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AJNR Am J Neuroradiol 1994, 15 (2) 365-371
http://www.ajnr.org/content/15/2/365

A Technique of Measuring the Precision of an MR-Guided Stereotaxic Installation Using Anatomic Specimens

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PURPOSE: To develop a method for direct measurement, using anatomic specimens, of the precision of MR-guided stereotaxic location and to describe its application to a 1.5-T MR unit with a Leksell stereotaxic frame. **METHODS:** Small pieces of gelfoam (1×1×1 mm), soaked in gadopentetate dimeglumine, were stereotaxically introduced into formalin-fixed human heads using a Leksell D (three experiments) or G (nine experiments) stereotaxic frame. The head and the frame were then introduced into a 1.5-T MR unit. The target coordinates (as set on the stereotaxic frame by one investigator) were then compared with the MR-determined stereotaxic coordinates (calculated independently by another investigator). The imprecisions Ex, Ey, and Ez in each direction were defined as the differences between the calculated and the chosen coordinates. **RESULTS:** Regarding the three targets studied with the D frame, mean imprecision Ex was 1.08 ± 0.50 mm (mean ± SEM), Ey 0.83 ± 0.58 mm, and Ez 0.75 ± 0.25 mm. For the nine targets studied with the G frame, Ex was 0.48 ± 0.17 mm, Ey 0.69 ± 0.14 mm, and Ez 0.82 ± 0.13 mm. Statistical analysis of the results showed no significant difference between Ex or Ey and half the size of a pixel, indicating that, in the axial plane, stereotaxic MR precision is limited only by pixel size. A statistically significant difference was observed in the coronal plane between Ez and half the size of a pixel, but it must be stressed that Ez remained smaller than 1 mm. **CONCLUSION:** MR-guided stereotaxic location can be used with confidence for most diagnostic, functional, and therapeutic procedures.

Index terms: Surgery, stereotaxic; Magnetic resonance, technique; Brain, magnetic resonance

AJNR Am J Neuroradiol 15:365–371, Feb 1994

Stereotaxic techniques are used to do biopsies of brain lesions, to carry out functional surgery on deep brain structures, or to perform stereotaxic radiosurgery. Computed tomography (CT) is the most commonly used method in stereotaxic neurosurgery. However, some authors still use positive-contrast ventriculography, principally for functional neurosurgery (1, 2). Recently, the use of magnetic resonance (MR) in stereotaxic conditions also has been proposed (3, 4). The advantages of using MR include the increased resolution of the lesion or target, direct nonreformatted multiplanar imaging and target coordinate determination, and reduced imaging artifacts produced by the stereotaxic frame (3). MR can be used for biopsies of CT-invisible or ill-defined lesions (5, 6), and it offers the potential for greater anatomic resolution for stereotaxic functional neurosurgery. The commissures, the thalamic organization, and individual anatomic variations are well defined by MR (3).

However, there is concern that image distortion on MR scans may displace intracranial targets from their true anatomic location (7). This can lead to errors in the determination of target coordinates and consequently to an unsuccessful stereotaxic procedure. Errors in the stereotaxic coordinates can also lead to life-threatening or

Received August 28, 1992; accepted pending revision November 2; revision received March 3, 1993.

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AJNR 15:365–371, Feb 1994 0195-6108/94/1502–0365

disabling complications. Distortions depend on the type of MR unit used (especially regarding static magnetic field strength and gradient coil design) and on acquisition parameters. They also could be related to eddy currents in the stereotaxic frame, although all manufacturers affirm that no distortion can be observed with their material (8, 9). A complete study of the distortions obtained with a given MR unit has been performed by some authors (10, 11). However, precision of a given MR stereotaxic system is difficult to extrapolate from this kind of study. Total imprecision during a given MR stereotaxic procedure is caused by the combination of: 1) intrinsic distortion of the images of the MR unit; 2) image distortions induced by the stereotaxic frame; and 3) mechanical imprecision of the stereotaxic frame. The result of the combination of these three types of errors is difficult to evaluate. It is thus extremely important to measure the precision of the MR stereotaxic system as a whole before clinical use. Recently, Kondziolka et al (12) have published a paper in which they compare MR and CT for stereotaxic coordinate determination in patients. However, to validate a given MR stereotaxic system with this method, it is necessary to perform MR and CT in stereotaxic conditions in a great number of patients. Moreover, this indicates only the difference of the results obtained with the two methods and not directly the precision of MR.

As part of a functional neurosurgery program, we needed to know the precision of our MR stereotaxic system. For this purpose, we developed a method to measure the precision of stereotaxic MR guidance using anatomic specimens. This method can be applied to any MR unit and quantifies the precision of a stereotaxic MR system.

Materials and Methods

Four human fixed heads were used in this study. A 40% solution of formaldehyde containing 5 g/L of sodium chloride was injected into both carotid arteries before the head was separated from the body. The total quantity of solution used was one-eighth of the body weight. The head was then plunged into a formaldehyde solution of the same percentage for 5 days. Thereafter the head was conserved at 4°C in a 10% solution of formaldehyde. An MR-adapted Leksell stereotaxic frame was fixed onto the head. Preliminary studies were performed with a Leksell D stereotaxic frame (Elekta Instrument, Stockholm, Sweden) on a Signa 1.5-T MR unit (General Electric, Milwaukee, Wis), and the other studies were performed on another Signa 1.5-T MR unit using a Leksell G stereotaxic frame. Both MR-compatible stereotaxic frames were constructed with nonferromagnetic aluminum alloy. A burr hole was performed at the level of the right or the left coronal suture. The head and the stereotaxic frame with MR-adapted fiducial markers were then introduced into the MR imager.

Fiducial markers on each side of the frame (Fig 1) were made with plastic disks containing channels filled with water and copper sulfate. The distance between the two lateral fiducial markers was 190 mm. The fluid-containing channels formed a square on each side. The dimensions of these squares were 120 × 120 mm. The center of each square was strictly aligned with the center of the frame. One diagonal of each square was also visualized with a channel. This diagonal can be used to calculate the Z coordinate on axial sections and the Y coordinate on coronal sections. We did not use this capability for our measurements, because the Z coordinate was directly measured on coronal sections. On an axial section, channels appeared as six bright spots (three on each side), four of which corresponded to intersections with the sides of the squares and two to intersections with the diagonals. The distances between the four spots at the extremities were 120 mm in the sagittal and 190 mm in the transverse dimension. The center of the four spots corresponded to the center of the frame in the X and Y directions. The two other spots corresponded to the diagonals of the squares. Their positions depended on the Z coordinate. On coronal sections, distances between the four spots at the extremities were also 120 mm in the sagittal and 190 mm in the transverse dimension. The center of these spots corresponded to the center of the frame in the X and Z directions. With the Leksell stereotaxic system, stereotaxic coordinates of the frame center are X = 100 mm, Y = 100 mm, and Z = 100 mm. The zero of the coordinate system is at the top, behind, and at the right of the head (Fig 1).

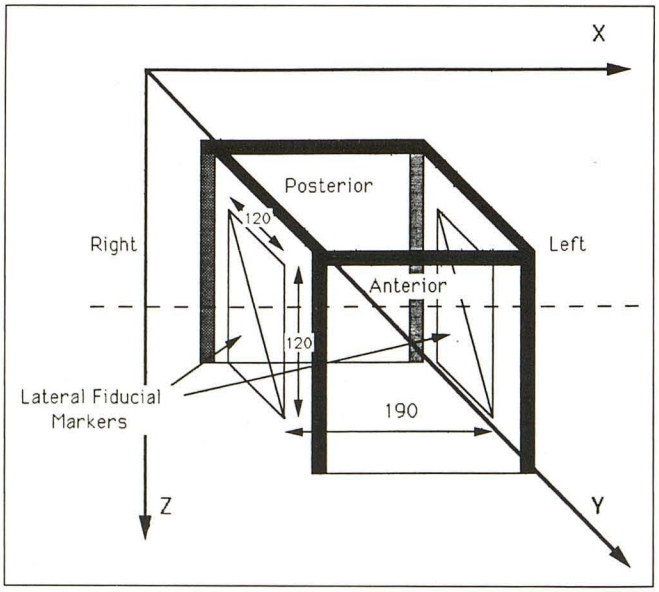

Fig. 1. Schematic drawing of the lateral fiducial markers and of the coordinate system of the Leksell stereotaxic frame.

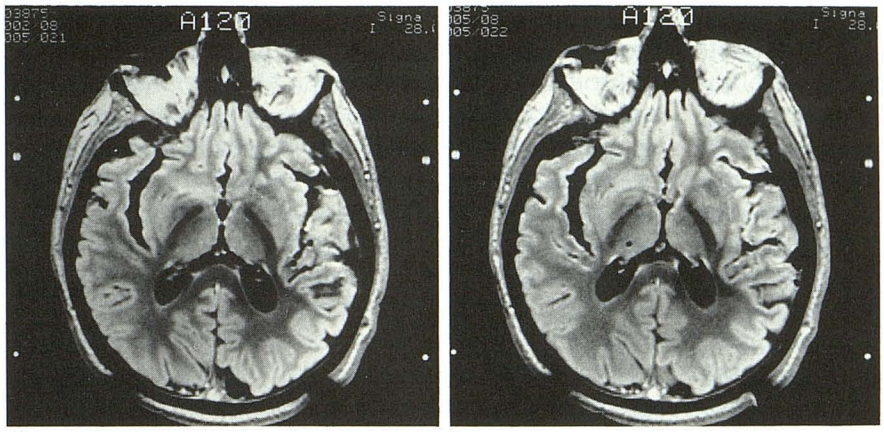

Fig. 2. T1-weighted (500/12/2) axial image of a formalin-fixed head studied with the Leksell G stereotaxic frame.

A, Before the introduction of the target. The six bright spots outside the head correspond to the lateral fiducial markers of the frame.

B, After the stereotaxic introduction of the target. The small piece of gelfoam with gadopentetate dimeglumine is clearly visible at the posterolateral part of the right thalamus (*arrow*).

Increasing X is going from right to left, Y from posterior to anterior, and Z from top to bottom.

After the introduction of the head and the stereotaxic frame into the MR imager, axial and coronal 500/20–12/2 (repetition time/echo time/excitations) spin-echo sequences were obtained. All acquisitions were obtained using the standard quadrature head coil of the MR unit. Section thickness was 3 mm, and the field of view was 240 mm. Matrix size was 256 × 256. The sections were acquired using the interleaved mode. Axial sections (from the midbrain to the upper part of the lateral ventricles, encompassing the thalamic, subthalamic, and lenticular nuclei region) and coronal sections (from the pulvinar to the frontal horns of the lateral ventricles) were obtained. Calibration of the gradients in the X, Y, and Z directions was checked on axial and coronal sections. On axial and coronal sections, the distance between the spots of the fiducial markers was checked in the sagittal (120-mm) and transverse (190-mm) dimension. If any significant discrepancy was found between the measured and exact distance, corrections of the appropriate gradients were done using the research mode of the Signa MR unit. These corrections were used for all further measurements. A set of coordinates was chosen on axial and coronal sections, delineating a parallelepipedic region in the thalamic, capsular, and lenticular regions of the chosen hemisphere. These regions avoided the lateral ventricle and the sylvian fissure. The size of this region was 20 mm in the X, 40 mm in the Y, and 20 mm in the Z direction. The stereotaxic frame and the head were then taken out of the MR unit, and a Backlund needle (Elekta Instrument) was stereotaxically introduced through the burr hole into the deep structures, within the defined parallelepipedic region. A 1 × 1 × 1-mm piece of gelfoam, soaked in gadopentetate dimeglumine, was then pushed through the needle into the brain. One investigator designated the coordinates of the positioned target at random (as set on the stereotaxic frame) inside the previously determined parallelepipedic region. The other investigator was not aware of the chosen values.

The stereotaxic frame with the head was then reintroduced into the MR unit, and 3-mm axial and coronal sections were obtained encompassing the parallelepipedic region. The stereotaxic coordinates of the introduced target were calculated by the second investigator on the basis of MR data. The X and Y coordinates were determined on axial images and the Z coordinate on frontal images. The coordinates were calculated using the standard software of the MR unit. The position of the frame center with reference to the landmark of the MR unit was determined from the position of the four spots seen at the extremities on the axial and coronal sections. The stereotaxic coordinates of the target were then inferred by calculating its distance from the center in two orthogonal directions. Calculated MR stereotaxic coordinates were compared with those chosen on the stereotaxic frame. The imprecisions Ex, Ey, and Ez in each direction were defined as the difference between the calculated and the chosen coordinates.

Twelve experimental studies were carried out. Three preliminary experiments were performed using the Leksell D frame on the first MR unit. Two heads were used for these experiments. One target was introduced in the first specimen and two in the second. Nine experiments were performed on two other heads using the Leksell G frame on the second MR unit. Five targets were introduced into one head and four into the other. We never introduced more than one target during a procedure, and measurements were done after stereotaxic introduction of each target.

Results

The typical aspect of a formalin-fixed brain with the short-repetition-time/short-echo-time MR sequence used in our experiments is shown in Figures 2 and 3. The signal-to-noise ratio and contrast were excellent. Gray/white matter contrast was inverted as compared with the results on living subjects. Small amounts of air inside the ventricles or at the sylvian fissure sometimes gave rise to limited magnetic susceptibility artifacts at the border of the ventricles or the sylvian fissure, but they did not interfere with the measurements. The introduced targets (Figs 2B and 3B) appeared as small round foci of signal void

with increased signal at the periphery. The position of the target was measured at the center of the round signal void area. Comparison of the frame (Xf, Yf, and Zf) and MR-calculated (Xc, Yc, and Zc) coordinates for the three targets studied with the D and the nine targets studied with the G frame is shown in Table 1. The measured imprecision in the X, Y, and Z directions (differences between the calculated and the frame coordinates) for the 12 targets is also given. Regarding the three targets studied with the D frame, mean imprecision in the X direction (Ex) was 1.08 ± 0.50 mm (mean ± SEM), in the Y direction (Ey) 0.83 ± 0.58 mm, and in the Z direction (Ez) 0.75 ± 0.25 mm. For the nine targets studied with the G frame, mean imprecision Ex was 0.48 ± 0.17 mm (mean ± SEM), Ey 0.69 ± 0.14 mm, and Ez 0.82 ± 0.13 mm. No statistically significant difference was found between the imprecisions measured with the D and the G frame (Student *t* test). Furthermore, with both frames, there was no statistically significant difference for the imprecisions in the X, Y, and Z directions (Student *t* test). Mean imprecision with the G frame was then compared with the theoretical limit of the precision of the method, conditioned by the use of a digitalized imaging method. This theoretical limit is half the size of a pixel; that is, a/2 = field of view/2 × 256 = 0.47 mm, where a is the size of one pixel. No statistically significant differences were found between Ex, Ey, and a/2, but Ez was significantly higher than a/2 (*t* test, $P = .05$).

Discussion

Accuracy of Stereotaxic MR

Stereotaxic neurosurgery is an actively developing field. Determining the stereotaxic coordinates of intracerebral targets using MR has become of increasing interest for cerebral biopsy, functional neurosurgery, and stereotaxic radiosurgery (3, 5, 6, 12). However, the possibility of image distortion has raised some questions regarding the accuracy of MR for stereotaxic co-

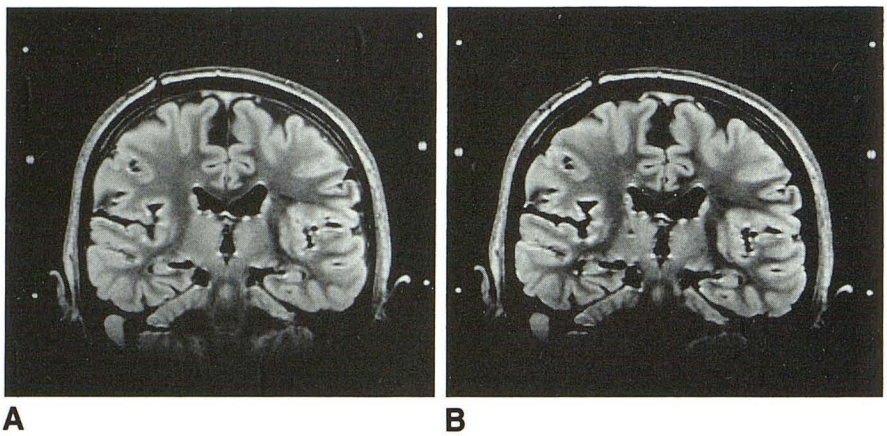

Fig. 3. T1-weighted (500/12/2) coronal image of the same head studied with the Leksell G stereotaxic frame.

A, Before the introduction of the target.

B, After the stereotaxic introduction of the target. The small piece of gelfoam with gadopentetate dimeglumine is also clearly visible at the posterolateral part of the right thalamus (*arrow*).

TABLE 1: Comparison of the frame coordinates (Xf, Yf, and Zf) and the MR-calculated coordinates (Xc, Yc, and Zc) for the 12 targets

No.	Xf	Xc	Ex	Yf	Yc	Ey	Zf	Zc	Ez
1	120	119	1	105	103	2	105	105.5	0.5
2	98	98.25	0.25	97	96.75	0.25	103	103.75	0.75
3	81	83	2	115	114.75	0.25	113	114	1
4	85	85	0	90	90.75	0.75	130	130.75	0.75
5	85	84.25	0.75	112	111.75	0.25	125	126.5	1.5
6	80	81.25	1.25	100	101	1	135	136.4	1.4
7	85	85	0	110	110.5	0.5	130	130.75	0.75
8	120	121	1	100	101.25	1.25	130	130.75	0.75
9	82	81.3	0.4	108	107.75	0.25	121	120.25	0.75
10	89	89	0	81	80.75	0.25	118	117.25	0.75
11	77	77	0	94	95.25	1.25	127	127.25	0.25
12	75	76	1	101	101.75	0.75	115	115.5	0.5

Note.—Ex, Ey, and Ez are the measured imprecisions. Targets 1 to 3 were studied with the D and 4 to 12 with the Leksell G Stereotaxic frame.

ordinate determination (7, 13, 14). The primary factors that introduce geometric distortion are inhomogeneity in the magnetic field and nonlinear magnetic field gradients. Verification of the homogeneity of the magnetic field and the search for image distortions caused by nonlinear magnetic field gradients are prerequisites for the use of an MR unit for stereotaxic purposes. These parameters were checked for both MR units used in this study (15, 16). However, the value of the precision of a stereotaxic procedure is difficult to extrapolate from such procedures, and, moreover, additional imprecision might be caused by distortions induced by the stereotaxic frame, including the fixation pins, and/or by mechanical imprecision of the frame.

Study of the precision of a given MR stereotaxic system is thus mandatory before use in clinical conditions. Some authors (3, 17, 18, 19) have compared stereotaxic data obtained with MR and CT studies. The largest and most recent series has been published by Kondziolka et al (12). Validation of a given MR stereotaxic system with this type of method requires that both CT and MR stereotaxic procedures be done in a relatively large series of patients. Moreover, this type of method only allows comparing the results obtained with the two methods. Differences observed between MR and CT coordinates can be caused by errors in MR but also in CT determination of coordinates. In order to avoid these practical and methodologic problems, we have thus developed a technique that determines the true precision of a given MR stereotaxic system independently of any other method of stereotaxic coordinate determination.

Principle of the Method

A stereotaxic procedure is based on the fixation of the head in a stereotaxic frame followed by the radiologic determination of the coordinates of the target with reference to the stereotaxic frame. A biopsy needle, an electrode, or any other tool then can be accurately positioned using the stereotaxic frame at the target coordinates. Imprecision of the procedure can be defined as the distance between the target (or the center of the target) and the needle's extremity. This imprecision is the sum of: 1) the errors in the radiologic determination of the coordinates; and 2) the mechanical imprecision of the stereotaxic frame. In the method we used, we measured the difference between the coordinates of the target as set on the stereotaxic frame and the coordinates of the target as determined with MR in stereotaxic conditions. With this method the error in positioning of the target results from the mechanical imprecision of the frame. Consequently, by measuring the position of the target in the stereotaxic frame, we add to this error the radiologic imprecision in the determination of the MR coordinates. The difference between MR coordinates and frame coordinates is thus the sum of both: 1) the imprecision of MR determination of the coordinates; and 2) the mechanical imprecision of the frame. Thus, the imprecision we measured is the same as that of the stereotaxic procedure in clinical conditions. In addition, it must be considered that with our method the positioning of the target inside the fixed brain could add a supplementary imprecision. Consequently, it can be stated that the real precision of a given stereotaxic system will be equal or superior to the precision measured using our technique. However, in order to obtain a correct measurement of the precision of the system, we stress the importance of positioning the target with great accuracy. The target must be at the zero point of the needle. In our experience, the use of a Backlund needle was decisive for the precision of the study. The opening of this type of needle is at its extremity (it is not a lateral hole as in the more frequently used Sedan biopsy needle). This allowed us to position the target with great accuracy.

Methodologic Considerations

Although our method allows measurements of stereotaxic MR precision in almost the same conditions as in a clinical procedure, some differences must be stressed: 1) Formalin-fixed cadaver brain is more rigid than live brain tissue; consequently, displacement caused by the needle-puncture wound could seem less likely with our method than it may be in vivo. However, during stereotaxic procedures in patients, displacement of the target by the needle (or electrode) has not been described as a potential pitfall even in very precise functional stereotaxic procedures (2, 3, 4, 12, 20, 21). 2) As in most current MR or CT stereotaxic clinical procedures, our studies were performed with no needle in place. However, it should be considered that clinical applications with a needle in place could introduce additional artifact and/or image distortion. 3) This method permits determination of the precision of a given MR stereotaxic system. Extrapolation of the result of one

system to another installation could be hazardous. In particular, the use of an MR unit with less homogenous magnetic field and/or gradient linearity defects could affect the precision of the procedure. Small symmetrical distortions of the images would probably not affect the precision of the method significantly (because this would not displace the center of the frame), but asymmetrical distortions could lead to very important imprecision.

Measurement of the Accuracy of the 1.5-T MR-Guided Stereotaxic System

Contrast observed in MR of formalin-fixed brains already has been described (22, 23). Inversion of gray/white matter contrast is caused by important T1 shortening, which predominates in the gray matter. The target we used in our experiments was chosen after unsuccessful preliminary attempts using injection of a paramagnetic contrast agent through the stereotaxic needle or introduction of pieces of glycerin. Injection of the paramagnetic agent in a formalin-fixed brain was impossible because the cerebral structures are stiff, and the pieces of glycerin were not seen on MR sections because of the high MR signal of the fixed brain.

After a preliminary study using the Leksell D frame, we studied the precision of our MR stereotaxic system using the Leksell G frame. The study was limited to a parallelepipedic region including the thalamus, capsular, and lenticular areas, because we were principally interested in the use of MR in functional neurosurgery, and we wanted to validate its use in the thalamic and lenticular regions. Locations of the targets are thus quite near the magnet isocenter, where errors caused by magnetic field inhomogeneity and gradient nonlinearity are at minimum. The measured precision is thus valid for this region of the brain. In addition, it should be stressed that stereotaxic MR is principally interesting for deep lesions (brain stem or basal ganglia lesions for example). In the case of a relatively peripheral lesion it is possible to fix the frame on the head in an asymmetrical position in order to place the lesion near the center of the frame and thus near the magnet isocenter. However, the use of stereotaxic MR for very peripheral lesions should necessitate measurement of the imprecisions observed in these regions.

Statistical comparison of the mean imprecisions found with the G frame in the axial plane ($Ex = 0.48 \text{ mm} \pm 0.17$; $Ey = 0.69 \text{ mm} \pm 0.14$) showed no significant difference with half the size of one pixel (0.47 mm). This value is the theoretical limit of the precision using a digitalized imaging technique. Considering the coordinate in the X direction (the demonstration is exactly the same in all the spatial directions), if the true coordinate X is in the nth pixel, that is, between $(n - 1)\,a$ and $n\,a$ (a is the size of one pixel), the MR-determined coordinate has the same probability to be $(n - 1)\,a$ or $n\,a$. Thus, imprecision is with the same probability $X - (n - 1)\,a$ or $n\,a - X$. Mean value of the imprecision is thus $1/2\,\{X - (n - 1)\,a + n\,a - X\} = a/2$. Thus, our results suggest that, in the axial plane, pixel size was the principal limiting factor for stereotaxic MR precision. With our imager, maximal matrix size is actually 256. The field of view in stereotaxic MR must be larger than the largest dimension of the frame (190 mm). It was thus impossible to obtain a pixel size much smaller than the one we used. In the future, the development of MR imaging with a matrix size of 512 should increase the precision of stereotaxic MR. In the coronal plane, a significant difference was observed between the observed mean imprecision (0.82 ± 0.13 mm) and half the size of a pixel. However, the imprecision of the method remained smaller than 1 mm.

Our results confirm that stereotaxic MR is a very precise method. This is in agreement with the conclusions of Kondziolka et al (12). These authors compared CT and MR values and found mean differences of 1.19 mm, 1.55 mm, and 2.04 mm, respectively, in the X, Y, and Z directions. Other authors (18, 19) have compared stereotaxic data obtained with MR and CT studies and obtained larger differences than Kondziolka et al. Mean differences of 1.0 and 3.75 mm in the X and Y directions, respectively, were reported by Bradford et al (18). Heilbrun et al (19) found the average error for the coordinates to be as great as 5 mm. All these papers report values of stereotaxic MR imprecision larger than the one we obtained by measuring directly the precision of stereotaxic MR. One of the reasons for these larger measured imprecisions is that the CT scan is used as the reference method, thus adding the errors of the two methods.

Utility of MR for Functional Neurosurgery

Stereotaxic ventriculography is still advocated by some authors (1, 2) to perform functional

neurosurgery, but numerical data on its precision, to our knowledge, are lacking. It is thus difficult to compare stereotaxic ventriculography results with our data on stereotaxic MR accuracy. The main drawback of ventriculography is that it is an invasive method and that data on the laterality of the thalamic target are purely statistical because direct visualization of the thalamus is impossible. Many authors propose the use of CT to avoid ventriculography (20, 21, 24). At present, determination of the thalamic nuclei position with CT is done indirectly. It is based on statistical data of nuclei position obtained from stereotaxic atlases (24, 25), after CT determination of the coordinates of the anterior and posterior commissures. CT determination of anterior commissure coordinates is often difficult in the cranio-caudal direction. More generally, accurate determination of the Z coordinate (cranio-caudal direction) using only axial sections is difficult with CT. Our results on stereotaxic MR precision confirm, as proposed by other authors (3, 4), that MR can be used for functional neurosurgery. The ability of MR to perform multiplanar imaging results in a greater accuracy in determination of anterior and posterior commissure coordinates than that of CT, especially in the cranio-caudal direction. Moreover, with appropriate sequences, MR permits direct visualization of the thalamus and part of the thalamic nuclei. Perhaps for the future, direct determination of intrathalamic targets can be envisioned.

In conclusion, we have presented a method that measures the precision of an MR stereotaxic installation using anatomic specimens. This method can be used on any MR imager with any MR-adapted stereotaxic frame. Applying this method, we have demonstrated that MR-guided stereotaxic location can be used with confidence for diagnostic, therapeutic, or functional procedures.

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