Effect of MR Distortion on Targeting for Deep-Brain Stimulation

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Abstract—Deep-brain stimulation (DBS) surgery involves placing electrodes within specific deep-brain target nuclei. Surgeons employ MR imaging for pre-operative selection of targets and CT imaging for designing stereotactic frames used for intra-operative placement of electrodes at the targets. MR distortion may contribute to target-selection error in the MR scan and also to MR-CT registration error, each of which contributes to error in electrode placement. In this study, we analyze the error contributed by the MR distortion to the total DBS targeting error. Distortion in conventional MR scans, both T1 and T2-weighted, were analyzed for six bilateral DBS patients in the typical areas of brain using typical scans on a 3T clinical scanner. Mean targeting error due to MR distortion in T2 was found to be 0.07 ± 0.025 mm with a maximum of 0.13 mm over twelve targets; error in the T1 images was smaller by 4%.

Index Terms—MR distortion, deep brain, targeting, registration, error.

I. INTRODUCTION

PLANNING for deep-brain stimulation (DBS) surgery commonly relies on the acquisition of both MR and CT pre-operative images. Typically, T1-weighted and T2-weighted images are acquired for the selection of targets, which are invisible in CT. Each of the MR images is then registered to a CT image that is utilized for mechanical targeting in the operating room. Common targets of DBS surgery are located within the subthalamic nucleus (STN), the ventral intermediate nucleus (VIM), or the globus pallidus internus (GPI). Because these structures are small (four to six millimeters in linear dimension in the case of the STN), DBS procedures require the placement of electrodes with millimetric accuracy at the desired target as targeting errors can translate into reduced effectiveness in controlling symptoms and/or increased levels of side effects. It is the purpose of this work to assess the contribution to targeting error arising from geometrical distortion in the MR image. We begin with a section providing some background of DBS targeting error.

II. BACKGROUND

Reports on the level of targeting errors in DBS patients have varied from a low of 1.2 ± 0.6 mm [1] to a high of 3.2 ± 1.4 mm [2]. The sources of these errors include (a) the mechanical targeting, which aims a probe at a position relative to the skull, (b) brain shift, which comprises any displacement of the target relative to the skull due to the motion of soft tissue, (c) rigid registration of the MR image to the CT image, and (d) geometrical distortion in the MR image itself.

Because of the rigidity of the skull, mechanical targeting error can be assessed by means of phantom experiments. The mechanical targeting is achieved by means of a fixture mounted to the skull via bone anchors. Approaches to this fixturing include traditional stereotactic frames, the NexFrame (Medtronic, Inc., Minneapolis, MN, USA), and the StarFix microTargeting Platform (FHC, Inc, Bowdoin, ME, USA). A 1994 phantom study of stereotactic-frame accuracies [3] reported an error level of 1.7 ± 1.0 mm for the Leksell G-frame (Elekta AB, Stockholm, Sweden), but it was based on CT technology that is now fifteen years old. A 2007 clinical study of electrode placement errors measured on DBS cases reported an error of only 1.2 ± 0.6 mm for the G-frame [1], suggesting that the accuracy level for this frame on current scanners may be better than that measured in 1994. A 2004 phantom study of the NexFrame reported an error level of 1.25 ± 0.6 mm [4], and a 2009 phantom study of the microTargeting platform reported an error level of only 0.42 ± 0.15 mm [5]. Clinical targeting errors for these devices have been measured at two to three millimeters [1, 2, 6], suggesting that the major contributions to the error observed with these devices must arise from sources other than mechanical error.

The error contributed by MR-CT registration was assessed in 1997 in a multi-center trial [7]. While that trial was not designed specifically for DBS, the results are applicable to any application which, like DBS, involves the
rigid registration of an MR of a patient to the CT of the same patient. The results of that trial showed that MR-CT registration of head images using the methods of mutual information and normalized mutual information [8], which are the same methods used for DBS today, provided a level of accuracy on the order of 0.7 mm.

The increase in error from phantoms to clinical cases, especially for the NexFrame and the microTargeting platform cannot be explained either by mechanical targeting error or by registration error and thus suggests that brain shift and MR distortion are together making a large contribution to the errors measured on patients. A recent in-vivo study of 25 patients utilized registered pre-op and post-op images to estimate the magnitude of brain shift in the DBS targeting region [9]. That study found a mean shift of 1.8 ± 0.8 mm. This measured error was independent of mechanical error, but it included error from brain shift, registration, and MR distortion. In a recent study performed on more than hundred implants [10], it was shown that the targeting error obtained with the microTargeting platform reduced from 1.99 mm to 1.24 mm when brain shift is taken into consideration. In this study, errors due to registration and MR distortions were not taken into account. It is the purpose of the present work to assess the contribution to targeting error from MR distortion alone.

III. METHODS
A. Patients and Targets

With IRB approval, six patients who underwent DBS implantation at Vanderbilt were selected for this study. All six patients had bilateral implantation. The STN was targeted in four of these patients and the VIM in the two remaining ones resulting in 12 targets with six on the left side and six on the right side.

B. Imaging

Both 3D T1-weighted gradient-echo images and multi-slice 2D T2-weighted spin-echo MR images were acquired for six patients as part of the normal planning procedure for surgical placement of electrodes for DBS at our institution. The images were acquired on a Philips Achieva 3 Tesla whole body MR scanner (Philips Healthcare, Best, The Netherlands).

A 3D magnetization-prepared T1-weighted turbo field echo (TFE) scan was acquired in 6 minutes 32 seconds with sensitivity encoding (SENSE) acceleration factor = 2.0, FOV = 256 × 170 × 256 mm in the anterior-posterior (AP), right-left (RL), and foot-head (FH) directions, numbers of acquired voxels = 256 × 170 × 256 (acquired voxel dimensions = 1 × 1 × 1 mm), flip angle = 5°, TR = 7.9 ms, TE = 3.7 ms, bandwidth/pixel = 217 Hz (readout gradient = 5.10 mT/m), inversion delay = 917 ms, inversion shot interval = 3000 ms.

A multi-slice 2D T2-weighted turbo-spin-echo (TSE) scan was acquired in 2 minutes 42 seconds with FOV = 240 × 192 × 90 mm in the AP, RL, and FH directions, numbers of acquired voxels = 400 × 256 × 45 (acquired voxel dimensions = 0.6 × 0.75 × 2 mm), flip angle = 90°, TR = 3000 ms, TE = 80 ms, bandwidth/pixel = 194 Hz (readout gradient = 7.58 mT/m, slice-selection gradient = 6.43 mT/m).

In addition to these two diagnostic scans, a field-map was generated for each patient. The acquisition for this map was a repeated 3D fast-field-echo (FFE) scan with different echo times to generate a field map, which was acquired in 9 minutes 11 seconds with a SENSE factor = 2.0, an acquired voxel size of 1 × 1 × 2 mm, FOV = 256 × 170 × 256 mm, flip angle = 30°, TR = 25 ms, TE = 7 ms, ΔTE = 1 ms, and bandwidth per pixel = 192 Hz. The following items should be noted for these three scans:

3D T1-weighted Scan
- The acquisition orientation was sagittal with frequency encode in the FH direction. Thus, readout distortion will appear in the FH direction.
- Distortion due to inhomogeneity is zero in the AP and RL directions, because they are phase-encoded.

Multi-slice 2D T2-weighted Scan
- The acquisition orientation was axial with frequency encode in the AP direction. Thus, readout distortion will appear in the AP direction.
- Distortion due to inhomogeneity is zero in the RL direction because it is phase encoded.
- The size of the distortion in FH direction, which is the slice-selection direction, is larger than that in AP direction by 17.9% because the strength of the readout gradient is 17.9% larger than the slice-selection gradient: 7.58 mT/m versus 6.43 mT/m.

3D Field Map Scan
- The acquisition orientation was sagittal with encoding directions matching those of the 3D T1-weighted acquisition.

In addition to the MR images, a CT scan was acquired for each patient on a Philips Mx8000 IDT 16 (16-slice acquisition, 120 kVp, 457 mA, 875 ms, slice thickness = 0.625 mm, pixel size 0.49 mm, FOV = 250 × 250 × 190 mm).

C. Measuring error contributions

Distortion may contribute to targeting error in two ways: First, distortion in the region of the target may cause the selected target point \( \mathbf{p} \) to be displaced from the true point \( \mathbf{p}_0 \) resulting in a displacement error \( \Delta \mathbf{p} = \mathbf{p} - \mathbf{p}_0 \). Second, when \( \mathbf{p} \) is mapped to the CT image during MR-CT registration, distortion may add a target registration error (TRE) resulting in a total displacement error:

\[
\Delta \mathbf{p}' = \Delta \mathbf{p} + \text{TRE}
\]  

In Eq. (1), \( \Delta \mathbf{p} \) is the error in position within the MR image caused by distortion before image registration. Each T1 and T2 MR image is registered to the CT image using mutual information [8], and TRE is the error added by that registration, when \( \mathbf{p} \), instead of \( \mathbf{p}_0 \), is transferred from MR
to CT by means of the registration. The former error, $\Delta p$, is the result of local distortion at $p_x$, while the latter error, $\text{TRE}$, results from the image-wide distortion pattern all of which conspire together to cause error in the MR-CT rigid registration.

We can calculate $\Delta p$ at any point in the anatomy by means of the field map along with information about the acquisition of the diagnostic image to be analyzed. As pointed out in the acquisition notes above, for both the T1 and T2 acquisitions there is a component of distortion along the readout direction, which we assign to be the $x$ direction. The signed magnitude $\Delta \alpha(x,y,z)$ of this displacement in units of millimeters at position $x,y,z$, can be calculated as follows:

$$\Delta \alpha(x,y,z) = \frac{W \Delta f(x,y,z)}{N \Delta F} = \frac{\Delta f(x,y,z)}{(\gamma/2\pi)G} \times 10^3,$$

where $W$ = width of the field of view in the readout direction (mm), $N$ = the number of pixels in the readout direction, $\Delta F$ = bandwidth per pixel (Hz), $\Delta f(x,y,z)$ = frequency shift (a signed quantity) at $x,y,z$ in the field-map image (Hz), $\gamma/(2\pi) = 4.2577 \times 10^4$ Hz/mT, and $G$ = gradient in the readout direction (mT/m).

As can be seen from Eq. (2), at each given position, $x,y,z$, the ratio of the magnitudes of readout displacement between T1 and T2 is equal to a constant—the inverse of the ratio of their respective readout gradients:

$$\Delta \alpha_{T1}(x,y,z)/\Delta \alpha_{T2}(x,y,z) = 1/r_{12},$$

where $r_{12} = G_{T1}^{(R)}/G_{T2}^{(R)}$ and $G_{T1}^{(R)} = G_{T2}^{(R)} = \text{strength of readout gradient for image of type } \alpha$. For our acquisitions, $r_{12} = 5.10/7.58 = 0.673$. For the T2 acquisition, there is a second component of displacement, $\Delta \alpha_{T2}(x,y,z)$, in the slice-selection direction. The ratio of the magnitudes of the two T2 components of distortion is also equal to a constant—the inverse of the ratio of the gradients in the respective directions:

$$\Delta \alpha_{T2}(x,y,z)/\Delta \alpha_{T1}(x,y,z) = 1/s,$$

where $s = G_{T2}^{(S)}/G_{T1}^{(S)}$ and $G_{T2}^{(S)} = G_{T1}^{(S)} = \text{strength of slice-selection gradient for the T2 acquisitions.}$ For our T2 acquisitions, $s = 7.58/6.43 = 1.18$. The displacement along the phase-encoding direction is zero, and hence $\Delta \alpha_{T1}(x,y,z) = 0$, $\Delta \alpha_{T2}(x,y,z) = 0$, and $\Delta \alpha_{T2}(x,y,z) = 0$. Thus, the ratio of the magnitudes of the vector displacement for the T1 and T2 images is

$$\Delta \alpha_{T2}(x,y,z)/\Delta \alpha_{T1}(x,y,z) = r_{12} \sqrt{1 + s^2}.$$}

For our acquisitions, the right side of Eq. (5) equals 1.04. Thus, the displacement magnitude is almost the same for the two sequences with T2 distortion being larger than T1 distortion by 4%. The displacements in the T1 images are along the FH direction, while the displacements in the T2 images are in a direction midway between the AP and FH directions (45 degrees from each).

The determination of TRE requires an indirect approach. We begin by employing Eq. (2) to produce a geometrically rectified T1 image. We produce that image by calculating the rectified intensity $I(x,y,z)$ at each point as follows:

$$I(x,y,z) = I_d(x + \Delta \alpha(x,y,z),y,z),$$

where the subscript $d$ means “distorted”, and then using the intensity to build a rectified image voxel-by-voxel. Because $\Delta \alpha$ is rarely an integral number of voxel widths, interpolation is required. In this study, we employed linear interpolation.

One approach to determining the component of registration error caused by MR distortion would be to (a) register the distorted MR to CT, (b) register the rectified (i.e., undistorted) MR to CT, and (c) compare the two transformations. However, in this case, the registration error due to MR distortion will be amplified by the error due to non-distortion sources: differences in MR and CT voxelations, noise in four images instead of two, and algorithmic errors in two registrations instead of one. While even an undistorted MR image will suffer some degree of registration error when MR-CT registration is performed, our goal is to determine as accurately as possible the component of registration error contributed only by distortion. We capture that component by the device of measuring the displacement resulting from a registration of the rectified image $I$ directly to the distorted image $I_d$. That registration was performed by using the entropy-based method of mutual information [8]. The displacement resulting from the registration at a point $x,y,z$ equals the TRE($x,y,z$) that could be expected from the MR-CT registration due to MR distortion.

IV. RESULTS

We performed the rectification of Eq. (6) and the subsequent $I-I_d$ registration, for the T1-weighted images for all six patients, calculating TRE($x,y,z$) at all points within a 10-mm cube centered on each of the twelve targets. The maximum geometric distortion over the entire head reached a maximum of 2.3 mm, but the maximum of the magnitude of TRE resulting from this error was only 0.001 mm over all target-centered cubes. To combine the data from all six patients, we used non-rigid registration to align each patient volume with a separate volume, which served as an atlas, using the method of Rohde at al. [11]. Then, we calculated the mean TRE over the six patients at each voxel within the region of head of the atlas. Figure 1 shows the pattern of this mean error via a color map (the scale is given at the right of the figure). The slow spatial variation observed in this figure is to be expected because the MR-CT registrations involve only rigid transformations, which themselves are spatially slowly varying displacements. This error is in all cases so much smaller than $\Delta \alpha$, as calculated via Eq. (2), that it can be neglected in Eq. (1).

Thus, the displacements can be determined by means of Eq. (2) alone. The resulting mean magnitude of displacement at all twelve targets is $0.071 \pm 0.025$ mm.
Figure 1. Magnitude of registration error. Left to right: sagittal, transverse, and coronal views of the brain, intersecting at the left subthalamic nucleus (not shown). Color indicates the mean MR-CT registration error at each point (scale at right).

Table 1 and Figures 2-5 summarize the results. In the table, the entries are total displacements in millimeters of the T2-weighted acquisitions for each patient and each target. Since the T2 values are consistently larger than the T1 values (4% larger), we display only the T2 displacements. Eight of the targets are STNs, and four are VIMs. To account for possible targeting inaccuracy, statistics are also reported for a 10 x 10 x 10 mm cube centered on each target. Figures 2 and 3 show the average anatomical positions of the left and right STNs, respectively, of Patients 1-4 after they were mapped onto the above described atlas. The targeted STNs are marked by an “X” in each view. While we display distortion values in our table for T2, we display these positions on the T1-weighted atlas image in Figures 2 and 3 because it depicts anatomy somewhat clearer than T2. The red outlined boxes show the volume of the atlas encompassed by Figures 4 and 5. The “X” in each of those images corresponds to the “X” that marks the targets in Figures 2 and 3. The small yellow box around each “X” shows the 10 x 10 x 10 mm cube centered on that target. In those latter images, we depict the mean displacement distance via a color map (the scale is given at the right of each figure) over this region for the left and right STNs, respectively.

V. DISCUSSION

The small size of the TRE that results from the effect of distortion on the MR-CT registration may be somewhat surprising, but it is due to the fact that rigid-body transformations are affected only by an average over local distortions. Thus, components of distortion in locally varying directions throughout the image tend to have a cancelling effect on a given angular or translational component of the rigid motion.

It should be noted that the magnitude of the MR
Table 1. Displacements due to MR Distortion in T2 acquisitions

<table>
<thead>
<tr>
<th>Patient</th>
<th>Target</th>
<th>Left (mm)</th>
<th>Right (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>At Target</td>
<td>Mean</td>
<td>SD</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>STN</td>
<td>0.065</td>
<td>0.039</td>
</tr>
<tr>
<td>2</td>
<td>STN</td>
<td>0.071</td>
<td>0.057</td>
</tr>
<tr>
<td>3</td>
<td>STN</td>
<td>0.132</td>
<td>0.108</td>
</tr>
<tr>
<td>4</td>
<td>STN</td>
<td>0.060</td>
<td>0.043</td>
</tr>
<tr>
<td>5</td>
<td>VIM</td>
<td>0.052</td>
<td>0.043</td>
</tr>
<tr>
<td>6</td>
<td>VIM</td>
<td>0.049</td>
<td>0.072</td>
</tr>
<tr>
<td>Mean</td>
<td>STN</td>
<td>0.082</td>
<td>0.062</td>
</tr>
<tr>
<td>Max</td>
<td>STN</td>
<td>0.132</td>
<td>0.108</td>
</tr>
<tr>
<td>Mean</td>
<td>VIM</td>
<td>0.050</td>
<td>0.057</td>
</tr>
<tr>
<td>Max</td>
<td>VIM</td>
<td>0.052</td>
<td>0.072</td>
</tr>
<tr>
<td>Mean</td>
<td>STN&amp;VIM</td>
<td>0.071</td>
<td>0.060</td>
</tr>
<tr>
<td>Max</td>
<td>STN&amp;VIM</td>
<td>0.132</td>
<td>0.108</td>
</tr>
</tbody>
</table>

distortion will depend on the scanning protocol, gradient strengths, sampling bandwidth, and B₀ inhomogeneity. All sources of B₀ are included in our analysis, whether from imperfections in the main magnets, shim quality, position in the scanner bore or from patient-born influences, including air-filled sinuses and dental work. The measurements presented here are for typical scans (T1 and T2) of typical patients in the typical areas of the brain used for DBS surgical planning on a typical 3T clinical scanner. For the protocols used at our institution, T2 distortion is slightly larger than T1 distortion (4%). While T2 distortion tends to be larger than T1, two factors must be considered to determine the relationship of their magnitudes for other protocols: (1) their respective readout gradient strengths and (2) whether they employ slice selection. In our study, a T1 acquisition is a volume acquisition, for which distortion is confined to the readout direction, while a T2 acquisition employs slice selection, for which there is an additional

Figure 4. MR distortion for T2-weighted acquisition. Left to right: sagittal, transverse, and coronal views of the ROI shown in Figure 2. Red cross locates the left subthalamic nucleus. Yellow box outlines the 10×10×10 mm region over which statistics are given in Table 1. The color maps the mean of the magnitude of distortion (scale at right).

Figure 5. MR distortion for T2-weighted acquisition. Left to right: sagittal, transverse, and coronal views of the ROI shown in Figure 3. Red cross locates the right subthalamic nucleus. Yellow box outlines the 10×10×10 mm region over which statistics are given in Table 1. The color maps the mean of the magnitude of distortion (scale at right).
distortion in the through-plane direction that adds to the readout distortion in quadrature. The size of the through-plane distortion depends on the magnitude of the slice-selection gradient. In every case, larger gradients provide smaller geometric distortion, both for readout and slice selection. If very fast scanning protocols, such as echoplanar imaging (EPI) are employed, distortion may increase dramatically. EPI is required to obtain time-varying functional responses from the cortex and are also employed for diffusion imaging. For these sequences, much larger distortion errors can be expected. However, for the standard acquisitions typically employed for imaging deep-brain targets, the magnitudes of the distortion shifts that we have observed are so small, that even relatively large reductions in gradient strengths or increases in $B_0$ inhomogeneity would still result in relatively small shifts.

The local distortions at the targets are much larger than the error contributed by TRE (two orders of magnitude larger), but even these relatively larger components are quite small. The maximum for all twelve targets for the T2 images was only 0.132 mm (see the entry at the bottom left of Table 1) with a mean magnitude of $0.071 \pm 0.025$ mm, and the maximum within all twelve 10-mm cubes surrounding these targets was only 0.294 mm (bottom right of Table 1) with a mean magnitude of $0.058 \pm 0.020$ mm. All these values are 4% smaller for the T1 images.

Finally, it should be noted that a second source of error, gradient nonlinearity, is not evaluated by our techniques, which are sensitive only to error due to static-field inhomogeneity. This second source also causes geometric distortion, but, unlike the first, it is consistent from patient to patient and is routinely corrected by calibration. Residual errors will remain, however, and they may add to the problem. Bounds on these residual errors are estimated during quality-assurance testing via phantom imaging. Repeated testing of the scanner used for the current study revealed that the maximum residual error due to gradient nonlinearity over a 200 mm head phantom was in every case less than 0.4 mm, which is smaller than the maximum error from field inhomogeneity measured by our method over the head, 2.3 mm, by a factor of almost six. We found that the TRE produced by inhomogeneity was negligible. Thus, the contribution of gradient nonlinearity to TRE can be expected to be negligible as well.

Furthermore, an analysis of the scanner’s algorithm to compensate for gradient nonlinearity during image reconstruction shows that its correction of local distortion within 30 mm of the center of the image is less than 0.05 mm. While the residual error after correction is not known, it is highly likely to be less than the error before correction. Thus, we will use the magnitude of the error correction as a conservative estimate of the magnitude of the residual error. DBS targets are clustered tightly within about 20 mm of each other and, as a result, can ideally be positioned in the scanner so that they lie within 10 mm of the image center and should typically fall within 30 mm. Thus, we estimate that error due to gradient nonlinearity will be less than 0.05 mm for reasonable centering. By comparison, the maximum local distortion due to inhomogeneity at the twelve targets was 0.132 mm. Thus, we estimate that distortion caused by gradient nonlinearity, while it is nonzero, can, for proper patient positioning, be ignored in comparison to the distortion caused by static-field inhomogeneity.

VI. CONCLUSIONS

The calculated displacement due to MR distortion in our in-vivo experiments using conventional imaging protocols on both T1 and T2 images on six bilateral DBS patients is no greater than 0.132 mm for all twelve targets and less than 0.3 millimeter within a 10-mm cubic region surrounding each of the twelve targets. In comparison to the other sources of error, including mechanical targeting, brain shift, and image registration, whose estimates in the literature range from one to three millimeters, such small displacements are likely to have a negligible effect on targeting accuracy in DBS surgery.

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