Accuracy of Customized Miniature Stereotactic Platforms

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Stereotaxy · Registration · Fiducial marker · Deep brain stimulation · Stereotactic surgery

Abstract
In this study, a new system was evaluated for implanting deep-brain stimulators based on a one-piece platform for each trajectory customized from a preoperative planning image. During surgery, the platform is attached to skull-implanted posts that extend through the scalp. The platform acts as a miniature stereotactic frame to provide guidance for parallel cannulas as they are advanced through a burr hole to the target. Accuracy is determined from a postoperative CT. For each implantation, the distance between the position observed in the postoperative image and the position calculated relative to the platform from the preoperative image is our measure of error. Because this measure incorporates the surgical error of electrode anchoring, brain shift between preoperative and postoperative scanning, and error in the measurement of the position of the electrode in CT, it will tend to overestimate the true error. The mean error was 2.8 mm for 20 implantations. These data reflect favorably the accuracy of this system when compared with others.

Introduction
The first FDA approval in 1998 of deep-brain stimulation (DBS) in the treatment of movement disorders has gained significant popularity \cite{1, 2}. The therapy has significant application in the treatment of tremor, rigidity, and drug-induced side effects in patients with Parkinson’s disease and essential tremor. The use of a 4-contact electrode, as shown in figure 1 (Medtronic No. 3387 or No. 3389 quadripolar lead; Medtronic, Inc., Minneapolis, Minn., USA) placed within deep brain targets ranging from 4 to 12 mm in diameter requires stereotactic neurosurgical methodology. Ideally, the optimal target for therapy should be located within the stimulation range of 1 or 2 contacts, each contact measuring 1.5 mm separated by either 1.5 mm (lead No. 3387) or 0.5 mm (lead No. 3389). Effective stimulation results when the contacts lie centrally within the treated nucleus \cite{3}. At our institution, we prefer that 2 contacts lie above and 2 below the ideal target. If the contacts are located as little as 3–4 mm away from the desired target, ineffective stimulation can result for several reasons: (a) failure to capture control of the group of neurons, (b) stimulation of nondesirable areas resulting in unpleasant stimulation or (c) necessity for higher stimulus intensities to produce the desired effect resulting in reduced battery life of the implant. For these reasons, targeting the specific neurons of interest for this...
therapy requires millimetric precision and allowance for variability among patients. Hence, the process of implantation of a DBS electrode follows a step-wise progression of (a) initial estimation of target location based on imaged anatomical landmarks, (b) intraoperative microanatomical mapping of key features associated with the intended target of interest, (c) adjustment of the final target of implantation by appropriate shifts in three-dimensional space, and (d) implantation of the quadripolar electrode with contacts surrounding the final desired target. In each of these steps, error may be introduced by the methodology of imaging the brain structures, planning the target, translating the target plan to an appropriate stereotactic device, manipulating the stereotactic device and interpreting the locations of test electrodes, and implantation of the final electrode.

Traditional methodology for carrying out this stepwise target localization and implantation procedure has been based on an externally fixed, rigid fixture, called a ‘stereotactic frame’ that encompasses the patient’s head and upon which the micromanipulating equipment can be mounted and maneuvered with precision (fig. 2). These various stereotactic frames have been optimized to obtain accurate images subsequently used to create the initial target trajectory and plan and then to reduce erroneous movement associated with passage of the test electrodes and the final implant [4, 5]. These frames typically require mounting the day of surgery, subsequent imaging with either CT and/or MR axial slices, and target planning prior to starting the actual procedure of intraoperative mapping and ultimate implantation of the electrode at the final target. Patients are minimally sedated or expected to be awake for the duration of this procedure due to the need for awake-neurological examination [6, 7]. Consequently, during image acquisition, motion artifact is encountered even if the patient is fixed to the CT or MRI table, especially if the patient has significant tremor for which they are being treated. This movement theoretically reduces the accuracy of image coregistration between CT and MRI. Furthermore, once images are acquired, time is required to process and coregister the images, perform the trajectory planning, and then translate the coordinates to the frame. The delay involved with frame placement, image acquisition, and trajectory planning can be several hours in some institutions and reduces critical operating room time necessary to perform the actual mapping of the relevant deep brain structures and testing of the electrode implant before the patient tires. This has prompted us to explore other methods in which image acquisition, coregistration, and trajectory planning may be accomplished prior to the day of stereotactic surgery.

Another patient discomfort encountered in the use of traditional frames is the need to secure the frame to the operating room table to avoid excessive movement and support the patient’s head due to the weight and bulk of the frame (fig. 2). Most patients with Parkinson’s disease that undergo stereotactic mapping of deep brain structures arrive off medications to allow examination for changes in rigidity associated with effective target identification when high frequency stimulation is applied. Patients often become intolerant to prolonged restriction in head movement associated with a rigidly fixed frame to the bed, thereby creating a patient tolerance dilemma when the target is difficult to determine and prolonged testing is needed. Sometimes targets are not identified easily and the procedure may last for the better part of
Accuracy of Miniature Platforms

The potential advantages of this type of stereotactic system have motivated us to pose the following questions in deciding whether this frame can accurately and precisely allow for implantation of DBS electrodes: (1) Is virtual planning of DBS stereotactic trajectories and rapid prototype manufacturing of customized miniature platforms a clinically practical process? (2) What is the electrode placement error (EPE) between the target designation and the initial electrode placement? Although the first question addresses the manufacturing and application of this technology, testing the accuracy of this method of stereotaxy requires a novel approach to answer the latter question. The reason for this is the fact that construction of each platform is customized and is based on a relationship to randomly fixed fiducial bone markers on the skull. Therefore, checking the accuracy of a given platform to a standardized reference frame, as is done with other standard stereotactic frames, is not feasible.

The FDA clearance of this platform was based on manufacturing standards that have already demonstrated submillimetric targeting accuracy, but the verification of that accuracy was based only on the manufacturer’s phantom experiments. There have heretofore been no studies of the accuracy of the system in a clinical setting. Consequently, the accuracy of this method of stereotaxy was tested at our institution by validating the location of the final target as seen by the implanted electrode, and comparing this with the initial planned target by which each platform was designed. In order to determine the practicality of this process and measure the error of implantation, we received IRB approval at our institution (Vanderbilt University IRB No. 01-0809) to investigate this process in an initial group of 30 patients. Experience with the first few patients showed that the answer to the first question is clearly yes, but the answer to the second question requires careful measurements on as many patients...
as possible. Much of the error is based on the stability and
certainty of the frame, but it also reflects the surgical pro-
cess as a whole, so it is not possible to answer the question
with phantom studies alone. It is the goal of the present
study to measure EPE for patients undergoing implanta-
tions of deep-brain stimulators.

**Materials and Methods**

Patients that were selected for either unilateral or bilateral DBS
implantation were candidates for this study. Either one (unilateral
implantation) or two (bilateral) platforms (fig. 3a) were thus manu-
factured for each patient, and each platform represents a planned
target. The method for placing the bone-based fiducial markers,
image acquisition and coregistration, platform planning and usage
are described next.

First, under anesthesia the fiducial bone markers are implanted
(Acustar™; Z-Kat, Inc., Hollywood, Fla., USA) [9]. CT (fig. 4) and
MR (not shown) volumes are acquired with the patient anesthe-
tized and head taped to the table to minimize motion. After imag-
ing, the marker caps are removed leaving only bone-implanted
posts in place, and the patient is sent home with wound care in-
structions. With the help of MR-CT registration software (VoXim,
FHC Inc., Bowdoinham, Me., USA), the surgeon selects the initial
target points based on standard AC-PC coordinates and associated
entry points on the surface of the skull. In addition, the centroids
of the markers and the directions of their posts are determined. For
each trajectory (unilateral or bilateral), a file is created containing
the directional instructions for the manufacturer (FHC) to construct
a unique platform that will mate with three marker posts. These
data are sent electronically to a fabrication plant where a custom-
ized platform is then manufactured and shipped to the site of
the surgery within 1 week.

Stereotactic mapping and electrode implantation occur within
1–2 weeks of the marker placement. The platform(s) are then ster-
ilized and used to mount routine microrecording and stimulating
electrodes using standard techniques described elsewhere [10, 11].
In brief, the patient is brought to the operating room and placed in
a recumbent position on a soft head holder. The miniature plat-
forms are then fitted to the bone marker posts, and local anesthetic
infiltrated at the intended entry points in the skull. A burr hole is
created coaxial with the trajectory provided by each platform. A
micropositioning drive (microTargeting® drive system, FHC Inc.,
Bowdoinham, Me., USA) is attached to each platform (fig. 5). Mi-
cro-electrode recording leads are advanced into the patient to the
respective initial target positions through the central guide tube of
the platform (not visible in fig. 5, but shown in fig. 3). Neurophysi-
ological data are noted and the target positions are reassessed in
the context of these data. The microelectrodes are removed and a uni-
polar macrostimulation lead (TM electrode; Radionics, Corp., Bur-
lington, Mass., USA) is inserted through an appropriate trajectory
as determined by the micro-electrode recordings. With the patient
awake, response to stimulation is monitored as the position of the
macrostimulating electrode is advanced. Voltages necessary for
effective stimulation are weighed against undesirable side effects and
a final target point is determined by the neurosurgeon. When the
final target is selected, the DBS lead is inserted to the measured
depth for which two contacts are located above and two are below
the target. The free end of the DBS lead is then buried beneath the
scalp to be connected to the internal pulse generators at a future
date. A second DBS electrode may be implanted on the contralat-
eral side on the same day. At the end of the procedure, the platform
is removed. Within 24 h of surgery, the imaging marker caps are
reattached to the posts, and a postoperative CT scan is acquired
(fig. 6). If no complications occur, the patient is discharged home.
within a day of this surgery. Furthermore, the final electrode position in the postoperative CT is then compared with the intended target measured relative to the platform, whose position is determined from the markers on the preoperative CT scan.

The goal of this study is to determine the geometrical accuracy of the system for electrode placement. The DBS electrode placement is altered from the target position initially chosen by the surgeon on the basis of physiological recordings and responses during the procedure. The situation is shown schematically in figure 7. The initial target (a) is missed by the micro-recording electrode when it is initially placed at (b). The discrepancy $\mathbf{ab}$ between these two positions is the EPE that we wish to measure. During surgery, the choice of position is altered by adjusting the depth of the electrode with the microdrive and by choosing alternate guide tubes in the platform (fig. 3b). The vector $\mathbf{ac}$ from (a) to the new target position (c), which is calculated by reading the markings on the microdrive and by noting the final guide tube selected, is the target displacement. The position (d) is the measured position of the DBS electrode on the postoperative CT. By subtracting $\mathbf{ac}$ from $\mathbf{ad}$ we get the vector $\mathbf{cd}$. If the displacement calculated by the system is accurate, then $\mathbf{cd}$ is equal to $\mathbf{ab}$. Our endpoint is the length of $\mathbf{cd}$. It is our measure of EPE. Inaccuracy in the system’s measurement of $\mathbf{ac}$, the surgical error of electrode anchoring, anatomical shift of the brain between the preoperative planning image and the postoperative CT, and error in the measurement of the position of the electrode in CT, will all be folded into the error that we measure and will tend to exaggerate our estimates of the mean or root-mean-square values of the length of $\mathbf{ab}$. The result in this case is that the EPE that we measure is an overestimate of the platform’s actual EPE.

The points that we use for these calculations must all be referred to the same image space. We choose the postoperative CT space. As the initial target position is given in the preoperative CT space, we must register preoperative to postoperative. That registration is effected as follows: The fiducial marker centroids are determined in both CT images using an automatic algorithm designed for the Acustar markers [12]. (CT images acquired at kVp = 120 V, exposure = 350 ms, 512 × 512 pixels ranging in size from 0.49 to 0.62 mm, slice thickness = 2 mm for 1 patient, 1.3 mm for 2 patients, 1 mm for all others.) The corresponding points are brought into registration by means of a rigid-body transformation determined using a standard algorithm (algorithm 8.1, 8). The determination of the electrode centroid, point (d) in figure 7 is accomplished by means of an algorithm designed especially for the electrodes employed in this study and reported on earlier [13].

**Results**

Twenty-one patients have been implanted, 9 unilaterally and 12 bilaterally for a total of 33 implantations. Eight patients – 5 bilateral and 3 unilateral – were excluded from our study on the basis of difficulties with the postoperative CT. All problems stemmed from the change in imaging procedure associated with the switch from a standard frame to the platform.

In 4 bilateral patients and 1 unilateral patient, some of the Acustar markers were not included in the postoperative CT field of view. This problem occurred because the CT technicians were unaware of the need to insure that the markers were in view for the purposes of this study.

In one bilateral and two unilateral patients, large motion artifacts were visible in the postoperative CT related to the patient being awake and free to move, typically due to preexisting movement disorder.

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**Fig. 6.** Postoperative CT image of patient after bilateral implantation. DBS leads are indicated by a and wire leads running under skin from DBS leads to pulse generator by b.

**Fig. 7.** Schematic of placement error. The image shows initial target chosen on preoperative image (a), initial placement of recording micro-electrode (b), final position of DBS electrode as calculated from adjustment ac made during implantation (c) and final position of DBS electrode centroid as measured in the postoperative CT (d). EPE is equal to vector $\mathbf{ab}$. $\mathbf{ab} = \mathbf{cd}$, which is in turn equal to the vector difference, $\mathbf{ad} – \mathbf{ac}$. 
Hence, out of a total of 33 implanted electrodes, 13 were discarded and 20 were used in the study. Note, however, that these problems were related only to the postoperative scan and were unrelated to the procedure itself. No data were discarded because of difficulty with electrode placement. A satisfactory physiological target was found and electrode positioned in all 33 implantations.

The x, y, and z components of the EPE were recorded for each electrode, where x, y, z are lateral, anterior-posterior, and inferior-superior components measured along the postoperative CT axes. This vector was determined by the technique described in the previous section (i.e. by subtracting \( \mathbf{a_c} \) from \( \mathbf{a_d} \)). Our results are summarized in table 1.

### Table 1. Statistics on measured EPE (in mm)

<table>
<thead>
<tr>
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<th>EPE</th>
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<tbody>
<tr>
<td></td>
<td>Lat</td>
<td>AP</td>
<td>SI</td>
<td>Mag</td>
</tr>
<tr>
<td>RMS</td>
<td>1.8</td>
<td>1.7</td>
<td>1.4</td>
<td>2.8</td>
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<tr>
<td>Mean</td>
<td>0.4</td>
<td>−1.1</td>
<td>−0.6</td>
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<tr>
<td>SD</td>
<td>1.8</td>
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<td>1.3</td>
<td>0.84</td>
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<tr>
<td>SEM</td>
<td>0.40</td>
<td>0.30</td>
<td>0.30</td>
<td>0.19</td>
</tr>
<tr>
<td>Mean Abs</td>
<td>1.4</td>
<td>1.4</td>
<td>1.1</td>
<td>2.7</td>
</tr>
<tr>
<td>Max Abs</td>
<td>4.0</td>
<td>3.2</td>
<td>2.9</td>
<td>4.1</td>
</tr>
</tbody>
</table>

RMS = Root mean square; Abs = absolute value. There were 20 implanted electrodes in 13 patients.

### Table 2. EPEs from Cuny et al. [14] (14 placements) and Starr et al. [6] (76 placements)

<table>
<thead>
<tr>
<th></th>
<th>EPE</th>
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</thead>
<tbody>
<tr>
<td></td>
<td>Lat</td>
<td>AP</td>
<td>SI</td>
<td>Mag</td>
</tr>
<tr>
<td>Cuny et al.</td>
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<td></td>
<td></td>
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<tr>
<td>RMS</td>
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<td>3.7</td>
<td>1.2</td>
<td>4.1</td>
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<tr>
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<td>−0.06</td>
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<tr>
<td>SD</td>
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<td>3.7</td>
<td>1.2</td>
<td>N/A</td>
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<tr>
<td>SEM</td>
<td>0.35</td>
<td>0.99</td>
<td>0.32</td>
<td>N/A</td>
</tr>
<tr>
<td>Starr et al.</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>RMS(^1)</td>
<td>1.8</td>
<td>2.0</td>
<td>2.2</td>
<td>3.4</td>
</tr>
<tr>
<td>Mean</td>
<td>−0.75</td>
<td>0.76</td>
<td>−0.69</td>
<td>N/A</td>
</tr>
<tr>
<td>Mean Abs</td>
<td>1.6</td>
<td>1.4</td>
<td>1.7</td>
<td>N/A</td>
</tr>
<tr>
<td>Max Abs</td>
<td>4.7</td>
<td>5.1</td>
<td>8.8</td>
<td>N/A</td>
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</tbody>
</table>

N/A = Not available.

\(^1\) Calculated from mean values and mean absolute values by assuming uncorrelated, normally distributed error.

### Discussion

The platform is a miniature stereotactic head frame. It is far smaller and lighter than standard stereotactic frames, and, unlike the standard frames, it allows free head motion. Furthermore, unlike the standard frames, it makes possible the simultaneous adjustment of dual probes during bilateral implantations. These advantages are significant, but only if the system provides requisite accuracy in the initial placement of the probe. While the initial placement is typically adjusted intraoperatively in response to microrecordings and to the observed effects of stimulation, error in that initial placement can add to the required number of passes of probe into brain and to increased procedure time. Sufficiently large errors can result in placements beyond the adjustment limits, thus making optimal placement impossible.

### Interpretation of Errors

The major goal of this study was to determine the accuracy of that initial electrode placement. On the basis of the registration of preoperative to postoperative CT image, as explained in the previous section, we measure the EPE in millimeters individually for each implanted stimulator. Statistics are given in table 1. Lat, AP, and SI, are the components of EPE in lateral, anterior-posterior, and inferior-superior directions. It can be seen, from the relative magnitudes of the standard deviations and the absolute values of the means of these three quantities, that there is no directional bias in the placement detectable in our study.

‘Mag’ is the magnitude of error. In the Materials and Methods section, several sources of additional error are listed that add to the measurement of EPE. An additional contribution arises from error in the registration of the preoperative and postoperative CT images based on the fiducial markers. These errors combine to exaggerate (to an unknown extent) our estimate of the true error. Thus, our best estimate based on these 13 patients and 20 implantations is that RMS EPE with the platform is below 2.8 mm and the mean absolute EPE is below 2.7 mm.

### Comparison with Published Data on Standard Frames

Two other groups have published measurements of EPE, each employing the Leksell stereotactic frame (Elekta Instruments, Stockholm, Sweden). Cuny et al. [14] presented error measurements for 14 bilateral implantations (28 electrode placements). They selected the target position on a preoperative MR, localized the electrodes on a
postoperative CT and used registration software to register the two images. Starr et al. [6] presented error calculations for 62 unilateral implantations and 7 bilateral (76 electrode placements). They selected the target position on a preoperative MR and localized the electrodes on a postoperative MR, using the center of an artifactual void in the image as the electrode centroid. The disparity between the two positions was determined relative to AC-PC coordinates in each image. As with our study, neither group observed directional bias in placement, but the EPE was considerably larger for these studies than for the present one. RMS EPE for the study by Cuny et al. [14] was 4.1 mm (from ‘3D MR imaging’ data in their table 1). RMS EPE for the study by Starr et al. [6] is not directly available but with the assumption of normally distributed error, we can estimate that their RMS EPE was 3.4 mm (from data in their table 5) [13]. Data from these two studies are given in table 2. The corresponding numbers in the two tables can be compared, except for the maximum absolute value, which depends strongly on the number of placements. It can be seen that most of the errors that we have measured for the platform are smaller than those measured for these studies of the Leksell frame.

Because of the retrospective nature of our comparison with these earlier studies and the difficulty of interpreting historic data, we do not claim that the new system is more accurate than the Leksell system. It does seem clear, however, that the overall accuracy of this new approach is very likely to be as good as, or better than, that reported so far with the larger frame.

Conclusions

Our goal in this study was to measure the initial EPE for patients undergoing the implantation of deep brain stimulators using the microTargeting platforms. Both initial target selection and final placement verification were based on CT images acquired while patients were anesthetized and immobilized. Our results are summarized in table 1 and are compared with similar error measurements, summarized in table 2, for the same procedure as described by other groups using a standard stereotactic frame. We observed an RMS placement error with the platform of 2.8 mm. This value compares very favorably with the corresponding values of 4.1 and 3.4 mm from the published results of groups using a standard frame. Furthermore, a satisfactory physiological target was found and electrode positioned in all patients we tested.

The additional advantages of increased patient comfort through separation of image acquisition and target planning from the day of stereotactic surgery, lack of restraint of the patient, the ability to reduce motion artifact by image acquisition under anesthesia other than the day of surgery, the potential to do bilateral procedures, and the light weight of this stereotactic methodology offer a unique alternative to traditional neurosurgical stereotaxy.

References