Thalamic surgery for tremor
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Chapter 5:
A comparison between three-dimensional MRI and ventriculography for target-coordinate determination in frame-based functional stereotactic neurosurgery

Abstract

Objective. The purpose of this study was to prospectively compare stereotactic coordinates obtained by ventriculography with coordinates derived from stereotactic computer-reconstructed three-dimensional magnetic resonance imaging (3D-MRI) in functional stereotactic procedures.

Methods. In 15 consecutive patients undergoing functional stereotactic procedures, both preoperative frame-based stereotactic 3D-MRI and intraoperative ventriculography were performed. Differences between 3D-MRI and ventriculography in X-, Y- and Z-coordinates of the anterior commissure (AC), posterior commissure (PC) and target area were calculated, as well as the 3-dimensional distance between the position of AC, PC and the target within stereotactic space as obtained by both methods.

Results. The position of the stereotactic MRI-fiducial markers measured using 3D-MRI compared well with the markers' known position embedded in the software, with a mean error of 0.4 mm and an average maximal error for an individual slice of 1.2 mm. For the individual coordinates, only for Y-PC a difference was found between 3D-MRI and ventriculography that significantly exceeded half the size of a pixel, the theoretical limit of precision when using a digitized imaging technique. However, the mean difference was smaller than 1 mm. The mean 3-dimensional distance between the 3D-MRI- and ventriculography-derived coordinates was 1.09 mm for AC, 1.13 mm for PC, and 1.29 mm for the targets.

Conclusion. These data show that there is sufficient agreement between ventriculography-derived and 3D-MRI-derived stereotactic coordinates, to justify the use of 3D-MRI target determination in frame-based functional stereotactic neurosurgery.

Introduction

There is a resurgence of interest in stereotactic functional neurosurgery, especially in the treatment of Parkinson’s disease, with many reports describing positive effects of pallidotomy, thalamotomy and continuous deep-brain stimulation of the thalamus, globus pallidus or subthalamic nucleus.¹⁻⁶ The mainstay of these procedures is accurate target determination, for which various techniques are available. Ventriculography is still used widely in target calculation but is increasingly being supplemented or even replaced by computed tomography (CT) or magnetic resonance imaging (MRI). Computed tomography has the advantage of limited inherent distortion of images, but is inferior in anatomical detail compared with MRI in stereotactic procedures.⁷ Magnetic resonance imaging provides excellent visualization of anatomical structures but has the possibility of image distortion due to gradient field nonlinearities and resonance effects. Techniques have been developed to minimize this distortion⁸⁻⁹ and reports were published using only MRI or CT-MRI fusion techniques for functional stereotaxy.¹⁰,¹¹

Three-dimensional MRI-reconstructions (3D-MRI) with frame-based stereotactic localizers have recently become available, and the purpose of this prospective study was to compare stereotactic coordinates obtained using ventriculography with coordinates derived from stereotactic computer-reconstructed 3D-MR images in frame-based functional stereotactic procedures.
Figure 1 3D-MRI for target localization in functional stereotactic neurosurgery

The MRI is made with the stereotactic frame fixed to the head, and fiducial markers (visible as dots at the perimeter of the images) attached to the frame. The 3D reconstruction provides stereotactic coordinates for the surgical target, in relation to the position of the anterior and posterior commissures.

Materials and Methods
Fifteen consecutive patients undergoing functional stereotactic procedures were entered into the study. In all patients both a preoperative frame-based stereotactic 3D-MRI and intraoperative ventriculography were performed for target calculation. Fourteen of these patients had Parkinson’s disease and underwent the following procedures for its treatment: posteroventral pallidotomy (nine patients), ventrolateral thalamotomy (two), and electrode implantation in the thalamus (two) or in the subthalamic nucleus (one). One patient had severe secondary dystonia of the arm, for which a thalamotomy was performed.

3D-MRI target localization
Lokal anesthesia was administered and the Leksell stereotactic frame (Elekta Instrument, Stockholm, Sweden) was affixed to the patients head, after which an MRI was obtained in the standard head coil by using a 1.5-Tesla MR imaging system (Vision; Siemens, Erlangen, Germany) and the Leksell MRI indicator for stereotactic localization. Image acquisition of a 3D dataset was performed using a high-bandwidth magnetization prepared rapid acquisition gradient echo (MPRAGE) sequence to minimize distortion. For the sequence, a 250-mm field of view was used, with a matrix size of 200 x 256 pixels (effective pixel size 1.25 x 0.98 mm), and a slab thickness of 170 mm, with
Figure 2 Ventriculography for target localization in functional stereotactic neurosurgery

The ventriculogram is made using a conventional X-ray after injecting contrast medium into the ventricles, with the stereotactic frame attached to the head. The position of the surgical target is calculated in relation to the position of the anterior and posterior commissures, which are visualized on the image.

170 continuous 1-mm axial T1-weighted images obtained through the cranium. The frequency-encoded readout gradient in the axial plane was anteroposterior (Y), and the phase-encoded direction was left-to-right (X).

The images were digitized, and 3D reconstruction was performed with commercially available software (Surgiplan; Elekta), which enabled direct reading of the stereotactic coordinates from the system. The positions of the anterior border of the posterior commissure (PC) and the posterior border of the anterior commissure (AC) were marked on the image, thus obtaining the AC-PC line with coordinates for AC and PC. The coordinates of the anterior border of the foramen of Monro (FM) were also determined, whereby the X-coordinate was taken as the mean of the left and right foramen.

The images were then automatically resampled using AC-PC as the central line for the axial, sagittal and coronal reconstructions, retaining the stereotactic coordinate space (Figure 1). Frame rotation relative to the AC-PC line is automatically corrected for, and can be read from the difference in the X-coordinate of the AC and the PC. Frame tilt relative to the midsagittal plane can be observed and corrected for in the reconstructed coronal images.

The target for the procedures was derived from the Schaltenbrand stereotactic atlas.\(^{12}\)
relative to the position of the AC (or FM) and PC, and entered into the system, after which target coordinates were given and marked on the images by the software. No visual correction of the target point was made. The 3D-MRI target determinations were performed by one investigator (P.R.S.) who was not involved in the subsequent ventriculography calculations.

**Ventriculography target localization**

After the MR imaging and calculations, the patient was transported to the operating room, and 10 ml of iohexol (240 mg I/ml) was injected into the lateral ventricle through a frontal parasagittal burrhole, just anterior to the coronal suture. Positive-contrast ventriculograms in lateral and anteroposterior projections were made with standard magnification using the C-arm attached to the stereotactic frame with fixed distance, after which stereotactic coordinates of AC, PC and the theoretical target point relative to the AC-PC line were calculated (Figure 2).

In two procedures the position of the AC was not visualized clearly on ventriculography. In those cases the FM was used for target calculation, with a target point relative to the FM-PC line using a Y- and Z-coordinate correction factor for the angle between FM-PC and AC-PC, which is 11.5° on average. The X-coordinate is the same whether FM-CP or AC-PC is used. All ventriculography target determinations were performed by one investigator (D.A.B.) who was not aware of the 3D-MRI coordinates obtained.

**Surgical technique**

The operation was then continued with the use of test stimulation as reported previously. Because it was our purpose to compare the theoretical targets obtained from 3D-MRI and ventriculography, and previously only ventriculography had been used in our functional stereotaxy, it was predetermined that the ventriculography-derived coordinates were to be used for the procedure.

Intraoperative testing with macroelectrode stimulation was used to confirm the optimal position for lesion or electrode placement, checking for maximal clinical effect with the lowest threshold high-frequency stimulation (130 Hz) and for absence of side effects with high- and low-frequency stimulation. After verification of the optimal position for intervention, either neuroablation or electrode implantation was performed.

**Data acquisition and statistical analysis**

Differences in X-, Y- and Z-coordinates of the AC (dX-AC, dY-AC, dZ-AC) and PC (dX-PC, dY-PC, dZ-PC) were calculated comparing 3D-MRI and ventriculography. The difference of the 3D position of AC in stereotactic space (d3D-AC) as obtained by both methods was calculated and the same was done for PC (d3D-PC). The mean, median and range of differences as well as the approximate 95% confidence intervals are reported. Because differences represent a distance, which cannot be negative, the distribution of the data was likely to be skewed. The confidence intervals were therefore calculated around the median in nonparametric analysis. Differences for X, Y, Z and the resulting 3D difference of the target (dX-T, dY-T, dZ-T and d3D-T) are reported separately. Because target coordinates were inferred from the position (that is, coordinates) of AC and PC, d3D-T is of course directly related to d3D-AC and d3D-PC.

The mean differences for all coordinates were also compared with the theoretical limit

<table>
<thead>
<tr>
<th>Coordinate</th>
<th>mean (mm)</th>
<th>median (mm)</th>
<th>range (mm)</th>
<th>95%CI (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>AC(13)&amp;FM(2)</td>
<td>0.44</td>
<td>0.4</td>
<td>0.0 - 1.1</td>
<td>0.2 - 0.6</td>
</tr>
<tr>
<td>dX-PC</td>
<td>0.38</td>
<td>0.3</td>
<td>0.0 - 1.4</td>
<td>0.2 - 0.5</td>
</tr>
<tr>
<td>dY-PC</td>
<td>0.78</td>
<td>0.5</td>
<td>0.0 - 3.1</td>
<td>0.3 - 1.8</td>
</tr>
<tr>
<td>dZ-PC</td>
<td>1.09</td>
<td>0.74</td>
<td>0.4 - 3.4</td>
<td>0.6 - 1.8</td>
</tr>
<tr>
<td>3D-difference</td>
<td>0.50</td>
<td>0.4</td>
<td>0.1 - 1.0</td>
<td>0.3 - 0.7</td>
</tr>
<tr>
<td>dX-PC</td>
<td>0.86</td>
<td>1.0</td>
<td>0.2 - 1.5</td>
<td>0.5 - 1.1</td>
</tr>
<tr>
<td>dY-PC</td>
<td>0.41</td>
<td>0.4</td>
<td>0.0 - 0.9</td>
<td>0.2 - 0.6</td>
</tr>
<tr>
<td>dZ-PC</td>
<td>1.13</td>
<td>1.19</td>
<td>0.3 - 1.8</td>
<td>0.9 - 1.4</td>
</tr>
</tbody>
</table>
of precision using a digitized imaging technique, which is half the size of a pixel ($a/2$: 0.63 mm for X, 0.49 mm for Y, 0.50 mm for Z). For this comparison $t$-tests were used against the value for $a/2$.

Results

In the process of reconstruction of the 3D-MRI from the axial MRI-slices, the system compares the measured position of the stereotactic fiducial markers on the localizer with the true values which are embedded in the software, giving a mean and maximal error for these positions across all slices. The highest mean error was 0.6 mm (average 0.4 mm, range 0.2 - 0.6 mm) and the highest maximal error in an individual slice was 1.9 mm (average 1.2 mm, range 0.7 - 1.9 mm).

The differences between 3D-MRI and ventriculography of the separate X-, Y- and Z-coordinates as well as the 3D difference of AC and PC are reported in table 1. Negative values for $dX$, $dY$ and $dZ$ in the raw data indicated a position more right-sided, posterior and superior on the 3D-MRI. For the calculations the absolute values were used to prevent underestimation of the mean differences in distance. The PC is always visualized clearly on ventriculograms; in two cases the AC was not visualized perfectly, and for these the FM was used for the calculations. Table 2 demonstrates the differences for X, Y, Z and the resulting 3D difference of the target ($dX$-T, $dY$-T, $dZ$-T and $d3D$-T). The mean 3D distance between the 3D-MRI and ventriculography derived coordinates was 1.09 mm for AC (13 patients) / FM (two patients), 1.13 mm for PC (15 patients) and 1.29 mm for the targets (15 patients).

The mean differences for all coordinates were then compared with the theoretical limit of precision when using a digitized imaging technique ($a/2$, Table 3). Only $dY$-PC was significantly higher than $a/2$ ($t$-test, $p=0.036$).

Discussion

In stereotactic functional neurosurgery for movement disorders, ventriculography is still used in many medical centers and often considered the gold standard for localization. There is longstanding experience with this method, it is relatively easy to perform, provides accurate and rapid localization of the commissures, and is free of image distortion. Image magnification can be avoided using teleradiography, or can be corrected for with standard arithmetic methods. There are a number of objections to ventriculography: imaging inaccuracies can arise if the geometric setup of radiography is not perfect, and an additional track through the brain is made, which adds potential morbidity. Intraventricular contrast medium can give rise to complications such as headache, meningismus or encephalopathy, although the neurotoxicity of current non-ionic monomeric contrast media is reportedly low.

Frame-based MRI provides direct noninvasive visualization of cerebral structures. Three-dimensional MRI reconstructions allow direct projections of the brain in axial, sagittal and coronal views with the inter-
commissural line as basis, keeping the entire volume registered in the stereotactic space. Target-structures can be obtained from stereotactic atlases and projected onto the patients' brain images, with direct translation to stereotactic coordinates. If necessary, correction can be made for frame rotation around the axial axis or frame tilt around the coronal axis, which is not possible with the method of ventriculography. An additional advantage of using 3D-MRI target localization is that the entry point of the electrode can be planned, taking care to avoid a sulcus or the lateral ventricle, the latter being of special importance in case of implantation of deep brain stimulation electrodes.

The problem of image distortion in MRI could lead to significant errors in localization. In one study of phantom imaging, considerable image distortion was measured using the Leksell frame, although this was most pronounced in the outer edges of the images. In our study, the measured position of the stereotactic fiducial markers at the perimeter compared well with their true position as embedded in the software, with a mean error of 0.4 mm and an average maximum error of 1.2 mm for an individual slice used for the 3D reconstruction. In another study of phantom imaging it was shown that, in the axial plane, stereotactic MRI precision when using the Leksell frame was limited only by pixel size, and observed errors in the coronal plane remained smaller than 1 mm. Distortion caused by object-induced magnetic field inhomogeneities on MRI is only manifest in the direction of the frequency-encoded readout gradient. Therefore, the largest discrepancies between ventriculography and MRI values were to be expected in the Y-direction. We found that \( dY-PC \) was significantly larger than half the size of a pixel, the theoretical limit of precision when using a digitized imaging technique. However, the mean difference was actually smaller than 1 mm.

In an earlier comparison of MRI and ventriculographic target localization for functional stereotaxy, a mean difference of 4.6 mm (range 0-19 mm) was found. However, in that study MRI data acquisition was two-dimensional sequential, with 4-mm slice thickness. Furthermore, in the aforementioned study the Cosman-Roberts-Wells frame was used, which was shown to cause severe distortions of up to 10 mm on MRI due to the magnetic properties of the frame, whereas the Leksell frame was found to be relatively free of geometric distortion.

Other investigators have described reliable use of 3D MPRAGE sequence imaging in combination with the Leksell frame in functional stereotaxy. In a comparative study between CT/MRI-derived and ventriculography/microelectrode-derived stereotactic coordinates, it was concluded that ventriculography is no longer required for safe and effective functional stereotaxy. In that paper the additional effect of microelectrode recordings and the patients' response to stimulation in determining the final target coordinates was not reported separately, leaving uncertainty about the comparison of CT/MRI studies and ventriculography.

Our study is the first to analyze target acquisition with 3D-MRI-derived data and to compare these data with those derived by ventriculography in a prospective, blinded fashion. The differences found between the two methods were small, especially when taking into consideration the size of the lesions produced or the electrodes implanted in these procedures. The data presented here cannot be assumed directly applicable to all circumstances, and in any clinical setting the combination of the particular stereotactic frame and 3D-MRI reconstruction technique needs to be tested for accuracy. Ensuing functional target verification with techniques such as microrecordings and micro- or macroelectrode stimulation to evaluate the clinical effect and prevent complications will remain necessary for final determination of the correct location for the intervention.

Our data indicate that there is sufficient agreement between ventriculography- and 3D-MRI-derived stereotactic coordinates, to justify the use of 3D-MRI target determination in frame-based functional stereotactic neurosurgery.
References


