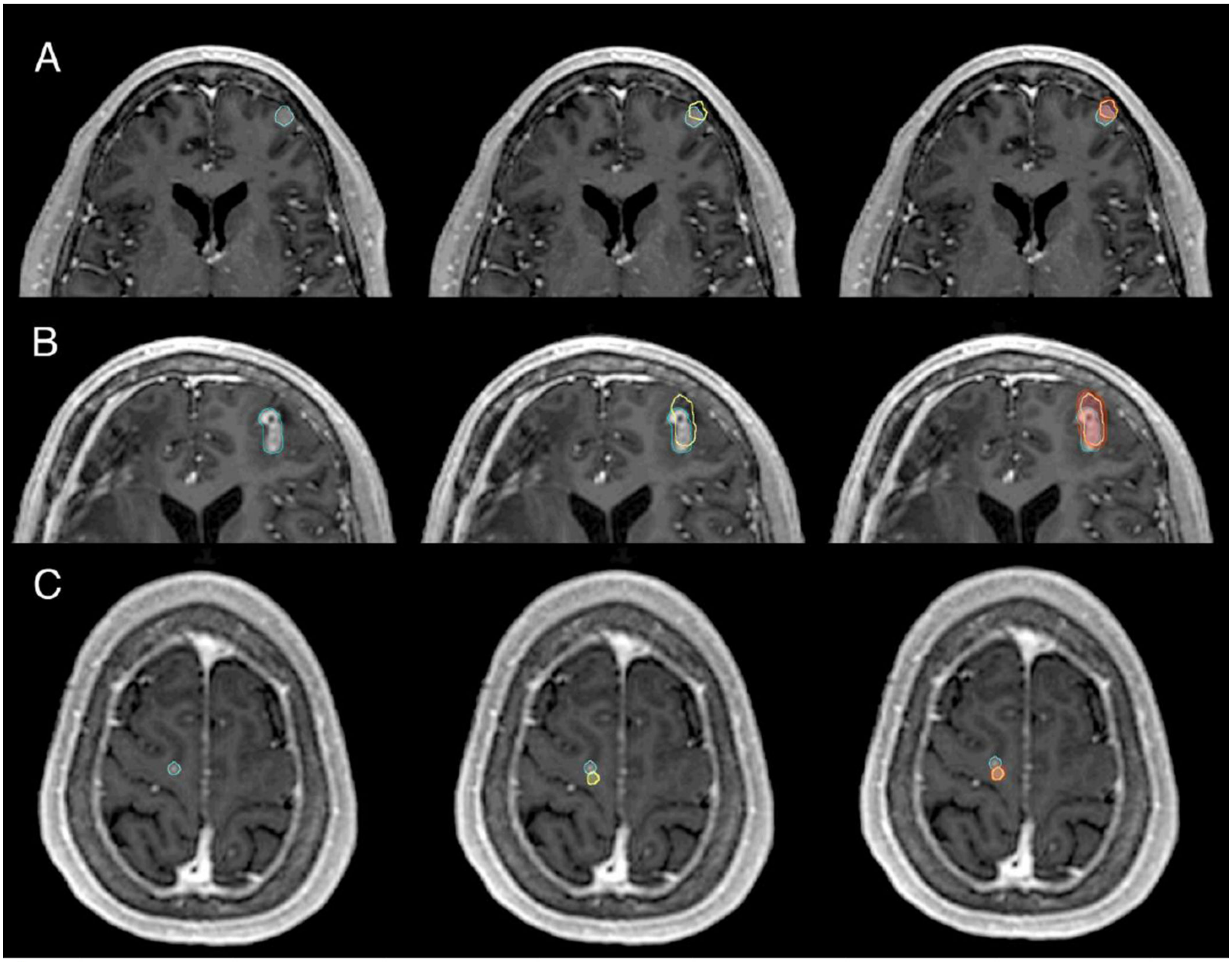

Fig. 1.

T1 post-contrast MRI for 3 example patients (A–C). True (corrected) GTV is outlined in cyan as contoured (left pane). Uncorrected/distorted GTV is outlined in yellow (middle and right panes). Displacement of centers of mass: 3.4 mm (A), 3.9 mm (B), 3.9 mm (C). Volume of true GTV **not covered** by ≥ 90% of prescription dose (shown in red, right pane): 28%, 96 mm^3 (A); 11%, 75 mm^3 (B); 75%, 62 mm^3 (C).

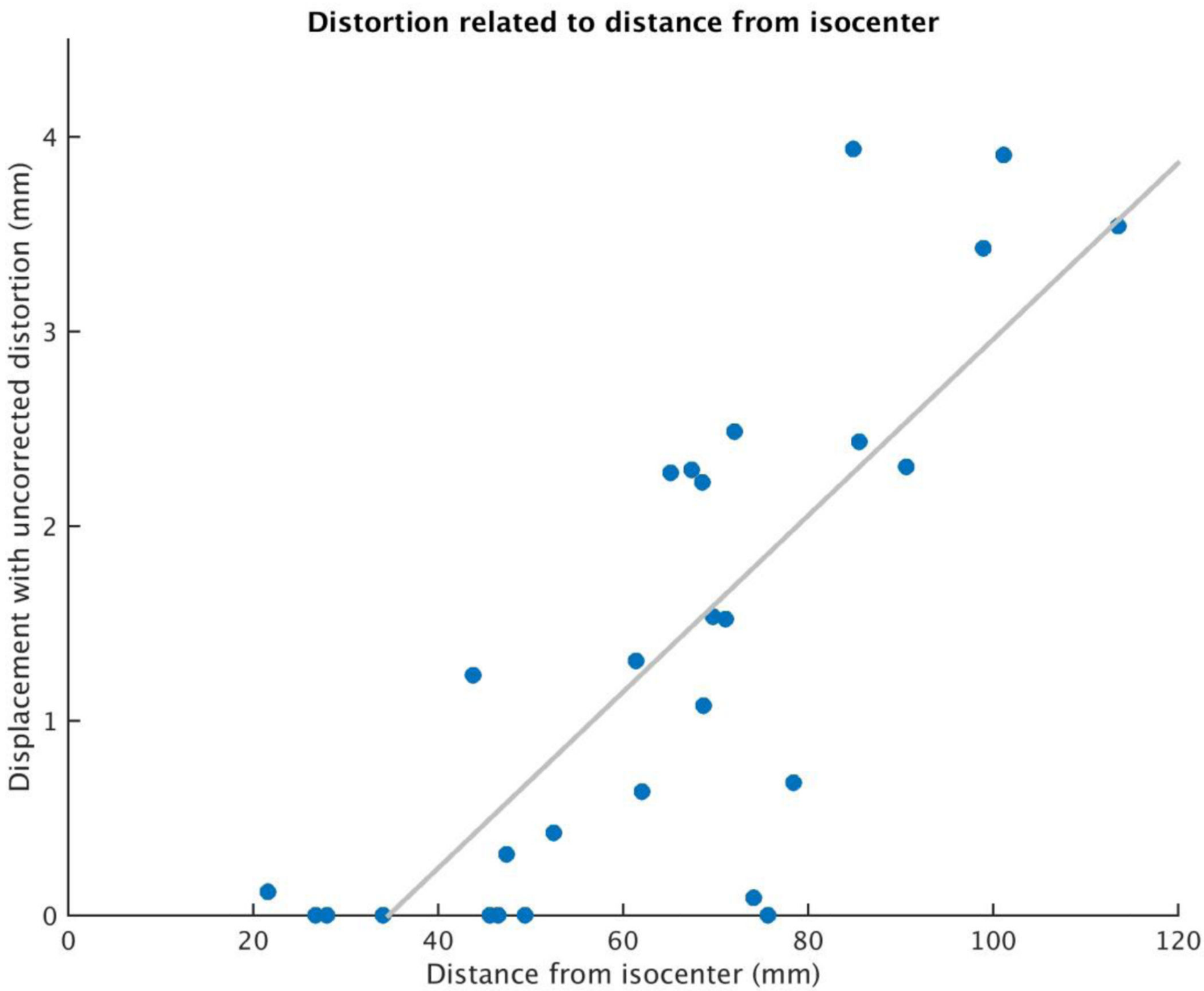

Fig. 2.

Distortion related to distance from isocenter. Displacement from distortion was measured as the Euclidean distance (in 3D) between the distorted and corrected positions of the center of mass of the target volume. Distance from isocenter was also measured for the target volume center of mass as a Euclidean distance in 3D. Each point on the scatter plot represents one study patient. Gray line is the least squares regression line. Correlation coefficient $r = 0.79$ ($p < 10^{-6}$).

TABLE 1

Cohort Characteristics (n = 28)

Descriptor	Number (%)	Median (IQR)
Tumor Location (Lobe)		
Frontal	14 (50%)	-
Parietal	5 (20%)	-
Cerebellum	4 (16%)	-
Temporal	2 (8%)	-
Occipital	2 (8%)	-
Brainstem	1 (4%)	-
Target Volume (mm3)	-	995.7 (261.55 – 4703.1)
Target equivalent sphere diameter (mm)	-	6.2 (3.9 – 10.4)

TABLE 2

Effect of Distortion on Target

Effect of distortion on target				
Distortion with patient in original position[††]				
	Median	1st quartile	3rd quartile	Worst case[§]
Displacement without correction[*]	1.2 mm	0.1 mm	2.3 mm	3.9 mm
with 2D-only correction	1.1 mm	0.1 mm	1.9 mm	3.8 mm
Dice similarity coefficient (0–1)[†]	0.86	0.79	1.00	0.15
Distortion with patient in alternate position[//]				
	Median	1st quartile	3rd quartile	Worst case[§]
Displacement without correction[*]	1.9 mm	0.5 mm	3.4 mm	7.8 mm
Dice similarity coefficient (0–1)[†]	0.79	0.60	0.92	0.00

[*] Displacement without correction = distance in millimeters that the center of mass of the GTV moves when MR images are corrected for distortion

[†] Dice similarity coefficient = degree of overlap between the uncorrected and corrected GTV (1 indicates full overlap; 0 indicates no overlap)

[††] Original position = refers to the actual MR images with the patient in whatever position he or she was scanned in

[//] Alternate position = refers to images modified to reflect a hypothetical, small change in the patient's positioning in the MRI system

[§] Worst case = results of the SRS case with the most severe distortion (i.e., maximal displacement)

TABLE 3

Effect of Distortion on Target Coverage and SRS Plan

Inadequacy of true target coverage with plans using uncorrected (distorted) target

	Criterion for inadequate plan	No. of cases (n=28)	Median *	Maximum *
	Distortion with patient in original† position			
(A)	>2% of target received <100% Rx dose	20	12.3%	98.2%
(B)	>10% of target received <90% Rx dose	4	43.0%	74.7%
(C)	>125mm^3 of target received <90% Rx dose	4	258 mm^3	427 mm^3
	Geometric miss by criteria (B) or (C)	8	-	-
	Distortion with patient in alternate†† position			
(A)	>2% of target received <100% Rx dose	25	17.8%	100.0%
(B)	>10% of target received <90% Rx dose	10	21.6%	92.4%
(C)	>125mm^3 of target received <90% Rx dose	10	359 mm^3	1966 mm^3
	Geometric miss by criteria (B) or (C)	13	-	-

*The median and maximum values (across patients) for each criterion are shown

†Original position=refers to the actual MR images with the patient in whatever position he or she was scanned in

††Alternate position=refers to images modified to reflect a hypothetical, small change in the patient's positioning in the MRI system

//Geometric miss=a plan is considered a geometric miss if it fails by either criterion (B) or (C)