Optimized magnetic resonance image-guided stereotaxis: a technique with validation based on the anterior commissure-posterior commissure line

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Some of the earliest successful frame-based stereotactic interventions directed toward the thalamus and basal ganglia depended on identifying the anterior commissure (AC) and posterior commissure (PC) in a sagittal ventriculogram and defining the intercommissural line that connects them in the midsagittal plane. The AC-PC line became the essential landmark for the localization of neuroanatomical targets in the basal ganglia and diencephalon and for relating them to stereotactic atlases.

Stereotactic functional neurosurgery has come to rely increasingly on magnetic resonance (MR) imaging guidance, and methods for accurately determining the AC-PC line on MR imaging are being developed. Our technique uses MR sequences that minimize geometric distortion and registration error, thereby maximizing accuracy in AC-PC line determinations from axially displayed MR data. The techniques are based on our experience with the Leksell G-frame, but can be generalized to other MR imaging-based stereotactic systems.

This methodology has been used in a series of 62 stereotactic procedures in 47 adults (55 pallidotomies and seven thalamotomies) with preliminary results equivalent or superior to results reported using microelectrode recordings. The measurements of the AC-PC line reported here compare favorably with those based on ventriculography and computerized tomography previously reported. The methodology reported here is critical in maintaining the accuracy and utility of MR imaging as its role in modern stereotaxy expands. Accurate parameters such as these aid in ensuring the safety, efficacy, and reproducibility of MR-guided stereotactic procedures.

Key Words * stereotactic surgery * stereotactic localization * pallidotomy * thalamotomy * Parkinson's disease * movement disorders

Beginning in the 1930s, techniques for neurosurgical intervention in the caudate nucleus, ansa lenticularis, and basal ganglia for the treatment of movement disorders have evolved in many ways.[2,4-6,10-13,17,22-27,29-35,38,42,43,45,46,51,56,57] Recently, advances in neuroradiological techniques for visualizing the basal ganglia structures and for intraoperative recording of cellular activity in these regions have improved the neurosurgeon's ability to determine lesion
Initially, ventriculography was replaced by computerized tomography (CT) scanning because of the greater reliability and accuracy noted (usually within 1-2 mm) with CT. Although CT produces little inherent distortion, it gives poor anatomical detail compared with magnetic resonance (MR) imaging, which provides much better definition and tissue contrast. The advantages of MR imaging for target identification, however, come with an inherent possibility of increased image distortion and consequent decreased accuracy in target localization. This distortion is induced not only by the magnet itself, but also by the object being imaged. Fortunately, methods now exist and are being refined to minimize this distortion, thereby making possible the successful use of MR imaging in stereotactic functional neurosurgery.

In this report, we present our preliminary work involving 62 stereotactic procedures (55 pallidotomies and seven thalamotomies) performed using MR imaging techniques that minimize distortion, most notably a three-dimensional (3-D) data set and magnetization prepared rapid gradient echo (MPRAGE) imaging, associated computer-based software, and a reformatting process that minimizes distortions created by the magnet and the imaged object. Together these techniques allow for precise placement of lesioning electrodes based on anatomically detailed and accurate imaging studies, without the need for intraoperative microelectrode recording. The efficacy, safety, and utility of this system are explored, and the usefulness of MR-based measurements of the anterior commissure-posterior commissure (AC-PC) line as validation of accuracy is described.

MATERIALS AND METHODS

Image Acquisition

Our imaging protocol for 3-D MR-guided stereotactic pallidotomy is composed of three image data sets: a precontrast 3-D volumetric acquisition 3-D MPRAGE; a postcontrast 3-D MPRAGE; and a proton-density T2-weighted spin-echo image acquisition. We used the precontrast 3-D MPRAGE as the primary localizing modality. It is critical that the parameters of the primary localizing modality be selected to minimize geometric distortions while preserving image contrast. The minimization of distortion was achieved by increasing the bandwidth of the stereotactic MPRAGE imaging relative to the bandwidth of the diagnostic MPRAGE image. The precontrast image volume is subtracted from the postcontrast image volume to yield an MR subtraction venography image volume. The proton-density T2-weighted image acquisition was added to the protocol to aid in distinguishing between the globus pallidus and the internal capsule. The orientation of the proton-density T2-weighted image sequence is in an oblique-axial plane parallel to the AC-PC line. Because of geometric distortions in the resulting T2-weighted imaging, the T2-weighted imaging data set was used only for target verification.
Fig. 1. All images were obtained in a single patient who underwent bilateral pallidotomy, 11 months apart, with excellent outcomes. Left: mid-sagittal; center: axial; and right: coronal T₁-weighted images with the AC and PC indicated.

The MR imagers used in image acquisition and planning for our stereotactic neurosurgical procedures were a 1.5-tesla SP Magnetom and a 1.5-tesla Vision (Siemens Medical Systems, Iselin, NJ). Imaging was performed with the patient secured in a Leksell G-frame (Elekta Instruments, Stockholm, Sweden). The precontrast MPRAGE image was acquired first (Figs. 1-3). Parameters for this sequence were a 280-mm field of view, a matrix size of 192 X 256 pixels, and a slab thickness of 230 mm, resulting in an effective slice thickness of 1.8 mm. As mentioned above, we used a relatively high bandwidth MPRAGE sequence to minimize geometric distortion. This slab thickness results in 128 contiguous 1.8-mm-thick sagittal T₁-weighted images through the entire cranium. Typically, high bandwidth sequences reduce distortion at the cost of decreased image contrast and increased image noise. However, the high bandwidth MPRAGE sequence minimizes distortion while preserving visualization of anatomical details such as the cortical gray-white junction.
Fig. 2. Oblique-axial T₁-weighted image obtained in the same patient, from which measurements were taken, with target overlay and AC-PC line indicated. The initial target calculation was 2 mm anterior to the midcommissural point, 20 mm lateral to the midline, and inferior to the AC-PC line at the level of the floor of the third ventricle.

Imaging parameters for the postcontrast MPRAGE sequence were identical to those for the precontrast MPRAGE sequence. Acquisition of the postcontrast data set was initiated within 3 minutes of injection of a contrast agent containing gadolinium. The timing of image acquisition following contrast effectively limited enhancement to blood vessels in this patient population.
The third sequence of images acquired was a spin-echo proton density T$_2$-weighted image image pair (Fig. 4). These images were acquired in an oblique-axial plane parallel to the AC-PC line as defined in the midsagittal plane. The T$_2$-weighted imaging emphasized the boundary between the globus pallidus and the internal capsule. These images were acquired as 4-mm-thick slices with a 10% gap between slices. Parameters for this sequence were a 280-mm field of view and a matrix size of 256 X 256 pixels, resulting in a resolution of 1.09 mm/pixel. As with the MPRAGE sequences, we increased the bandwidth of the T$_2$-weighted imaging sequence relative to the diagnostic sequence to minimize geometric distortions. Additionally, because geometric distortions are largest at the edges of images, the field of view was chosen such that the stereotactic fiducials did not appear in this region. However, the geometric distortions that remained in the T$_2$-weighted imaging sequence were still too large to be clinically acceptable. Therefore, the T$_2$-weighted imaging sequence was not used as a primary targeting modality, but only for target verification.
Initially, we included stereotactic CT images as a primary targeting modality in our 3-D stereotactic planning protocol because of suspected geometric distortions in MR images. However, acquisition of CT images was abandoned after we ruled out any gross deviations in stereotactic localization between the CT images and the MPRAGE images. In addition, recognition of an appropriate target location in the globus pallidus was difficult in CT images so that there was no apparent benefit from including CT in our imaging protocol. Recently, we performed a phantom study using our 3-D stereotactic MR protocol and compared stereotactic coordinates obtained from this data set with the corresponding positions in CT images. Preliminary results indicate that the coordinates obtained from the MPRAGE sequence compare favorably with those obtained from CT images.

Geometric distortion in the T2-weighted imaging sequence was measured using the method of Maciunas, et al. [40] This method requires acquisition of two T2-weighted imaging sequences, which differ only in that the readout gradient of the second is reversed. Using this method, geometric distortions can be observed by subtracting the first image sequence from the second. The distortion can be measured as one-half the distance between a structure in the first image and the same structure in the second image. In the central slices, the magnitude of the absolute distortion of our T2-weighted imaging was on the order of 1 mm with a maximum of 2.5 mm. The magnitude of this distortion was independently verified. (JM Fitzpatrick, personal communication, 1995). The distortion increased as we moved away from the center of the magnet, as described by Walton, et al. [58] Therefore, we confine our target verification, when using T2-weighted imaging, to slices acquired in the center of the magnet.
Stereotactic Registration

The stereotactic registration techniques applied to the 3-D MPRAGE image volumes used for frame-based, stereotactic surgical planning were similar to techniques used to register images to intraoperative neurosurgical navigational devices. We used a point-to-point least squares match between known stereotactic positions/fiducials and positions extracted from the imaging data.[3] This procedure registered the entire MPRAGE image volume into stereotactic coordinate space. Once the 3-D image volume had been registered, the image volume could be resampled into images of any orientation. These resampled images remained registered to stereotactic coordinate space.

Stereotactic registration of the oblique-axial T2-weighted image sequence was performed using the method described in Lemieux, et al.[39] Registration of both the 3-D MPRAGE and T2-weighted imaging data sets into stereotactic coordinate space allowed direct mapping of stereotactic coordinates of targets chosen in the 3-D MPRAGE image volume into the T2-weighted imaging.

Target Localization and Trajectory Planning

Initial target localization was performed using the method previously published by Laitinen.[33] The 3-D MPRAGE T1-weighted imaging data were resampled into coronal, axial, and sagittal images. Additionally, the 3-D image volume was reformatted into oblique-axial images parallel to the intercommissural line as defined in the midsagittal plane. The resulting oblique-axial data set was coregistered with the 3-D MPRAGE image volume and, by association, with the Leksell coordinate space. Therefore, stereotactic coordinates for a surgical target selected in an oblique-axial image were reported immediately. The oblique-axial image through the intercommissural line allows direct visualization of the structures used to define Laitinen's pallidotomy targeting scheme, as well as the globus pallidus and the internal capsule.

To aid the pallidotomy planning procedure, a graphical representation of Laitinen's target system was overlaid on the set of reconstructed oblique-axial images (Fig. 2). The overlay described a pallidal target 2 mm anterior to the midpoint of the intercommissural line and 20 mm lateral to the midline of the third ventricle.[33,49] The inferior (z) coordinate was placed at the level of the floor of the third ventricle as identified in the sagittal plane. The resulting surgical target was illustrated in a coronal T1-weighted image (Fig. 3).[33,34] The initial pallidal target was mapped onto a stereotactically registered oblique-axial T2-weighted image. The T2-weighted image was used to determine the lateral extent of the posterior ventral pallidum. Distortion in this lateral direction was minimal because it was perpendicular to the readout direction. The modified surgical target was mapped onto the resampled sagittal, axial, and coronal views of the T1-weighted image volume. The three cardinal views of the surgical target illustrated its proximity to the optic tract and other structures (Fig. 1). If necessary, corrections to the computer-planned target could be made by adjusting the target to avoid the optic tract, the internal capsule, or any visible vessels near the target.
The surgical entry point was determined from a 3-D reconstruction of the 3-D MPRAGE image volume. This 3-D reconstruction represented skin surface, cortical surface, and cortical surface vasculature. The cortical surface was extracted from the $T_1$-weighted imaging data using a model-based active surface technique that yielded a 3-D representation of the brain surface (Fig. 5).[50] Magnetic resonance subtraction venography was used to visualize cortical veins larger than 1.5 mm in diameter.[28] The volume of the patient's skin surface was extracted from the 3-D MPRAGE image volume with a simple interactive threshold technique. Entry points were ideally placed anterior to the coronal suture, 0 to 30\° anterior to the vertical plane, and with a right-left angle 0 to 10\° from the vertical plane. Incorporating the MR subtraction venography into the planning procedure allowed the entry point to be planned away from cortical surface veins. The entry point could also be planned on a gyral apex to avoid sulcal veins.

The final stage in the surgical planning process involved reviewing the proposed trajectory interactively in the three cardinal views. This procedure allowed the surgeon to assess the surgical safety of the chosen neuroanatomical path prior to implementation of the surgical plan.

**Surgical Technique**

The surgical technique used in pallidotomy was essentially similar to that previously published by Laitinen and colleagues[32-34] and will not be detailed here. All surgeries were performed by the same surgeon (E.R.L.). The pallidal target and entry site were defined by MR imaging, as above, contralateral to the patient's more symptomatic side.
RESULTS

This methodology has been used in a series of 62 stereotactic procedures performed in 47 adults (55 pallidotomies and seven thalamotomies), and several calculations relating to the AC-PC line have been determined (Fig. 6). As measured from T1-weighted 3-D MR (MPRAGE) data and verified on axial reconstruction through both commissures, the length of the AC-PC line ranged from 22.9 to 32.3 mm, the mean was 27.07 mm, the standard deviation (SD) was 1.84 mm, the median was 26.9 mm, and the mode was 26.9 mm. Determinations obtained using ventriculography include ranges between 21 mm and 28.5 mm, a mean of approximately 25 mm, and a SD of 1.5 mm. As determined by CT scanning, the reported range is 22 to 28 mm, the mean is 25.2 mm, and the SD is 1.67 mm. The measurements of the AC-PC line reported here compare favorably with those previously reported based on ventriculography and CT scanning. Our measurements define anatomical variability and allow the estimation of confidence limits for the MR-based measurements that are essential for accurate targeting of the lesion.

Additionally, a second set of slightly different measurements was made using the same sequences from the same group of patients. These measurements were not used for the actual surgical planning, but rather for comparison. The measurement technique for this second set was altered slightly to match the technique most commonly used in ventriculography and CT scanning. Specifically, in the second set, the AC point was chosen at the posterior margin of the AC because it is the posterior margin that is most easily and clearly defined on ventriculogram and CT. For our surgical planning measurement set, however, the center of the AC was used to place the AC point because MR imaging can clearly and consistently define and visualize the central portion of the AC.

In the second set of AC-PC line measurements, the range was 22 to 28.3 mm, the mean is 25.43 mm, the SD was 1.57, the mode was 25.6 mm, and the median was 25.3 mm. These numbers are extremely close to those previously published for measurements obtained by ventriculography and CT and they are consistently and only very slightly smaller than those obtained in our preoperative planning, which is optimized for MR. The range of the differences in the first measurement set minus the second measurement set was -0.4 to 4.1 mm. The mean was 1.6 mm, the SD was 1.0, the mode was 2 mm, and median was 1.7 mm.

DISCUSSION

In this paper we present data illustrating the use of a 3-D data acquisition set (3-D MR, MPRAGE) that is highly T1-weighted and minimizes distortion. This technique is used for a variety of measurements and target selections. Specifically, it is used to determine the AC-PC line, from which the posterior ventral pallidum is then localized. Our determinations of the AC-PC line compare favorably with previously reported measurements based on cadaver studies, ventriculograms, and CT scans.

Stereotactic functional neurosurgery was greatly aided by the introduction of CT scanning for more accurate targeting. After the development of MR imaging, it was thought by many that the improved tissue contrast and anatomical definition of MR imaging would be useful in stereotactic functional
neurosurgery. However, early work with MR imaging revealed that there was significant magnetic distortion on spin-echo sequences, particularly on $T_2$-weighted images, and determinations of the AC-PC line made from sagittal images were often inaccurate. Initially, this distortion precluded the use of MR imaging for accurate stereotactic targeting. These magnetic distortions are due both to gradient field nonlinearities and to resonance offsets. The resonance offsets can be further subdivided into two categories according to their source: magnetic field inhomogeneities and chemical shift. The magnetic field inhomogeneities may be induced by the magnet itself or by the imaged object. Magnetic field inhomogeneities induced by the imaged object and gradient field nonlinearities are the two most important sources of distortion in MR imaging. The capability to correct chemical shifts has existed for several years, by preprogramming adjustments in the computer software of MR imagers and/or by the use of suppression techniques to remove the effects of these shifts. Techniques are being developed to minimize all of these distortions; however, the neurosurgeon using MR imaging must confirm that distortion corrections have been made on the particular imager being utilized. Provided this is done, MR imaging can produce nearly distortion-free images to guide stereotaxis.

A major cause of distortion in earlier studies using MR imaging for stereotaxis was the two-dimensional imaging technique. In two-dimensional MR imaging, several slices are acquired sequentially to image a 3-D volume. With the 3-D MR imaging technique used in our study, the whole image field of view is excited at once and only weak slice selection is used so that the well-known "potato chip" and "bowtie" effects seen with two-dimensional MR imaging are no longer an issue. In 3-D MR imaging, therefore, the only gradient field distortion that can be produced is barrel aberration. Even this aberration, however, can be corrected by previously measuring the known distortions of the magnetic fields generated by that particular MR imager's gradient coils and adjusting accordingly. Using these methods, the gradient field nonlinearities can be reduced to insignificance.

Resonance effects, another cause of MR imaging distortions, depend on the gyro magnetic ratio of the imaging subject and the magnetic field. As mentioned previously, resonance effects are divided into chemical shifts or magnetic field inhomogeneities and these are produced by changes in the gyro magnetic ratio and by changes in the magnetic field, respectively. It is important to note that magnetic field inhomogeneities are manifest only in the read out or frequency encoding direction of the image. Therefore, we performed our imaging so that the critical parameter, namely measurement of the AC-PC line, was performed in the phase-encoding direction, which does not receive geometric distortion. Fortunately, chemical shifts can be suppressed by selective saturation methods and because the area of greatest clinical interest, the basal ganglia, does not possess an interface of tissues with widely differing MR signals (such as fat adjacent to air or bone); the degree of chemical shift there is both minimal and easily corrected.

Magnetic field inhomogeneities can be produced either by the imager or by an object. Imager-induced distortions, although not expected in an MR imager creating a constant main magnetic field, result from residual imperfections that do exist. With recent improvements in imagers and with the developments of shimming methods, imager inhomogeneity can be corrected so that it no longer creates significant inaccuracy. Using the imaging techniques we describe, the only remaining significant distortion is object-induced distortion. This distortion depends on both the shape and composition of the imaged structure. Fortunately, this distortion only occurs in the gradient read out (frequency encoding) direction and so it can be avoided to a significant degree by the choice of imaging systems. Recently it has been noted that by correcting gradient field and magnetic field inhomogeneities, 1 mm geometric accuracy
may be achieved using MR imaging, making it comparable to CT.[7,8,53,54]

Other sources of inaccuracy that can influence MR-guided stereotactic procedures are also addressed in our current study. We used the Leksell G-frame system, which has been found in numerous studies to be comparable or superior in accuracy to other major frame-based systems.[14] In addition, a major source of error, not often appreciated, is placement of the field of view and the location of the fiducials used for transformation from image space to stereotactic space. A recent study that reported increased distortion using MR imaging for stereotaxis used a system in which the fiducials were placed near the limits of the coils, with a field of view that was too small.[58] This placement-created distortion is not seen in our study because of our less peripheral placement of the fiducials. In the same study, additional distortion was created because phantom placement prevented the G-frame post from being placed in its usual location, thereby creating an overestimation of the error associated with the Leksell G-frame system. Additional error was introduced by using a system that lacked a direct mechanism for the transfer of the MR imaging data to the radiosurgery planning work station. In the published study, all images were transferred to film and then imported to their planning system using a flat bed image scanner, creating further distortion in the images. Furthermore, there is no mention in their study of the use of any of the MR distortion corrections outlined above. Finally, it should be noted that the degree of distortion would be expected to be greatest where there is a tissue interface in a more peripheral location of the imaging system. This raises the absolute maximum measured error in their study, thereby disproportionately elevating the mean error. For the majority of stereotactic procedures, namely brain tumor biopsy and functional stereotaxy, which take place within the centrally imaged substance of the brain, the magnetic susceptibility is relatively homogeneous and magnetic field inhomogeneity is much less problematic.

We note in our study an accuracy within 1 to 2 mm, supported by a phantom quality control analysis. Intraoperatively, however, we use several techniques to assure further the safety and accuracy of our planned target. Impedance measurements can be used to aid in the placement of the electrodes and occasionally can provide confirmatory information. Also, by closely examining the patient's vision, sensorimotor function, and speech during prelesion stimulation at low and then high frequencies, the neurosurgeon can predict and prevent potential injury to the optic tract, the internal capsule, and adjacent structures that could be produced by the therapeutic radiofrequency lesion. Additionally, we routinely evaluate the target site with a microvascular Doppler device placed down the stereotactic cannula.

In anticipation of the occasion when the pallidotomy target might yield unacceptable neurological symptoms on prelesion stimulation and assessment, an alternate target in the ipsilateral ventrolateral thalamus is chosen during stereotactic planning. On one occasion we effectively switched to the alternate target to perform a thalamotomy.

Brain segmentation techniques used in our patients provided a combination of safety and accuracy not previously available in stereotactic ablative procedures. More specifically, by performing brain segmentation, a 3-D surface reformation of cortical gyri and sulci is provided enabling the surgeon to pick an entry point passing through the middle of a gyrus, avoiding sulci and potential vascular injury. The MR subtraction venography also enables the surgeon to place the burr hole optimally away from major cortical veins. Trajectory slices enable the surgeon to determine which nuclei or tracts the electrode will pass through on its way to the target. These segmentation procedures are simple and easy to perform and can be included in the software programming of the data acquired from the MR image.

We experienced only one major complication in our series. This involved a large intracerebral
hemorrhage that occurred approximately 5 hours after the procedure was performed, eventually culminating in death. We have otherwise observed no infections, permanent visual field deficits, or motor deficits. Transient worsening in speech and swallowing function occurred in two patients early in the series.

Although long-term outcome studies have not yet been performed, we have noted preliminarily in our parkinsonian patients that many have experienced marked improvement in all of the major symptoms of Parkinson's disease including bradykinesia, hyperkinesia, "freezing," tremor, and rigidity. Objective analysis of this data and of outcome is currently in progress.

In summary, we describe a stereotactic planning technique for the treatment of Parkinson's disease, but have also used the same technique to perform successful thalamotomies for severe tremor and centromedian thalamotomies for pain. The techniques could likely be generalized for use in other stereotactic funcrional procedures. We have found that by using the 3-D MPRAGE sequence described, specific software adaptations, and an appropriate imaging protocol, it is possible to minimize distortion and improve accuracy. This allows for precise determination of the AC-PC line and, thus, the stereotactic target, using MR imaging alone. This translates into accurate and efficient targeting of the posterior ventral lateral pallidum, through a safe trajectory, to maximize good outcome in the treatment of patients with Parkinson's disease.

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