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 and changed in many ways. Recently, advances in neuroradiological procedures for visualizing the basal ganglia structures and in intraoperative recording of cellular activity in these regions have improved the neurosurgeon’s ability to determine lesion location. Ventriculography was replaced by computerized tomography (CT) scanning because of the greater reliability and accuracy noted (usually within 1–2 mm) with CT scanning. Although CT scanning 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, consequently, 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 magnetic distortions created by the magnet and the object being imaged. Together these techniques allow for...
precise placement of lesioning electrodes based on anatomically detailed and accurate imaging studies, without the absolute requirement for intraoperative microelectrode recording. The efficacy and safety 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.

Clinical Material and Methods

Image Acquisition

Our imaging protocol for 3-D MR-guided stereotactic pallidotomy is comprised 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 relative to the bandwidth of the diagnostic MPRAGE. 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 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.

The MR imaging systems 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 and 2). Parameters for this sequence included a 280-mm field of view, a matrix size of 192 × 256 pixels, and a slab thickness of 230 mm, resulting in an effective slice thickness of 1.8 mm. As mentioned earlier, we used a relatively high-bandwidth MPRAGE sequence to minimize geometric distortion. This slab thickness results in 128 contiguous 1.8-mm-thick sagittal T2-weighted images through the entire cranium.

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 after injection of a contrast agent containing gadolinium. The timing of image acquisition following administration of the contrast agent effectively limited enhancement to blood vessels in our patient population.

The third sequence of images acquired was a spin-echo proton-density T2-weighted image (Fig. 3). These images are acquired in an oblique–axial plane parallel to the AC–PC line as defined in the midsagittal plane. The T2-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 the

Fig. 1. Midsagittal (left), axial (center), and coronal (right) T1-weighted images with the AC and PC indicated. All images shown in Figs. 1 through 5 were obtained in a single patient who underwent pallidotomy bilaterally, 11 months apart, with excellent outcomes.

Fig. 2. Left: Oblique–axial T1-weighted image, from which measurements were taken, with target overlay and AC–PC line indicated. The initial target calculation is 2 mm anterior to the midsagittal point, 20 mm lateral to the midline, and inferior to the AC–PC line at the level of the floor of the third ventricle. Right: Coronal T1-weighted image obtained 2 mm anterior to the midsagittal point. The target point, floor of the third ventricle, and optic tract are indicated.
slices. Parameters for this sequence were a 280-mm field of view and a matrix size of $256 \times 256$ pixels, resulting in a resolution of $1.09 \text{ mm/pixel}$. 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 scans as a primary targeting modality in our 3-D stereotactic planning protocol because of suspected geometric distortions in the MR images. However, acquisition of CT scans was abandoned after we ruled out any gross deviations in stereotactic localization between the CT and MPRAGE images. In addition, stereotactic coordinates obtained with the 3-D stereotactic MR protocol were compared with the corresponding positions on CT scans. A total of 29 comparisons between MR and CT images were made, similar to phantom studies previously published. The results indicate that the coordinates obtained from the MPRAGE sequence compare favorably with those obtained from CT scans.

Geometric distortion in the $T_2$-weighted imaging sequence was measured using the method of Maciunas, et al. This method requires acquisition of two $T_2$-weighted imaging sequences, which differ only in that the readout gradient of the second is reversed. In the central slices, the magnitude of the absolute distortion of our $T_2$-weighted imaging was on the order of 1 mm with a maximum of 2.5 mm. The magnitude of this distortion was independently verified by a medical imaging expert at another institution (JM Fitzpatrick, personal communication, 1995). The distortion increased as we moved away from the center of the magnet, as described by Walton, et al. Therefore, we confined our target verification, when using $T_2$-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. 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 remain registered to stereotactic coordinate space.

Stereotactic registration of the oblique–axial $T_2$-weighted imaging sequence was performed using the method described in Lemieux, et al. Registration of both the 3-D MPRAGE and $T_2$-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 $T_2$-weighted imaging.

**Target Localization and Trajectory Planning**

Initial target localization was performed using the method previously published by Laitinen, et al. The 3-D MPRAGE $T_2$-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 mid-sagittal plane. The resulting oblique–axial image 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 graphic representation of Laitinen’s target system was overlaid on the set of reconstructed oblique–axial images (Fig. 2 left). 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. 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 $T_2$-weighted image (Fig. 2 right). The initial pallidal target was mapped onto a stereotactically registered oblique–axial $T_2$-weighted image. The $T_2$-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 $T_2$-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 T₁-weighted imaging data using a model-based active surface technique that yielded a 3-D representation of the brain surface (Fig. 4). Magnetic resonance subtraction venography was used to visualize cortical veins larger than 1.5 mm in diameter. The volume of the patient’s skin surface was extracted from the 3-D MPRAGE image volume with a simple interactive threshold technique. Incorporating MR subtraction venography into the planning procedure allowed the surgeon to plan the entry point 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 and the perioperative care used in our patients were essentially similar to those previously published by Laitinen and colleagues and will not be detailed here. All surgeries were performed by the same surgeon (E.R.L.). Follow-up CT scans were routinely obtained on postoperative Day 1 (Fig. 5).

Results

This methodology has been used in a series of 62 stereotactic procedures in 47 adults (55 pallidotomies and seven thalamotomies) and several calculations relating to the AC–PC line have been determined. 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 reported ranges of between 21 mm and 28.5 mm, a mean of approximately 25 mm, and an SD of 1.5 mm. As determined by CT scanning, the reported range is 22 to 28 mm, the mean 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. For our surgical planning measurement set, however, the center of the AC was used to place the AC point.

In the second set of AC–PC line measurements, the range was 22 to 28.3 mm, the mean was 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 scanning, and they are consistently and only very slightly smaller than those obtained in our preoperative planning, which is optimized for MR imaging. The range of the differences of 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 mm, the mode was 2 mm, and median was 1.7 mm.

In the phantom study we performed using our 3-D stereotactic MR protocol, we compared stereotactic coor-
ordinates obtained from this data set with the corresponding positions on CT scans. The mean total difference between points on MR and CT images was 2.2 mm with an SD of 1.1 mm. The mean differences between individual x, y, and z coordinates were 0.6, 0.6, and 1.9 mm, respectively, with corresponding SDs of 1, 0.5, and 1.1 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 T$_1$-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.

Initially, magnetic distortions precluded the use of MR imaging for accurate stereotactic targeting. These magnetic distortions are due both to gradient field nonlinearity and to resonance effects. We used techniques to minimize these distortions; however, the neurosurgeon using MR imaging must confirm that these distortion corrections have been made on the particular imaging system that is being utilized. Provided this is done, MR imaging can create nearly distortion-free images and accurate coordinates.

A major cause of magnetic distortion in earlier studies using MR imaging for stereotaxis was the two-dimensional (2-D) imaging technique. In 2-D 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 2-D 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 nonlinearity can be reduced to insignificance.

Resonance effects, the other subgroup of magnetic distortions, are subdivided into chemical shifts or magnetic field inhomogeneities, which are produced by changes in the gyro-magnetic ratio or by changes in the magnetic field, respectively. Because the basal ganglia does not possess an interface of tissues with widely differing MR signals, the degree of chemical shift in that location is both minimal and easily suppressed by selective saturation methods and/or software modifications. The magnetic field inhomogeneities are manifest only in the readout 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.

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 development 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 magnetic field inhomogeneity distortion. This distortion depends on both the shape and composition of the imaged structure. Fortunately, this distortion only occurs in the gradient readout (frequency-encoding) direction, and therefore it can be avoided to a significant degree by choice of imaging systems. Recently it has been noted that by correcting gradient field and magnetic field inhomogeneities, geometric accuracy within 1 mm may be achieved with MR, making it comparable to CT scanning.

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. 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. In a recent study, Walton, et al., reported increased distortion using MR imaging for stereotaxis. These researchers used a system in which the fiducials were placed near the limits of the coils, with a field of view that was too small. This placement-created distortion was not seen in our study because of our less peripheral placement of the fiducials used for stereotactic planning. 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. This was avoided in our study. We avoided additional error by using a system that allows for direct transfer of MR imaging data to the radiosurgery planning workstation. In the study published by Walton, et al., all images were transferred to film and then imported into their planning system using a flat-bed image scanner, creating further distortion in the images.

We note in our study an accuracy within the 0.5- to 2-mm range, supported by a phantom quality control analysis. Given the fact that the minimum effective slice thickness is 1.8 mm with our MR protocol and 3 mm for CT scanning, the phantom study results seem quite encouraging. 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 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 lesioning. Additionally, we routinely evaluate the target site with a microvascular Doppler device placed down the stereotactic cannula.

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 sur-
face reformation of cortical gyri and sulci is provided, enabling the surgeon to pick an entry point passing through the middle of a gyrus, avoiding the 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.

Although long-term outcome studies have not yet been performed, we have noted preliminarily in our parkinsonian patients that many patients have experienced marked improvements in all of the major symptoms of Parkinson’s disease including bradykinesia, hypokinesia, “freezing,” tremor, and rigidity. Objective analysis of these 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 technique could likely be generalized for use in other stereotactic functional procedures as well. 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 outcomes in the treatment of patients with Parkinson’s disease.

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