An Image Correction Protocol to Reduce Distortion for 3-T Stereotactic MRI

**BACKGROUND:** Image distortion limits application of direct 3-T magnetic resonance imaging for stereotactic functional neurosurgery.

**OBJECTIVE:** To test the application of a method to correct and curtail image distortion of 3-T magnetic resonance images.

**METHODS:** We used a phantom head model mounted on a platform with the dimensions and features of a stereotactic frame. The phantom was scanned within the head coil of a Philips Achieva 3T X series (Philips Medical Systems, Eindhoven, the Netherlands). For each scan, 2 images were obtained—the normal and the reversed images. We applied the inverted gradient correction protocol to produce a corrected $x$, $y$, and $z$ coordinates. We applied the Cronbach test or coefficient of reliability to assess the internal consistency of the data.

**RESULTS:** For all analyzed data, the $P$ value was $>.05$, indicating that the differences among the observers were not statistically significant. Moreover, the data rectification proved to be effective, as the average distortion after correction was 1.05 mm. The distortion varied between 0.7 mm and 3.7 mm, depending on the target location.

**CONCLUSION:** This study examined a rectifying technique for correcting geometric distortion encountered in magnetic resonance images related to static field inhomogeneities (resonance offsets), and the technique proved to be highly successful in producing consistently accurate stereotactic target registration. The technique is applicable to all routinely used spin-echo MRI.

**KEY WORDS:** Distortion, Inverted sequence magnetic resonance, Stereotactic technique

**ABBREVIATION:** RF, radiofrequency
3-T magnets. The current work tested algorithms for the correction of image distortion in 3-T MRI.12

PATIENTS AND METHODS

A phantom head model (Micromar, São Paulo, Brazil), consisting of an acrylic resin cylinder mounted on a platform with the dimensions and features of a stereotactic frame (Figure 1), was manufactured. A system of 3 fiducial markers was attached to the phantom frame to provide 9 points for target calculation. Each fiducial marker consisted of an N-shaped tubular silicone channel filled with a 3% boric acid solution to provide the best magnetic resonance (MR) signal. The Micromar Stereotactic Frame System was used for imaging acquisition and stereotactic localization (Hitchcock frame modified). The Frame System is manufactured from Celeron, a material produced from phenolic resin and cotton wool. The system was nationally developed and is the most widely used in Brazil. The combination of the fiducial markers and the imaging system allowed each image slice, obtained from the phantom, to contain 3 reference points on the right/left and superior sides of the image (Figure 2). During image acquisition, the phantom was positioned exactly as a patient would be.

The signal in MRI comes from the interactions between the magnetic moments of certain nuclei in the object and the magnetic field generated by the imager. Because of their strong signal and abundance in living tissue, the protons of the hydrogen atom are the predominant nuclei of interest in most MRI scans.5-7 These protons are excited by a radio-frequency (RF) field oscillating at a frequency tuned to match that of protons immersed in a static magnetic field of a specific magnitude B. In the most common imaging technique, “slice selection,” a carefully controlled, highly linear gradient, Gz, in the z direction is impressed on the static field, so that only protons in a plane perpendicular to the z axis experience the same value of B. An RF pulse of a specific frequency applied in the presence of Gz excites only protons lying in such a plane. If, however, there is some nonuniformity to the static field, the excited protons lie in a warped surface. 5-7 We will let Be (x, y, z) be the deviations of the static field from nonuniformity. Be might be caused by imperfections in the main magnets of the imager or by the magnetizability of the object being imaged. To express the warp due to Be, we will let x1, y1, z1 be the incorrect coordinates provided by the imager and x, y, z be the actual coordinates of the protons.5-7

\[ z_1 = z + \frac{B_e(x,y,z)}{G_z} \quad (1) \]

The derivations of this equation and the following one are provided elsewhere.5-7 These excited protons each produce a sinusoidal signal with the same phase. Their phases are deliberately dispersed by the application of a gradient Gx in the y direction of a finite time. This dispersion in the presence of a linear gradient effectively encodes the y coordinate of each proton into its phase.5-7 Finally, a signal is recorded during the application of a third gradient Gy in the x direction. This signal is the sum of signals from protons whose frequencies are, because of Gx, a linear function of x. By repeating the process of slice selection with Gx, phase encoding with Gy, and readout of signal in the presence of Gz, for many different values of Gz, the imager obtains a set of signals all of which arise from the same (warped) slice of the object.5-7

A 2-dimensional Fourier transform of this signal produces a distribution of signal in x, y that represents the distribution of the magnetic field M of the protons. There is no distortion in the y coordinate arising from Be, but there is distortion in x:

\[ x_1 = x + \frac{B_e(x,y,z)}{G_x} \quad (2) \]

In addition to the distortion in position, there is a distortion in image intensity. If we let i(x, y, z) represent the undistorted image and i'(x1, y1, z1), the distorted image, then
\[ i_1(x_1, y, z_1) = i(x, y, z)/J_1(x_1, y, z_1), \]  

(3)

where \( J_1(x_1, y, z_1) \) is the Jacobian of the transformation from \( x, y \) to \( x_1, y, z_1 \), and we have used \( y_1 = y \). It can be shown that

\[ J_1(x_1, y, z_1) = 1 + \frac{1}{G_x} \frac{\partial B_x(x, y, z)}{\partial x} + \frac{1}{G_z} \frac{\partial B_z(x, y, z)}{\partial z}. \]

(4)

It is possible to correct the errors in \( x, y \), and the intensity described by these equations. This procedure is called rectification.\(^5\)\(^,\)\(^6\) We describe 2 rectification techniques, one called image rectification, and one called point rectification.\(^5\)\(^,\)\(^7\) The former technique is described in detail elsewhere.\(^5\)\(^,\)\(^7\) A considerable simplification is possible in both rectification techniques because of the typically large value of \( G_x \). From Equations 1, 2, and 4, it can be seen that for sufficiently large \( G_x \), we can ignore those terms involving \( G_z \). Thus, we now have \( z_1 = z \), and Equations 3 and 4 become

\[ i_1(x_1, y, z) = i(x, y, z)/J_1(x_1, y, z), \]

(5)

\[ J_1(x_1, y, z) = 1 + \frac{1}{G_x} \frac{\partial B_x(x, y, z)}{\partial x}. \]

(6)

For both rectification techniques, we acquire an additional image, \( i_2 \), using identical imaging parameters except for a reversal in the direction of the readout gradient. For \( i_2 \), we have

\[ x_2 = x - \frac{B_x(x, y, z)}{G_x}, \]

(7)

\[ i_2(x_2, y, z) = i(x, y, z)/J_2(x_2, y, z), \]

(8)

\[ J_2(x_2, y, z) = 1 - \frac{1}{G_x} \frac{\partial B_x(x, y, z)}{\partial x}. \]

(9)

Combining Equations 2 and 7, we find that

\[ x = (x_1 + x_2)/2. \]

(10)

Point rectification is based on this equation.\(^5\)\(^,\)\(^7\) We identify a point, such as an N-bar cross section or a target, in each image, and we use the average of these 2 positions as the rectified point. In image rectification, we combine Equations 2 and 5 through 9 to obtain a rectified image,

\[ i(x, y, z) = \frac{2i_1(x_1, y, z)i_2(x_2, y, z)}{i_1(x_1, y, z) + i_2(x_2, y, z)}. \]

(11)

We identify points directly on this image.\(^5\)\(^,\)\(^7\) However, for practical purposes, we performed only point rectification because stereotactic interventions are based in defined point coordinates.

The phantom was scanned within the head coil of a Philips Achieva 3T X series (Philips Medical Systems, Eindhoven, the Netherlands). The MR scans consisted of T1-weighted, transverse, spin-echo slices. The repetition time (TR) was 600 ms, and the echo time (TE) was 15 ms; the field of view was 380 mm; a 256 × 256 pixel array was acquired; a slice thickness of 4 mm with a 1-mm gap was used. For each scan, 2 images were obtained—the normal and the reversed images. The correction protocol was performed as previously described for the 1.5-T system.\(^7\) A computed tomography (CT) scan was used for MR geometric accuracy calculations. The CT scan images were obtained using the High Speed CT scanner (GE Medical Systems, Milwaukee, Wisconsin). The instrument was set to obtain 2-mm axial slices, without a gap, using a 512 × 512 matrix, with a 0.65 × 0.65-mm pixel array.

To ensure internal consistency and limit possible divergence among observers, the Cronbach and Friedman tests were applied, respectively.\(^8\) Eleven volunteers were selected to measure the 3 marks (\( B, C, \) and \( D \)) that were clearly identifiable in the CT and MR (Figure 2) images. Through these measurements, the \( x, y \), and \( z \) coordinates for each point were obtained. The image analysis was conducted using MSA 3.50 software (Micromar Software Assistant; Micromar).

### RESULTS

Previous authors\(^6\) concluded that clinically relevant MR distortion occurs only in the laterolateral (\( x \)) axis, and these conclusions were confirmed in the current study. Therefore, the statistical analysis was performed using only the \( x \) axis data. To assess the internal consistency of the data, the Cronbach test or coefficient of reliability was applied (Table 1).

The Cronbach coefficients range from 0.000 to 1.000. To understand these data, Perrin classified the range as follows: (a) 0.000-0.599, unsatisfactory reliability, meaning that 1 or more volunteers had to be excluded or evaluated separately, depending on the type of study; (b) 0.600-0.699, satisfactory reliability, implying no statistical reason to exclude a volunteer; (c) 0.700-1.000, high reliability.\(^8\) The current data achieved a calculated Cronbach coefficient of 0.999 (highly reliable). The purpose of the Friedman test is to access differences among observers. If the \( P \) value is <.05, there are differences. If \( P \) value is >.05, there are no differences, as we have shown. We applied this test because variability among observers can be a source of error.\(^8\)

Tables 2 and 3 illustrate the data obtained in the \( x \) axis for the CT and rectified MR images. Table 4 shows raw data obtained from the observers/volunteers. The Friedman test was applied on these data to assess reproducibility and possible differences among the observers/volunteers.\(^8\) For all the analyzed data, the \( P \) value was >.05, indicating that the differences among the observers

### Table 1. Cronbach Coefficient for Data Reliability

<table>
<thead>
<tr>
<th>Image</th>
<th>( \alpha ) Cronbach Coefficient</th>
<th>( P ) Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Computed tomography</td>
<td>&gt;0.999</td>
<td>&lt;.001</td>
</tr>
<tr>
<td>Normal magnetic resonance image</td>
<td>&gt;0.999</td>
<td>&lt;.001</td>
</tr>
<tr>
<td>Reversed magnetic resonance image</td>
<td>&gt;0.999</td>
<td>&lt;.001</td>
</tr>
</tbody>
</table>
were not statistically significant. Moreover, the data rectification proved to be effective, as the average distortion, after correction, was 1.05 mm (range of distortion after correction, 1.05-2.75 mm). The distortion before correction varied between 0.7 mm and 3.7 mm, depending on the target location. The distance of the target from the center of Cartesian coordinate system determined the level of distortions, as the theory predicts; the least distortion was encountered for point B (near the image center), whereas the greatest distortion was evident for point C (on the image periphery). Figure 3 illustrates normal MR, inverted MR, and the subtraction of the 2 images. As evident in the Figure 3, the distortion occurs only in the x axis or the read-out phase.

**DISCUSSION**

In 1999, when the first 3-T MRI systems became available, they were not readily used because of poor RF coil design and impractical protocols. Nevertheless, they were used in laboratories and for brain imaging for many years. However, in those restricted applications, demonstrated improvements in the signal-to-noise ratio, partial and temporal resolution, and spectral resolution were achieved using similar techniques as for the 1.5-T systems. However, with the signal-to-noise ratio gains, there was an accompanying increase in magnetic field inhomogeneity. The 3-T system’s higher resonance frequency results in an augmented interference in the RF signal that produces spurious intensity variations across the image. Therefore, one of the major shortcomings of a 3-T system is the increased RF field inhomogeneity, which is also one of the major sources of image distortion. Additionally, the extension of the gradient field nonlinearity varies considerably from system to system. Therefore, the development of a method for distortion correction is mandatory to enable 3-T MRI stereotaxy. However, one must keep in mind the different nature of each type of distortion and the influence on each x, y, z coordinate (ie, static field inhomogeneity vs gradient field nonlinearity). The first is related to resonance offsets (chemical shifts and magnetic field inhomogeneities), and it mainly affects the read-out phase. The last is produced by nonlinear electric currents that pass through the orthogonal coils. Gradient field nonlinearities produce variable types of distortion, namely, barrel aberration (present in 2- and 3-dimensional MRI) and the potato chip and bow tie effect (the last 2 only present in 2-dimensional MRI). Currently, all manufacturers have provided some form of software corrections for 2-dimensional MRI. However, this may not be universally available for 3-dimensional MRI.

Several methods have been proposed for distortion correction. As demonstrated by Chang and Fitzpatrick, phase encoding is insensitive to field inhomogeneity. Some authors have tried to take advantage of this insensitivity. Regrettably, these techniques demand long image acquisition times because they sacrifice the parallel data assemblages from the frequency encoding to derive the immunity from field inhomogeneity. Some authors provided by the phase encoding. Another concept includes the use of distortion field-based mapping. Such methods use the creation of distortion maps based on image acquisitions from a known phantom shape. Yet another method includes Fourier transformation of the images to correct intensity distortions. However, a major hindrance of these methods is the lack of geometric distortion correction, which is the main concern in stereotactic neurosurgery. Sekihara et al proposed a more comprehensive method using a spin-warp technique. By applying a pulse sequence with 2-phase encoding directions in a uniform phantom, they were able to calculate a field map. The image obtained from the object was then rectified by using a known map to account for the geometric and intensity distortions.

As Chang and Fitzpatrick pointed out, the distortion correction methods described have several disadvantages. The phantom cannot satisfactorily represent the human or animal to be imaged; hence, the variations in the field produced by the

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**TABLE 2. Mean Values for the 3 Points Marked by Each of the Volunteers (Same Points for All Volunteers) Using Computed Tomography**

<table>
<thead>
<tr>
<th>Volunteers</th>
<th>n</th>
<th>Mean</th>
<th>Significance (P)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 CT</td>
<td>3</td>
<td>24.33</td>
<td></td>
</tr>
<tr>
<td>2 CT</td>
<td>3</td>
<td>24.33</td>
<td></td>
</tr>
<tr>
<td>3 CT</td>
<td>3</td>
<td>24.33</td>
<td></td>
</tr>
<tr>
<td>4 CT</td>
<td>3</td>
<td>24.67</td>
<td></td>
</tr>
<tr>
<td>5 CT</td>
<td>3</td>
<td>24.67</td>
<td>.44</td>
</tr>
<tr>
<td>6 CT</td>
<td>3</td>
<td>24.67</td>
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</tr>
<tr>
<td>7 CT</td>
<td>3</td>
<td>24.67</td>
<td></td>
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<tr>
<td>8 CT</td>
<td>3</td>
<td>24.33</td>
<td></td>
</tr>
<tr>
<td>9 CT</td>
<td>3</td>
<td>24.33</td>
<td></td>
</tr>
<tr>
<td>10 CT</td>
<td>3</td>
<td>24.33</td>
<td></td>
</tr>
<tr>
<td>11 CT</td>
<td>3</td>
<td>24.67</td>
<td></td>
</tr>
</tbody>
</table>

*There was no significant variation among the individuals. CT(x), computed tomography x coordinate data.

**TABLE 3. Magnetic Resonance Mean x Coordinates After Image Correction (MRx)**

<table>
<thead>
<tr>
<th>Volunteers</th>
<th>n</th>
<th>Mean</th>
<th>Significance (P)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 MR x</td>
<td>3</td>
<td>25.20</td>
<td></td>
</tr>
<tr>
<td>2 MR x</td>
<td>3</td>
<td>25.23</td>
<td></td>
</tr>
<tr>
<td>3 MR x</td>
<td>3</td>
<td>25.37</td>
<td></td>
</tr>
<tr>
<td>4 MR x</td>
<td>3</td>
<td>25.22</td>
<td></td>
</tr>
<tr>
<td>5 MR x</td>
<td>3</td>
<td>25.20</td>
<td></td>
</tr>
<tr>
<td>6 MR x</td>
<td>3</td>
<td>25.20</td>
<td>.152</td>
</tr>
<tr>
<td>7 MR x</td>
<td>3</td>
<td>25.20</td>
<td></td>
</tr>
<tr>
<td>8 MR x</td>
<td>3</td>
<td>25.23</td>
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</tr>
<tr>
<td>9 MR x</td>
<td>3</td>
<td>25.22</td>
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<tr>
<td>10 MR x</td>
<td>3</td>
<td>25.22</td>
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</tr>
<tr>
<td>11 MR x</td>
<td>3</td>
<td>25.20</td>
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</tbody>
</table>

*There was no significant error among the volunteers selecting the same 3 points.
magnetic susceptibility of the object are ignored. This means that the correction compensates only for distortions other than those produced by the object being imaged. In addition, the distortions from sources other than the object may vary with time; thus, the field map may not be accurate at the time of image acquisition. Finally, if the field map is extrapolated from coordinates near fiducial sites, the coordinates away from the fiducial may limit the precision of the correction process.

The method applied in this study was the one developed by Chang and Fitzpatrick and has been previously applied, successfully, in stereotactic 1.5-T MR images. The method involves the acquisition of 2 images using identical pulse sequences, except for changes in the magnetic gradients. The 2 distorted images are then combined to produce a rectified image. This technique corrects the geometric distortions in routinely used spin-echo MR images caused by static field inhomogeneity. For pulse multidirectional diffusion sequences such as echo planar imaging, different protocols have to be applied. The reason is that each on and off field gradient transition produces eddy currents to some degree. If the eddy current (and its associated magnetic field) decays to an inconsequential value between the time of the applied field gradient transition and the image readout, a spatially dependent change in image phase will result with no discernible distortion. However, when the eddy current decays slowly, so that a residual field remains during the image readout, the field behaves like an additional spatial encoding gradient and causes distortion of the image. In conclusion, the distortion from eddy currents and phase encoding gradient (\(G_y\)) has to be taken into account on those sequences.

As demonstrated in Tables 2 and 3, the mean geometric distortion was reduced to less than 1 mm, which is an acceptable stereotactic error. However, the technique has some limitations. The correction remains greater at the center of the image than at the extremities (where the distortion can reach 3.7 mm), which has to be taken into account by the surgeon. Moreover, the time between MR and reversed MR slice acquisition may add error.

### TABLE 4. Raw Data for x Coordinate Values Including Each Type of Image Acquisition, Chosen Point, and Volunteer

<table>
<thead>
<tr>
<th>Volunteers</th>
<th>1</th>
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<th>3</th>
<th>4</th>
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<th>6</th>
<th>7</th>
<th>8</th>
<th>9</th>
<th>10</th>
<th>11</th>
</tr>
</thead>
<tbody>
<tr>
<td>CT (x)</td>
<td>B 0.1</td>
<td>0.1</td>
<td>0.1</td>
<td>0.1</td>
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<tr>
<td></td>
<td>C 36.3</td>
<td>36.3</td>
<td>36.3</td>
<td>36.3</td>
<td>36.3</td>
<td>36.3</td>
<td>36.3</td>
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<tr>
<td></td>
<td>D 36.5</td>
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<td>37.5</td>
<td>36.5</td>
<td>36.5</td>
</tr>
<tr>
<td>MR (x)</td>
<td>B 1.5</td>
<td>1.5</td>
<td>1.5</td>
<td>1.5</td>
<td>1.5</td>
<td>1.5</td>
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<tr>
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<tr>
<td>rMR (x)</td>
<td>B 0.8</td>
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<td>0.8</td>
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<td>0.8</td>
<td>0.8</td>
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<td>0.9</td>
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</table>

CT(x), computed tomography x coordinate data; MR(x), magnetic resonance x coordinate data; rMR(x), reverse magnetic resonance x coordinate data.
because the magnetic field is not perfectly stable. Furthermore, the stereotactic frame system has to be manufactured from nonmetal components. The reason is that the magnetic field inhomogeneities related to object-induced distortion would be unacceptable. Nevertheless, the simplicity of the technique makes it applicable to any situation without the need for complex or time-consuming calculations after the pulse sequence has been setup to obtain a reversed image, which, however, can only be done in open software MRI devices. Moreover, MR physicists have a pivotal role in the process because they are responsible for finding and adjusting, among thousands of program lines in the MR software source code, the specific MR program line that can reverse the gradient without changing any other parameter. Additionally, the MR physicist’s presence in the entire process is crucial for the multidisciplinary team to understand and adjust the distortions observed.

CONCLUSION

The development of 3-T MRI instruments enabled the generation of higher quality images, but they are also subject to greater distortion, limiting their utility for stereotactic procedures. This study examined a rectifying technique for correcting geometric distortions encountered in the MR images, and the technique proved to be highly successful in producing consistently accurate stereotactic target registration. The technique is universally applicable to all routinely used spin-echo MR images and corrects for geometric distortion accountable by static field inhomogeneity. The image quality was also found to be preserved throughout the rectification process. Clinical use can be practical and efficient, with automated rectification rapidly producing images for stereotactic procedures.

Disclosure

The authors have no personal financial or institutional interest in any of the drugs, materials, or devices described in this article.

REFERENCES


Acknowledgments

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COMMENTS

In this paper, the authors describe a method to correct the distortion for images acquired with a 3-T MRI. Accounting for distortion is particularly important in functional neurosurgery to ensure the accurate placement of intracranial electrodes. Novel deep brain stimulation targets that are located farther away from the center of the brain are likely to be even more affected by distortion. An algorithm to correct distortion of 3-T MR images is therefore timely. The technique described by the authors requires modifying some component of the MRI software to allow the acquisition of a reversed image along with the true image. The reversed image is created using the inverse of the magnetic gradient used for the true image. Taking advantage of the otherwise mathematical equivalence of the 2 images, a rectification of the distortion is obtained by averaging the position of their points. Based on their calculations with a phantom head, the authors found an average distortion of 1.05 mm after rectification, but up to 3.7 mm near the periphery. Although promising, this technique will require further verification in humans to determine whether the targeting accuracy is truly improved.

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As the quality of MRI improves, MRI-guided and MRI-verified deep brain stimulation, without the use of physiological recording or clinical testing, is increasing in popularity. First used in a number of European centers, 1-6 this approach is also starting to gain traction in the United States. 7,8
Therefore, this paper addresses a timely and important subject in stereotactic functional neurosurgery. Reducing image distortion is essential for accurate surgical planning to benefit from the improved tissue contrast offered by the higher field strength of 3-T MR scanners.9-10 The point rectification technique described requires a second acquisition of each MR image, identical except for a reversed read-out gradient.11 Point locations are identified separately on the forward and reverse images with the midpoint used as the corrected coordinate. By applying this technique in a phantom, the authors demonstrate the feasibility and reproducibility of the method and show a reduction in the range of point distortions after rectification by comparison with a CT reference.

MRI distortion is caused by errors between the theoretical and actual magnetic field. Distortion correction requires an understanding of the different origins of these errors. The method described here corrects for static field errors, largely caused by magnetic susceptibility of the patient’s tissues.12 In most sequences, these occur only along the readout direction and are precisely reversed by readout gradient reversal. However, as the authors note, other errors will not be eliminated by this approach. Spatial gradient nonlinearity causes distortions that depend only on position and will not change sign with read-out gradient polarity.12-14 Some MRI systems offer independent correction for this. Other MRI sequences (notably echo planar imaging) demonstrate different, more complex distortion effects.15-17 The complexity of this situation and the potential for serial correction of different errors underline the importance of close collaboration between surgical, scientific, and radiological experts in this field.

The technique used here has the potential for wide applicability. Many MRI scanners allow user control of read-out polarity, and rectified coordinates can then be calculated without specialist software. Point rectification is thus much more easily applied than full image-based correction, which would require additional image post-processing. Development for human application at 3-T will require assessment of target point reproducibility in brain tissue to confirm that a net increase in stereotactic accuracy can be achieved.

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