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A Comparison between Magnetic Resonance Imaging and Computed Tomography for Stereotactic Coordinate Determination

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The spatial accuracy of magnetic resonance imaging (MRI) has not been established for stereotactic surgery. Magnetic susceptibility artifacts may lead to anatomical distortion and inaccurate stereotactic MRI coordinates, especially when targets are in regions of the brain out of the center of the magnetic field. MRI-guided stereotactic localization, however, provides better multiplanar target resolution than is available with computed tomographic (CT) scanning. Therefore, we compared the accuracy of stereotactic coordinates determined by MRI and CT studies in 41 patients (53 targets). Coordinates were measured in each plane and as vector distances between the target and the center of the stereotactic frame on axial or coronal MRI studies. Absolute axial plane MRI and CT distances varied an average of 2.13 ± 1.59 mm. The mean difference in measurements in the *X* (left-right) dimension was 1.19 mm and 1.55 mm in the *Y* (anterior-posterior) dimension. Central targets (located less than 2 cm from the frame center) had a mean MRI-CT difference of 2.09 ± 1.79 mm; peripheral targets (greater than 2 cm from the frame center) differed by 2.17 ± 1.3 mm. The voxel volumes were calculated for all compared images. Although differences between the physical properties of data acquisition with each imaging modality could explain the observed CT-MRI discrepancies, a 1-pixel difference in target selection could account totally for all the variance observed. MRI field strength (0.5 vs. 1.5 T) did not correlate with coordinate determination accuracy. We conclude that MRI-guided stereotactic localization can be used with confidence for most diagnostic, functional, and therapeutic stereotactic procedures. (*Neurosurgery* 30:402–407, 1992)

Key words: Computed tomography, Magnetic field, Magnetic resonance imaging, Stereotactic surgery

INTRODUCTION

The advantages of using magnetic resonance imaging (MRI) rather than computed tomographic (CT) scans in stereotactic surgery include the increased imaging resolution of the lesion or target (using contrast enhancement or different pulse sequences), direct nonreformatted multiplanar imaging and target coordinate determination, and reduced imaging artifacts produced by the stereotactic frame (13). The use of MRI is especially beneficial when performing stereotactic surgery in patients with brain lesions or normal anatomical targets that are poorly demonstrated by CT scanning or contrast ventriculography. Stereotactic biopsies (13, 17), functional stereotactic surgery (6, 7, 9, 11, 18), and stereotactic radiosurgery (10) all require MRI guidance for such patients.

Because inhomogeneities in magnetic field gradients can lead to geometric image distortion (magnetic susceptibility artifacts), the accuracy of stereotactic MRI guidance has been questioned (6, 14). Optimal imaging can be attained by the frequent calibration of the MRI unit to standard test phantoms, the use of nonferromagnetic frames and fiducial systems, and the immobilization of the patient (5, 7, 8). To determine whether MRI provides a consistent and accurate method to obtain stereotactic coordinates, we compared MRI-determined stereotactic target measurements with those obtained with CT scans. We also examined the possible effects of different magnetic field strengths during stereotactic imaging and whether MRI was more reliable in imaging central brain targets than in imaging peripheral targets.

PATIENTS AND METHODS

Stereotactic target coordinates were obtained in 41 patients using both MRI studies and CT scans. We compared these images for a total of 53 targets: 16 in patients undergoing stereotactic biopsies, 27 in patients undergoing stereotactic radiosurgery, and 10 in patients undergoing functional stereotactic surgery (thalamotomies or capsulotomies). This study was completed in 2 years using resources at two university teaching hospitals (University of Pittsburgh and University of Toronto). The Leksell stereotactic frame (Elekta Instruments, Tucker, GA) and coordinate-determination system were used in all patients to obtain both the CT and MRI data. The MRI-compatible stereotactic frame was constructed from a nonferromagnetic aluminum alloy. All imaging was performed using General Electric scanners (General Electric Medical Systems, Milwaukee, WI): for nonionic contrast-enhanced CT imaging, a 9800 CT scanner (field of view, 250 mm) was used; for MRI studies, either a 1.5-T Signa MRI scanner (26 patients, 30 targets) or a 0.5-T MAX scanner (15 patients, 23 targets). MR images were obtained at 3- or 4-mm slice intervals with no interval between slices. All CT slices were 5-mm thick and were scanned at 3-mm intervals. Using similar slice intervals, we attempted to obtain comparison images with a similar *Z* (superior-inferior) coordinate. With MRI studies, coordinates were obtained from short TR sequences, with or without paramagnetic contrast material, unless the lesion could be demonstrated better using long TR sequences. Only spin-echo sequences and no gradient-echo sequences were used. The application of the stereotactic frame, imaging, and surgery were performed with patients under local anesthesia.

After the first images were performed, we obtained reference mea-

surements of the distance between the fiducial markers on the stereotactic frame. These markers represented fixed points in space with a separation of 120 mm in the sagittal dimension and 190 mm in the transverse dimension. In all patients, these measured distances fell within ±0.7 mm of the actual distances (Fig. 1).

All stereotactic coordinates first were obtained using the standard distance-measurement software of the CT or MRI scanner. We then confirmed these coordinates using a manual localization technique that allows a scaled grid to be superimposed on the hard copy images. Stereotactic targets included the site for the tumor biopsy; the site for the irradiation isocenter placement during radiosurgery; or, for functional procedures, the anterior or posterior commissures or the internal capsule. Although we attempted to study the same pixel on both CT and MRI studies of a given target, we accepted the possibility that we could have selected different targets by a factor of 1 pixel. We therefore calculated the pixel sizes for each patient to determine the degree of error potentially caused by intraobserver variation, independent of the physical characteristics of the imaging technique. To calculate the pixel size, we recorded the field of view (FOV) and the matrix size ($n_{phase} \times n_{frequency}$) used, then applied the formula $FOV^2/n_{phase} \times n_{frequency}$. For example, a pixel in an image obtained using a 250-mm FOV and a 256 × 192 matrix would measure 1.28 mm^2 (*X* dimension, 0.98 mm; *Y* dimension, 1.3 mm). Thus, a 1-pixel difference in CT and MRI target selection would lead to a maximum discrepancy of 1.95 mm in the *X* dimension, 2.6 mm in the *Y* dimension, and 3.25 mm in the hypotenuse. MRI scans were obtained using the following techniques: a) FOV = 260 mm, matrix = 256 × 256 (n = 27); b) FOV = 240 mm, matrix = 256 × 192 (n = 3); c) FOV = 240 mm, matrix = 256 × 256 (n = 1); and d) FOV = 250 mm, matrix = 256 × 192 (n = 22).

To compare target-coordinate determinations, we selected the axial plane CT and MRI studies of the same lesion and brain anatomy (Fig. 2). Because the superior–inferior stereotactic coordinate was dependent on the axial image chosen, we did not use this coordinate to compare CT and MRI measurements. Each comparison measurement was determined by the absolute distance between the stereotactic frame center and the target in two planes (Fig. 3).For statistical analysis, the mean differences were compared with a *t* test for paired or independent samples as appropriate. The null hypothesis was that no difference should exist except that caused by an intraobserver selection variance of 1 pixel (an acceptable intraobserver variation; allowed

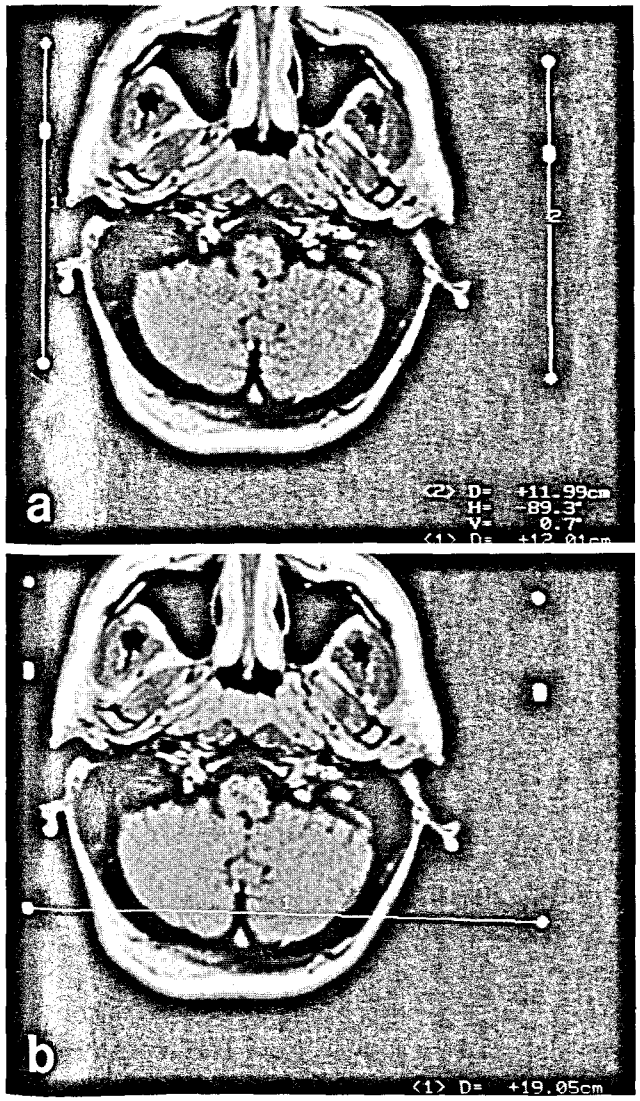

FIG. 1. Axial T1 spin-echo MRI studies show distance measurements between the side-plate fiducial markers (*A*) (actual distance, 120 mm; measured distances, 119.9 and 120.1 mm) and the transverse fiducial markers (*B*) (actual distance, 190 mm; measured distance, 190.5 mm).

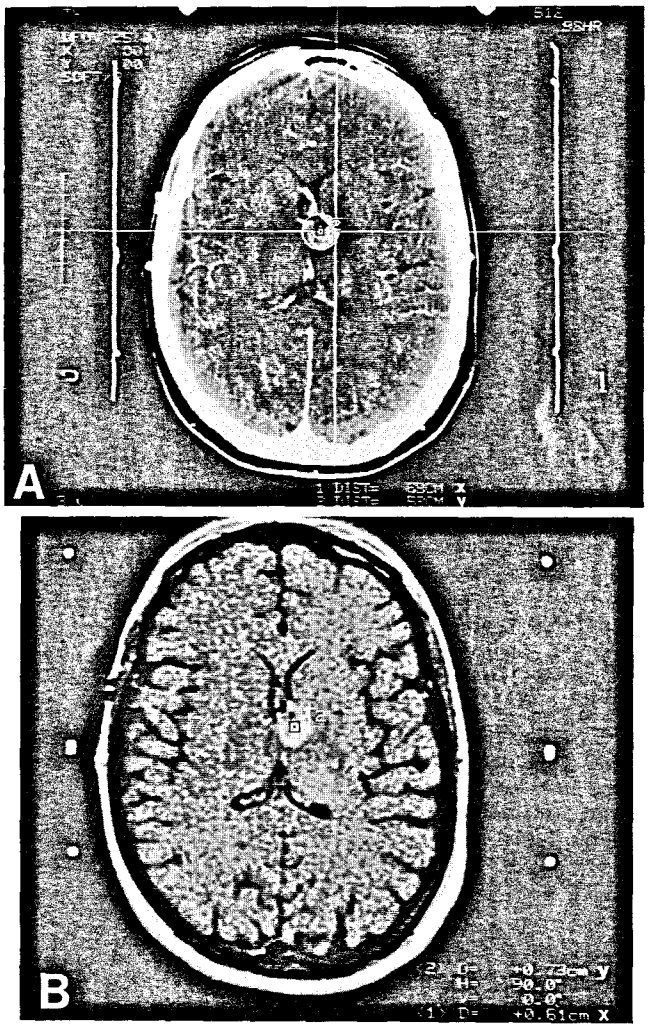

FIG. 2. Axial CT image with contrast enhancement (*A*) and T1-weighted spin-echo MRI study with gadolinium diethylene-triaminepentaacetic acid (*B*) show an anaplastic astrocytoma of the internal capsule and thalamus at the time of the stereotactic biopsy. The *center of the square* marks the target coordinate used in comparing the two techniques.

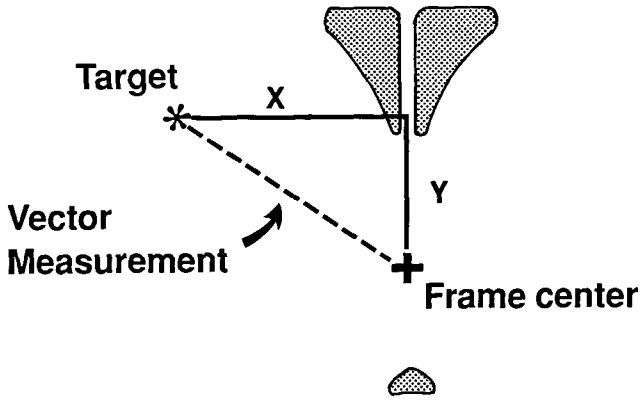

FIG. 3. The two-dimensional vector distance from the target to the center of the stereotactic frame was calculated as the hypotenuse of the triangle formed by lines X and Y in the axial plane. X represents the left-right measurement (X coordinate), Y represents the anterior-posterior measurement (Y coordinate), and the *cross* refers to the center of the stereotactic frame.

error of 2.82, 3.13, 2.66, or 3.25 mm, depending on the MRI technique used). Statistical significance was determined at the $P < 0.05$ level.

Because reformatted coronal CT images are not feasible for stereotactic target determination, we could not compare coronal plane MRI- and CT-derived coordinates. Instead, we examined the consistency between axial CT and coronal MRI coordinates in the same patient for 28 of the 53 targets. Such a comparison of measurements in different planes is totally dependent on image slice selection by the surgeon. To obtain appropriate images for comparison, we chose those that included a similar definition of the target. This comparison is important, because potential inhomogeneities in the magnetic field gradient can differ in different planes. The accuracy of coronal target determination is most important for lesions shown best in the coronal plane, such as pituitary tumors to be treated by radiosurgery. We did not use MRI studies in the sagittal plane.

Our study examined two other factors. To determine whether MRI was more accurate for central (defined herein as targets within 2 cm of the frame center) than for peripheral targets, we compared the measurement discrepancies between MRI and CT studies for central targets and peripheral targets (>2 cm from the frame center). We also stratified vector measurements according to the magnetic field strength of the MRI unit used.

RESULTS

Axial plane comparison

For each of the 53 targets, we compared the vector distances from stereotactic targets to the frame center on both MRI and CT studies. The mean difference obtained by subtracting the MRI distance from the CT distance was 2.13 mm [standard deviation (SD), 1.59]; this difference is not significant if we accept the null hypothesis allowing a 1-pixel error. We also studied this distance difference in the X and Y dimensions individually. In the X dimension, the mean difference between MRI and CT measurements was 1.19 mm (SD, 1.00); in the Y dimension, it was 1.55 mm (SD, 1.5). These dimensional differences were not significant when a 1-pixel dimension error was allowed (Fig. 4).

Coronal plane comparison

For all 28 targets studied, we found a mean vector difference of 2.04 mm (SD, 1.14) between coronal plane MRI measurements and axial CT measurements. This difference was not significant using the null hypothesis of 1-pixel error. When we compared the coronal MRI–axial CT plane difference with the axial MRI–axial CT plane difference, for 28 targets that had both of these studies, the mean axial–coronal difference was 0.13 mm. These differences between planes were not significant.

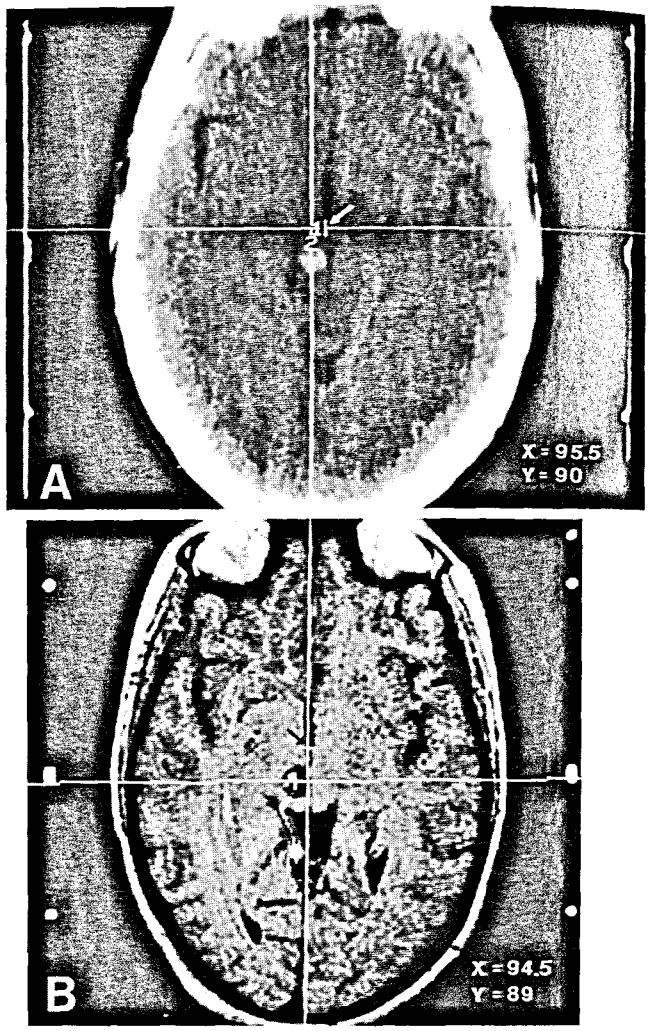

FIG. 4. Axial CT image with contrast enhancement (A) and T1-weighted spin-echo MRI study (B) show an angiographically occult vascular malformation within the right midbrain. The center of the stereotactic frame ($X = 100$, $Y = 100$) is identified by an *arrow*. The individual X and Y coordinates for the compared target obtained with each imaging method (*lower right corner*) differed by 1 mm in each dimension.

Central versus noncentral targets

For centrally located targets in the axial plane (n = 31), the mean difference between MRI- and CT-derived measurements was 2.09 mm (SD, 1.79). The X measurement mean difference was 1.24 mm (SD, 1.47), and the Y measurement mean difference was 1.09 mm (SD, 1.65). In the 22 peripheral targets, the mean vector difference was 2.17 mm (SD, 1.3). When compared with central targets, this difference was not significant. The mean X measurement difference for peripheral targets was 1.13 mm (SD, 1.66); for the Y measurement, it was 0.87 mm (SD, 1.28). Independently, the X and Y measurement differences between central and peripheral targets were not significant. Similarly, in the coronal MRI plane, a comparison between 19 central-target vector measurements (mean MRI–CT difference, 2.1 mm; SD, 1.04) and 9 peripheral measurements (mean difference, 1.93 mm; SD, 1.39) revealed no difference.

Magnetic field strength differences

In the 30 targets studied using a magnetic field strength of 1.5 T, the mean difference in MRI and CT axial plane vector measurements was 2.04 mm (SD, 1.77). In the 23 target measurements obtained with the 0.5-T unit, the mean difference was 2.24 mm (SD, 1.67). The variation in measurements obtained at these different magnetic field strengths was not significant ($P > 0.1$).

DISCUSSION

Accuracy of MRI

Stereotactic localization depends on the spatial accuracy of the images used. CT scans or plain roentgenograms maintain linear accuracy by using x-ray photons for data acquisition (7, 14). The possibility of magnetic susceptibility artifacts and image distortion has caused some to question the accuracy of MRI for stereotactic coordinate determination (6, 8, 14, 16, 18). MRI stereotaxis is potentially superior to other imaging techniques, because MRI enables nonreformatted imaging, provides better anatomical resolution, can define the target using different pulse sequences or contrast enhancement, produces no ionizing radiation, and minimizes imaging artifacts caused by the frame itself. Despite these advantages, current limitations in the use of MRI include restricted availability, limited use of general anesthesia, longer image-acquisition time compared with CT scanning, and higher cost.

A fundamental prerequisite for high-quality MRI studies is a stable magnetic field. The primary factors that introduce geometrical distortion are inhomogeneity in the magnetic field and nonlinear magnetic field gradients (15). Field homogeneity may be disrupted by imperfections in the manufacturer's magnet construction, temporal fluctuations in the power supply, thermal instability, internal or external ferromagnetic objects, or susceptibility artifacts at air–fat or air–water interfaces (e.g., at the nasal cavities or the thoracoabdominal junction) (5). The most common artifact is caused by patient movement. In stereotactic MRI, the frame must be held rigidly within the coil, either by using a frame holder that is shaped to fit the curve of the coil or by taping the frame directly to the coil. Although not so important in intracranial imaging, chemical shift artifacts are important at fat–water interfaces. The fat protons precess more slowly than the water protons in the same slice; the signal for the fat protons then may be misregistered to an incorrect location (5). The lack of air–water or air-tissue interfaces in the brain should limit the occurrence of susceptibility artifacts that depend on the sequence method.

The interaction between the magnetic resonance coil and the imaged subject produces a charge density on the subject owing to induction (4), although the magnitude of any created distortion is unknown. Eddy currents are residual magnetic gradients that can result in more rapid dephasing of magnetization when noncylindrical or nonspherical shapes are imaged (3, 4). Such distortion is more noticeable in images obtained away from the magnet isocenter (5). We could not identify distances away from the magnet isocenter as a specific reason for error in this study, because stratification of the data according to central and peripheral targets revealed no differences.

Various authors have compared stereotactic data obtained with MRI and CT studies in small numbers of patients. Lunsford et al. (13) reported a difference of 0 to 2 mm in individual X or Y measurements in 3 patients. In 6 patients, Andoh et al. (1) found that X dimension measurements differed by a mean of only 0.03 mm and that Y dimension measurements differed by a mean of 1.7 mm (1). Mean differences of 1.0 and 3.75 mm in the X and Y dimensions, respectively, were reported by Bradford et al. (2). Heilbrun et al. (8) studied 9 targets with MRI and CT scanning using the Brown-Roberts-Wells stereotactic system; they found the average error for the coordinates to be as great as 5 mm. Finally, in a comparison between MRI and ventriculographic information in the X and Y dimensions in 6 patients undergoing functional surgery, Villemure et al. (18) found differences of 0 to 3 mm for the two imaging methods.

In the present series of 53 targets, the mean two-dimensional vector difference was 2.13 mm in the axial plane. Differences of approximately 2 mm also were found in the coronal plane and in the central-peripheral target study. We identified no specific inaccuracy in a single direction, that is, the X measurement was no more or less accurate than the Y value. The magnetic field strength (either 1.5 T vs. 0.5 T) of the MRI unit had no effect on coordinate error.

Other possible sources for observed error in this study included inhomogeneous shimming of the magnet (suboptimal tuning of individual shim coils to achieve magnetic field homogeneity) and selection by the surgeon of close but not identical pixels for comparison of CT and MRI studies (an error dependent on voxel volume). Quality-assurance measures designed to minimize magnet inhomogeneity and servicing of the magnet (15) were performed at 2-week intervals, according to the manufacturer's specifications. Regarding the second potential source of error, all observers were experienced in stereotactic coordinate determination and made a diligent effort to select the same anatomical target on both CT and MRI studies. As noted earlier, however, a 1-pixel error in target matching could cause a potential discrepancy of 1.95 (X) or 2.6 (Y) mm in each dimension, resulting in a 3.25-mm discrepancy in the hypotenuse vector measurement. A variance of this magnitude could occur not only in a comparison of MRI with CT scanning, but also in a comparison of sequential CT scans, depending on CT-pixel size. The potential inconsistency between CT and MRI coordinates caused by a 1-pixel selection difference alone can explain all of the CT-MRI differences found in this study. Although susceptibility artifacts may exist, they were not larger than the allowed surgeon's error in this statistical evaluation. To minimize this error, pixel or voxel size can be reduced by using a small FOV and a large matrix MRI technique. Another source of error could have been the failure to compare the exact same image slices on MRI or CT [in the superior-inferior dimension (for axial images) or in the anterior-posterior dimension (for coronal images)]. Although we could not eliminate this error, we tried to reduce it by using 3-mm slice intervals for all scans, to provide more images so that identical views of the target were available for comparison. Using more images and selecting individual slices for comparison, we thought that any error in the Z coordinate was reduced as much as possible, but not eliminated.

To diminish any potential distortion caused by the stereotactic frame itself and to reduce the voxel volume by decreasing the FOV, Rousseau et al. (16) developed a fiducial system of four copper sulfate-filled boxes, separately attached to the outer table of the skull. We think these modifications unnecessary, because the greater error of variation in surgical target selection remains unchanged.

Choice of stereotactic imaging technique

For most patients, CT scanning remains the imaging technique of choice because the data acquisition time is shorter than with MRI studies, imaging resolution is adequate, it is more accessible, and the cost is lower. For functional surgery, the biopsy of a small lesion in a vital location, or the placement

of a radiation isocenter during radiosurgery, accuracy within 1 mm is mandatory. For these indications, the potential error associated with stereotactic MRI technique must be weighed against the deficiencies associated with stereotactic CT scanning. Although CT scanning-derived information is anatomically accurate, MRI studies provide superior imaging resolution for most clinical problems. Because it is noninvasive, CT scanning has supplanted ventriculography at many centers as a method to identify the anterior and posterior commissures in functional neurosurgery (12). MRI studies provide much better visualization of the commissures than does CT scanning, but they are insufficient to act as the sole guidance for a thalamotomy (7–9); for this purpose, electrophysiological recording or stimulation remains essential. During a thalamotomy or a deep-brain electrode placement, stereotactic MRI techniques offer further benefits over stereotactic CT techniques by providing improved commissural imaging, thalamic nuclei recognition, and graphic depiction of the internal capsule.

For the biopsy of lesions that are poorly seen on CT scans but well visualized on MRI studies, the value of MRI guidance outweighs possible spatial variation. During the biopsy of a small lesion in a critical location, potential MRI targeting errors are counterbalanced by the additional information that MRI studies often provide (e.g., the vascular anatomy around a pineal or brain stem lesion). We prefer stereotactic MRI guidance during radiosurgery if resolution is poor on preoperative axial CT images. Axial CT scanning inadequately depicts pituitary microadenomas and small macroadenomas, especially with regard to their proximity to the optic chiasm, because of regional bony artifacts; we use coronal MRI studies routinely for this purpose. Additionally, because MRI far exceeds CT studies in the ability to demonstrate angiographically occult vascular malformations, we now use MRI studies alone to guide the treatment of these lesions (10).

SUMMARY

A difference of approximately 2 mm was identified between two-dimensional comparison measurements using stereotactic MRI studies and CT scans. This difference was identified on well-calibrated MRI units, did not increase with distance from the center of the coil, and was similar on both 1.5- and 0.5-T MRI units. Although magnetic susceptibility artifacts can occur, this difference can be explained by a 1-pixel difference in target selection between the two methods, which is an acceptable intraobserver variation. The importance of any potential inaccuracy associated with MRI stereotaxis usually is surpassed by the greatly increased target resolution and brain visualization associated with MRI techniques.

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COMMENT

Because of the potential image distortion from magnetic susceptibility artifacts, the accuracy of magnetic resonance imaging (MRI) studies for stereotactic localization has been questioned. In this study, the authors compare the position of 53 targets in 41 patients as shown on both MRI and computed tomographic (CT) localization scans. Relating the position of the selected target and the distance of the selected target from the center of the stereotactic frame as seen on both images, they concluded that, in a properly calibrated MRI machine, a difference of 1 pixel in intraobserver target choice could account for distance variability. The authors conclude that this difference is not significant.

This study also addresses the various other factors that potentially affect MRI image distortion in stereotactic localization. These include chemical considerations, such as magnetic field inhomogeneities resulting from air, fat, and water inter-

faces, as well as the physical fixation and positioning of the head frame relatively parallel with the gantry.

Although the authors note that MRI images are usually acquired with a 256 × 192 or 256 × 256 matrix, they do not address directly the fact that routine CT images use a 512 × 512 matrix, a format factor that also might influence the 1-pixel difference. As future MRI advances allow 512 × 512 matrix image acquisition, this variable should be reviewed, since the higher-resolution 512 × 512 matrix could further reduce the 1-pixel differences. Additionally, the fixation that the authors describe allows a direct manual measurement of the distance of the target to the frame center and may have an advantage over indirect diagonal rod computations, which can be affected by inhomogeneities from the air–rod interface. A manual method has potential error if the head frame is not parallel with the image acquisition plane. For example, the tilt of the head frame by 2° from the gantry could account for a 1-pixel difference in distance measurement.

In summary, this important paper needs to be read by all neurosurgeons using MRI techniques for stereotactic localization. MRI stereotactic localization can be accurate only if the neurosurgeon understands the potential distortion effects of magnetic field inhomogeneity and how to reduce such effects by proper machine calibration, proper frame alignment, and redundant systems for target selection and coordinate determination, including, if possible, concurrent CT scanning.

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