A Real-Time System for 3D Neurosurgical Planning

John C. Goble, John W. Snell, Ken Hinckley and Neal F. Kassell

Neurosurgical Visualization Laboratory
Department of Neurological Surgery
The University of Virginia
Charlottesville, VA 22908

Abstract

We have designed and implemented a computer-based system that permits rapid acquisition of digital medical images, multi-modality registration and segmentation, and three-dimensional planning of minimally invasive neurosurgical procedures. The system, known as Netra, is optimized for real-time planning: imaging, pre-processing and planning are performed on the morning of surgery in clinically useful times. We have tested the system on procedures such as needle biopsies, depth electrode placements and craniectomies for arteriovenous malformations, aneurysms and tumors.

We describe in this paper the core algorithms of our system, and discuss issues related to implementation, validation and user acceptance. We focus on techniques for physician interaction that encourage active participation by the surgeon as principal operator of the visualization and planning system.

1. Introduction

New computer and communications technologies are fundamentally changing the role of medical imaging in the care of patients with neurologic disease. Previously restricted to diagnosis, the images from magnetic resonance scanning (MR), computerized tomography (CT), positron emission tomography (PET) and others have become merely the starting point for systems that permit the physician to visualize and quantify the extent of disease and to plan therapeutic interventions.

Another trend emerging from the availability of these new computational resources is the increased availability of three dimensional imagery. The interpretation of diagnostic imaging results has been a two dimensional task: projection radiography compresses the image content from an inarguably 3-D patient onto a 2-D piece of film. Even imaging modalities which are inherently 3-D have traditionally been interpreted as a stack of 2-D slices -- the root tomo figures prominently in the names of these remarkable instruments. But if planning systems are to be genuinely useful to the surgeon, they must accurately predict the spatial relationships of internal and external landmarks, diseased and damaged tissues and essential structures within the brain. We can now depict these relationships from arbitrarily chosen viewpoints in a realistic fashion - all in a matter of seconds using commercial, off-the-shelf hardware.

We describe in this paper the development of Netra, a real-time system that permits the neurosurgeon to explore surgical approaches that are extremely difficult to visualize from the information present in the traditional axial, sagittal and coronal projections. The system incorporates extensive segmentation, image integration and registration capabilities, as well as frame-based stereotactic planning algorithms. All imaging, image pre-processing and planning phases are compressed into a clinically useful time frame - the two or three hours just prior to surgery. Our goal is to demonstrate that computer-based visualization and planning systems can positively impact both cost-of-care and patient outcome.

1.1 What is Real-Time?

The practical constraints of surgery define the notion of "real-time." With current pressures on the health care delivery system in the United States, there are time and cost constraints that effectively bound access to radiological services and severely limit the time available for surgical planning. For instance, multiple MR image acquisitions using T1 and T2-weighted pulse sequences might permit the use of powerful multi-spectral image segmentation and classification algorithms. In practice, however, there is rarely time to do more than the minimum imaging required for stereotactic registration - the patient's diagnosis has already been

\[\text{Netra} \text{ is loosely translated as "the all-seeing eye" from Sanskrit}\]
established and an operating room (with a staff of five or six residents, nurses, orderlies, etc) may be standing idle awaiting the completion of imaging and planning. Nor is the patient's insurance company likely to cheerfully bear the cost of both MR and CT on the same morning unless a convincing argument can be established with respect to the necessity of this course.

Similarly, access to the principal neurosurgeon responsible for the case on the morning of surgery is similarly limited: the surgeon will be responsible for several other procedures in various stages of completion, and his ability to commit time to computerized surgical planning will be severely compromised. The planning system must rapidly and reliably produce the planning information needed to perform the minimally invasive procedure.

To place the notion of “real-time” in perspective, a typical MR-guided depth electrode implantation procedure begins in the operating room with stereotactic frame placement at about 0730. By 0800, the patient is in the MR suite, for about 30 minutes of scanning. By 0900, the patient has been transported to the neuroangio suite and is prepared for contrast angiography to delineate the arteries and veins in the brain. If no complications are encountered, images will be available by approximately 1000.

In the meantime, the MR images have been digitally transferred from the MR scanner to the planning computer via ethernet. The images have been stereotactically registered, and image segmentation and classification has been accomplished. With the availability of the subtraction angiograms, the procedure awaits the arrival of the neurosurgeon who will perform the surgery and the neurologist responsible for overall care of the epileptic patient.

By 1030 all images are available, and the patient has been taken to a holding area near the operating room. Usually the planning for this type of procedure takes no more than 30-45 minutes, with the surgeon and the neurologist co-operating in the selection of structures in the brain that are likely candidates for electrode placement. Once planning is complete, the patient is prepared for the placement surgery, which ordinarily will take no more than two or three hours.

In the context of a large academic neurosurgery department, minimally invasive procedures must be closely scheduled, with intense communication and co-operation between neurosurgeon, neuroradiologist, neurologist and surgical staff. The planning system must produce safe, reliable trajectories with a high degree of confidence, all in at most two or three hours. “Real-time” is determined not so much by technological challenges as by the scheduling and synchronization issues in these complicated cases.

1.2 Neurosurgery as a Testbed

Despite the challenges noted in the previous section, neurosurgery has become an early adopter of computer-based techniques for a variety of historical and practical reasons:

- Planning systems are dependent on the availability of digital diagnostic imaging modalities for the base image data. Many of these imaging systems have been initially developed and used in a neurologic context. Neurosurgeons are intensive users of medical imaging in their routine practice.

- The structures in the brain remain relatively fixed with respect to landmarks in the head. Problems associated with cardiac and respiratory motion of significant neurologic structures are minimized.

- The consequences of surgical insult to the brain are substantial. Minimally invasive procedures are attractive because they result in reduced frequency of complications, shorter intensive care stays and improved patient outcome when compared to traditional open surgical techniques.

- Stereotactic surgical instruments and imaging adapters for neurosurgical applications are well developed and commercially available.

As a result, neurosurgery is attractive as an initial application domain for these techniques. Early investigations indicate that many of the techniques described in this paper are easily extendable to orthopedic and cranio-facial surgery, but we restrict this report to our experience in neurological surgery.

1.3 Other Planning Systems

*Netra* builds upon extensive research and a fast growing literature in neurological surgery planning. In fact, the earliest concept of defining structures in the brain in a 3-D coordinate system dates from R.H. Clarke’s work early in the century. The first human stereotactic procedure, a thalamotomy, was reported in 1947 by Spiegel and Wycis. The early 1950s saw extensive development of stereotactic techniques and specialized surgical instruments at centers in France and Sweden by Talairach and Leksell, respectively.

The introduction of the EMI Scanner in 1973 began a series of revolutionary improvements in our ability to visualize normal anatomy and pathology within the brain, and subsequently, our ability to accurately plan stereotactic surgical interventions. In 1982, Gildenberg reported the use of 2D CT images for stereotactic surgical planning. Concurrent advances in the spatial
resolution available from CT and in 3D rendering algorithms enabled the construction of realistic perspective renderings from multiple CT slices. As early as 1983, for instance, Udupa demonstrated realistic 3D renderings of bony structures in the head.18

More recently, dramatic improvements in the computational and graphics capabilities of commercially available workstations have spawned increasingly sophisticated and capable surgical planning systems. Kelly, in 19889, reported his extensive experience with computer-assisted stereotactic resection of metastatic tumors using a system based on CT. Newer systems exploit the increased soft tissue contrast and more nearly isotropic voxel aspect ratio available from MR. Volumetric pulse sequences such as MP-RAGE 10 can be used to rapidly acquire a set of T1-weighted images in which the voxels are nearly isotropic. The sequence is relatively immune to spatial distortions resulting from susceptibility artifacts and is often used in our laboratory as a basis for 3D surgical planning.

There is also intensive research and commercial interest in intra-operative navigation systems that do not require use of a stereotactic frame.(2,3,6,15,20) Most of these systems include a surgical planning component, but do not use the fiducial indicators common to the frame-based systems. Frameless systems appear extremely promising for some procedures, and we are currently adding support for some frameless techniques to Netra.

2. System Overview

A principal goal of Netra is to act as a framework for a continuing program of research in techniques for pre-surgical planning, intra-operative navigation, image measurement and analysis and physician-computer interface design. Written entirely in C++, the system is inherently modular and can easily be configured for specific research and clinical support projects. A “frozen” version of the system is used to plan stereotactic biopsies, depth electrode placements and other minimally invasive procedures on a routine basis. Figure 1. depicts the logical and physical flow of data in the system.

Netra consists of several principal software components, including:

- A simple, distributed database for medical images, segmented data, regions of interest and other resources related to the planning system. The database server will be DICOM compliant and will generate queries against the recently installed Radiology PACS system.
- An intelligent image loader. The loader recognizes a number of common image file formats, including GIF, TIFF, PPM and Sun Rasterfiles. It also can load ACR/NEMA v2.0 data files, including most vendor’s private blocks, and also recognizes some purely proprietary formats such as those for GE 9800 and Picker CT scanners, Luminys film digitizer files, etc.
- Support for Asynchronous Transfer Mode (ATM) links in the local area and wide area environments over 100 Mbit/sec multi-mode fiber connections. Many current imaging devices do not have ATM capability, so planning workstations and file servers are also connected through nominal 10 Mbit/sec Ethernet connections.
- Complete stereotactic registration software for the Leksell system, including MR, CT and projection angiographic localizers.
- Software for semi-automatic 3D MRI image segmentation. This module uses a knowledge-based approach to classify each voxel in the volume into one or more tissue classes.
- A planner module that permits surgical trajectory selection from any stereotactically registered modality.

2.1 Hardware Configuration

The principal planning computers are essentially commercial high performance UNIX workstations connected via a private ATM network using multi-mode fiber. To permit planning in real-time, these machines must be robust. A typical configuration:

- Processor: Hewlett-Packard HP9000/735.
- 2 Gbytes of fast-wide SCSI disk with NFS access to a 20Gbyte file server.
- , At least 196 Mbytes of memory.
- 48 CRXZ dual 24bit frame buffers
- ATM and Ethernet network interfaces.
- Custom user interface devices.
Planning computers are located in the Neurosurgical Visualization Lab, in a Neuroradiology reading area, and in the Gamma Knife radiosurgical facility. An 100Mbit/sec multi-mode fiber connection also links NVL to the principal neurosurgical operating room suite via ATM switches. Simultaneous data, JPEG-compressed video and high quality stereo audio can be transmitted.

Figure 1. Logical and physical data flow during neurosurgical planning.
3. Image Acquisition and Database Operations

3.1 Medical Image Acquisition

Acquiring the original source images from the digital imaging modalities in clinically useful times remains a challenge. Many MR and CT scanners still in active and useful service are based on computers that are not well suited to networking. Vendor software has traditionally used proprietary file formats, undocumented file compression techniques, and little ability to co-exist in mixed vendor environments. More recent acquisitions are usually based on commercially available processors, and can often be networked more easily - although transfers are likely to be limited to Ethernet speeds.

In our facility, the principal MR scanner used for surgical planning is running third party TCP/IP software running over an Ethernet transport layer. The imaging required for typical surgical planning may easily exceed 20 Mbytes, and transferring these data sets to our planning workstations over a heavily loaded hospital 10BaseT network can be a major problem.

Currently, the source images are stored on a workstation that functions as a file server, with several Gbytes of fast SCSI storage and an attached 20Gbyte optical jukebox. This storage is adequate for clinical operations but requires frequent migration of image files to backup 4mm DAT tapedrives when the surgical case load is high.

3.2 Database Server Operation

We have implemented a database server that is implemented using Internet stream sockets. This strategy provides for a reliable and persistent two way communication link between one or more planning processes and the database server/loader process. The server supports queries from authorized machines against the database, and responds to requests for patient lists, study and sequence data, and provides a pointer into a distributed file system that holds the actual image data in a format native to the original source. The server also decodes the images from their native format, populates header block variables associated with each image and delivers them to the requesting process in a simple internal format.

The database also registers “resources” related to subsequent processing information and associates this data with the base image data. Segmented volume data, regions of interest, volume calculations and other image related data can be stored and recalled from the database on a per-user basis.

4. Pre-Processing

Once the basic image data is available to the planning application, the images must be registered into the stereotactic coordinate system established by the Leksell frame. Registration algorithms differ according to the imaging modality and consist of both 2D and 3D operations.

4.1 Multi-modality Registration

CT scans and subtraction angiography images are currently registered using 2D algorithms and standard Leksell localizers. Angiographic images are generally produced using right and left internal carotid contrast injections in a bi-plane angiographic suite equipped with Puck film changers. Late phase images are obtained to delineate the venous structures. Photographic subtraction of the cut film images are then digitized using a high resolution film scanner.

We have evaluated the use of digital subtraction angiography for vessel delineation. This would potentially reduce the total contrast load required and would permit direct import of the DSA images. We have found that image intensifier spatial nonlinearities - usually in the form of barrel or pin cushion distortion, can significantly compromise the accuracy of the stereotactic registration, even if this distortion is not apparent when the images are viewed for diagnostic purposes.

The stereotactic localizer box, with four opaque markers on each face, is attached to the frame and remains visible even in the subtracted images. The projection of the markers onto the films is analyzed using a method adapted from Siddon. Our physicians prefer to specify target and entry points of the surgical trajectory from MR images, and then evaluate the projection of the path on the registered angio subtractions.

Axial and coronal MR images may be registered using well known three fiducial techniques similar to those used in CT scanning. We have developed a new, more accurate technique that takes advantage of the three dimensional nature of the MR volume and performs a 3D registration.

A 3D transformation between the MR image volumes and the Leksell coordinate system is established by localizing the vertices of the left and right side fiducial boxes as they appear in the 3D rendered image. The initial points indicated by the user are
projected along a line orthogonal to the screen and into the image volume until an intersection with the segmented fiducial marker is reached. Since the Z depth is known, we can recover the screen co-ordinates of the individual vertices.

These estimates are refined automatically by searching for neighboring bright corners in the image volume. These eight points are mapped by the transform into their known positions in the Leksell coordinate system. The 3D transformation allows for stereotactic localization to be achieved on any oblique slice through the volume as well as on any object surfaces defined by segmentation procedures.

Following computation of the affine transformation between the image co-ordinates and Leksell space surgical co-ordinates, the location of the fiducial marks can be rendered into the 3D volume image. The modeled marks should exactly overlay the actual fiducial locations, providing visual verification to the physician that the registration has been accomplished correctly.

### 4.2 Image Segmentation

Image segmentation is implemented in the system in order to accurately define 3D objects of interest. This is necessary to effectively visualize the surfaces of the head, brain, vasculature or other objects using standard volume visualization techniques. Presentation of the image data as a 3D perspective view is required to facilitate rapid understanding of complex structures and spatial relationships during the surgical planning process.

The segmentation process requires the use of a priori knowledge of imaging modality characteristics and head anatomy in order to overcome object boundary ambiguities in the image data. The a priori anatomical knowledge is provided in the form of a polygonal surface model of brain surface as well as four voxel-based models for typical brain composition, brain shape, gray/white matter distribution, and ventricle shape. Use of the Talairach proportional grid system provides approximate registration of the models with a given subject. Once registered, the models allow image preprocessing using two techniques: presumptive class histograms and retrospective model-based gradient removal.

Selective histograms can be made by restricting the voxels used in histogram construction to those with high probability of being in a particular class according to our models. If the models are adequate, these presumptive class histograms will greatly emphasize individual class peaks compared to an indiscriminate histogram of the entire image.

Presumptive class histograms are used to initially classify the image of the head. Presumptive CSF, gray matter, and white matter histograms are constructed, and the intensity distributions for each class are obtained by modelling the histograms as a sum of gaussians.

Operations using image intensities often assume the substances have spatially invariant intensities; this assumption may be incorrect in many cases due to imaging artifacts. Particularly troubling in magnetic resonance imaging is the spatial inhomogeneities introduced by the radio frequency coil. Previous work has used phantoms and filtering (e.g. high-pass or divide by low-pass) techniques to remove the slowly varying intensity gradients. We have implemented a retrospective model-based method to estimate the imager sensitivity using an order two, three-dimensional polynomial.

The problem of spatial intensity variations in MR images has been studied by a number of researchers since the mid-1980’s, and two assumptions are commonly made:

1. The observed intensities $I_{\text{obs}}(\hat{p}) = I(\hat{p})$ are a product of the RF signals generated by the relaxing protons, $I_{\text{true}}(\hat{p})$, and the sensitivity profile $S(\hat{p})$ of the RF coil used to detect the generated signals:

$$I_{\text{obs}}(\hat{p}) = S(\hat{p}) \cdot I_{\text{true}}(\hat{p})$$

2. The sensitivity profile $S(\hat{p})$ varies slowly relative to the higher spatial frequency components of anatomical object borders.

Removal of the gradient, then, is simply a matter of determining $I_{\text{true}}(\hat{p})$ from $I_{\text{obs}}(\hat{p})$. We use the a priori knowledge provided by the registered models and treat the imaged brain as our phantom. Rather than estimating $S(\hat{p})$ by using a phantom, lowpassed image, or user-defined points, we estimate $I_{\text{true}}(\hat{p})$ for likely brain voxels in a subsampled fashion, calculate an averaged sensitivity profile at each point in the subsampled grid, and then fit a polynomial to the averaged sensitivities. Since the voxel-based models are used to select the data points, user interaction is reduced to the initial registration of the image to Talairach space.

We fit a second-order three-dimensional polynomial $\gamma(\hat{q})$ to $S(\hat{q})$ using singular value decomposition to find the best least-squares approximation. Since our fit polynomial $\gamma(\hat{q})$ describes a continuous surface, we can compute the corrected intensity at any voxel $\hat{p}$ of the image as $I_{\text{true}}(\hat{p}) = I(\hat{p}) / \gamma(\kappa \cdot \hat{p})$ where $\kappa$ is used to map voxels to subvolumes.
4.2.1 Active Surfaces

Analysis of the presumptive class histograms allows the image to be automatically broken into foreground and background regions corresponding to head/brain and air/bone/cerebrospinal fluid respectively. The head and brain are typically spatially connected within the foreground region and must be separated by selectively breaking these connections. This task is performed by the application of an active surface template, which elastically models the outer surface of the brain and spinal cord.

The surface model is based on the physically-based deformable models described in [17]. The active surface is an energy minimizing spline which is characterized by a set of intrinsic and extrinsic constraints. The intrinsic constraints govern the material properties of the surface, while the extrinsic constraints link the surface to the image data through "forces" which deform it. The equation of motion is given by

\[
\begin{align*}
\mathbf{w}_2 \left[ \frac{\partial^4 \mathbf{v}}{\partial x^4} + 2 \frac{\partial^4 \mathbf{v}}{\partial x^2 \partial y^2} + \frac{\partial^4 \mathbf{v}}{\partial y^4} \right] \\
-\mathbf{w}_1 \left[ \frac{\partial^2 \mathbf{v}}{\partial x^2} + \frac{\partial^2 \mathbf{v}}{\partial y^2} \right] = f(\mathbf{v})
\end{align*}
\]

where \( w_1 \) and \( w_2 \) determine the elasticity and stiffness of the surface respectively. The surface is linked to the image data by the extrinsic forces defined as

\[
f(\mathbf{v}) = \frac{\delta P}{\delta \mathbf{v}}
\]

Our image potential function is based on a vector distance transform (VDT) of the foreground region which has the effect of reducing unwanted local minima which typically occur in gray-level based functions. The gradient of the VDT always points toward the nearest foreground boundary and therefore tends to restore surfaces from the interior of the foreground.

In order to solve the system given by Equation (5), the domain is discretized with equal spacing \( h \) between the grid points and the derivatives of the Laplacian and biharmonic parts of Equation (5) are approximated by finite differences. The system may be written in matrix form as

\[
\mathbf{A} \mathbf{v} = \mathbf{F}.
\]

We have used the simple Gauss-Seidel iterative solution method, lagging the nonlinear terms in \( \mathbf{F} \). The iteration process is stopped when the average distance of the surface from the background falls below a suitably low value.

The major substructures of the brain are modeled independently by separate active surfaces. These subsurfaces are connected by enforcing mutual boundary conditions alone adjoining edges. This allows for the natural formation of folds or articulations between substructures. The resulting surface model is comprised of approximately 80 thousand nodes and requires 3-4 minutes of computation on an HP Apollo 700 series workstation. The equilibrium configuration of the active surface model is then used to separate the brain from the foreground region of the image.

4.2.2 Prioritized Growth

While model-based knowledge is essential to making segmentation more automatic, current limitations in computational power and memory prevent any fully automated model-based segmentation system from correctly identifying the variety of images encountered in clinical practice. Therefore, the segmentation process, while nearly automatic for "normal" images, should allow user labelling to override model-based labelling when pathology is encountered.

After determination of the brain boundaries using the active surface method, all foreground and background areas are broken into regions using watershed analysis. Foreground areas are broken into regions using watershed analysis on image gradients while the background areas are split by watershed analysis of distances from the foreground.

After splitting of an image into numerous regions, one would like a data structure which compactly represents the spatial relationships of the regions and allows efficient analysis of the represented regions. The region adjacency graph (RAG) is a popular data structure satisfying these requirements. In a RAG, regions are represented as nodes (or vertices or points) and the adjacency of two regions is represented by an edge (or arc) between the two corresponding nodes.

Two RAGs are created - one for the background area and one for the foreground area. Adjacencies between foreground and background regions are not included in either graph. The next step merges the different regions into coherent 3D objects. Within
Talairach space, universally present head structures like the brain and ventricular system are easily localized. For these structures, anatomical models can automatically provide the initial labelling of “core” nodes located in areas almost always occupied by a particular structure. However, in clinical settings, a number of unexpected objects (e.g. tumors, cysts) can occur in a variety of locations. In order to segment these objects, the voxel-based models must be supplemented by user labelling. This requirement is met by our two stage approach of (1) initial labelling followed by (2) prioritized growth of structures.

The nodes are initially labelled by either the registered anatomic models or a knowledgeable user. From these initially labelled core nodes, we slowly grow our target structures towards the more ambiguous border areas using a prioritized growth algorithm. The prioritized growth algorithm works sequentially through an ordered list of unlabeled nodes. The priorities (which determine ordering) are computed, in part, by the strength of a region’s connections to previously labelled neighboring regions. Labelling halts when all nodes have been visited; therefore, the algorithm is guaranteed to terminate after visiting each unlabeled node once.

Cadaver studies were performed to validate the segmentation results. The brain was segmented from MR images of whole cadaver heads and compared with images of the brain after dissection. Segmented volumes compare with actual volume measurements within 5%.

5. Surgical Trajectory Planning

Once the image registration and segmentation have been performed, the data is ready for surgical trajectory planning. A multiple window GUI permits the physician to input and validate surgical co-ordinates, 3D trajectories and frame settings. Using the 3D MRI volume data set, the operator can select target neurological structures from the canonical sagittal, coronal and axial views. Moving the slice selector in any plane selects corresponding resampled images in the other views.

Once the target has been selected, the 3D volume rendered view is used for burr hole selection. The volume rendering requires approximately 5 seconds on the machines described in Section 2.1, and pre-selected views can be stored and recalled. Using transparency, important structures within the brain can be effectively avoided. MR angiography data, including both arteries and veins, SPECT images, and even EMG signals can be registered into the co-ordinate system and rendered for surgical orientation.

Figure 3. depicts a typical 3D view. The Leksell frame and the upright posts that secure the patient’s skull are depicted graphically to warn the physician of trajectories that may be physically blocked intra-operatively. The Micro-stereotactic arc is iconically depicted for physician orientation in the OR.

Less visible in this figure are the depiction of the cortical vein structure derived from a MRI volume subtraction technique. While beyond the scope of this paper, the technique permits accurate delineation of these veins that are often troublesome to the surgeon performing minimally invasive techniques.

Figure 4. depicts the surgical trajectory, which has now been displayed with the same perspective distortion of the original image. Note that intersection of the trajectory does not necessarily indicate that the path will intersect the vessel: the structure may be much deeper in the image. With two views, we must depend upon the physician’s knowledge of the neurovascular structures to ascertain that the proposed trajectory is really safe.
Neurosurgery occurs in three dimensions and deals with complex three-dimensional structures. The neurosurgeon must be able to visualize these structures and understand how a proposed surgical intervention will impact different regions of the brain. To facilitate this task, we have implemented a computer-based neurosurgical planning system which incorporates a 3D interface based on the surgeon’s everyday skills for manipulating real-world tools with two hands.

We use a commercially available six-degree-of-freedom tracking system called the FASTRAK, manufactured by Polhemus Navigation Systems, Inc. to monitor the location of the props. Each prop is instrumented with a small magnetic receiver which generates a signal in response to pulsed magnetic waves sent out by a nearby magnetic field transmitting box. The FASTRAK system processes the signals generated by the receiver to determine the location (x, y, z) and orientation (yaw, pitch, roll) of each receiver relative to the transmitting box, and returns this information to the host computer via an RS232 serial port connection.

6.0 User Interface Considerations

Neurosurgery occurs in three dimensions and deals with complex three-dimensional structures. The neurosurgeon must be able to visualize these structures and understand how a proposed surgical intervention will impact different regions of the brain. To facilitate this task, we have implemented a computer-based neurosurgical planning system which incorporates a 3D interface based on the surgeon’s everyday skills for manipulating real-world tools with two hands.

We use a commercially available six-degree-of-freedom tracking system called the FASTRAK, manufactured by Polhemus Navigation Systems, Inc. to monitor the location of the props. Each prop is instrumented with a small magnetic receiver which generates a signal in response to pulsed magnetic waves sent out by a nearby magnetic field transmitting box. The FASTRAK system processes the signals generated by the receiver to determine the location (x, y, z) and orientation (yaw, pitch, roll) of each receiver relative to the transmitting box, and returns this information to the host computer via an RS232 serial port connection.
When the user picks up the doll’s head and rotates it, the computer-generated 3D image of the individual patient’s head seen on the screen is automatically updated to match the orientation of the doll’s head. The surgeon can also control the image zoom factor by moving the doll’s head towards or away from his or her body. Since moving the head left-right or up-down is typically not useful, we have found it helpful to constrain the (x, y) position of the polygonal brain to the center of the screen. This simplifies the task and surgeons find it natural.

The on-screen image or “virtual head” is actually updated approximately 15 times per second, so from the user’s perspective, the motion of the doll’s head and the motion of the virtual head appear to be tightly coupled. Ideally, as the surgeon rotates the doll’s head, he or she would like to see a detailed volume-rendered image of the individual patient’s head (Figure 5).

The two-handed interface techniques permit fast, intuitive interaction between the neurosurgeon and complex planning software. These techniques form a basis for surgical planning software that is well accepted by the surgical community.

![Figure 5](https://example.com/figure5.png)

Figure 5 Physician using two-handed interface tools.

**7.0 Conclusion**

We have implemented a 3D computerized stereotactic planning system which integrates many recent developments in computer workstation, graphics and networking technologies as well as our own research in knowledge-based image segmentation and 3D user interface design. Our priorities for system design have emphasized real-time performance, both in terms of rapid image acquisition and user interface interactivity.

By presenting the neurosurgeon with a 3D oriented system which integrates all the available image information and by providing intuitive means for its manipulation, we have described a fast and accurate brain segmentation method that provides a basis for 3D visualization and surgical planning. The method permits 3D models of patient anatomy to be constructed within minutes of image acquisition, making 3D image acquisition and visualization clinically practical.

**References**

Figure 6 Simulation of planned surgical trajectory.

Figure 7 Actual trajectory executed in the operating room.