A Model-Based Optimal Planning and Execution System with Active Sensing and Passive Manipulation for Augmentation of Human Precision in Computer-Integrated Surgery

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Abstract

Researchers at IBM and NYU Medical Center have recently begun development of a model-based system for optimal planning and augmented execution of precise osteotomies to correct craniofacial malformations. In these procedures, the facial bones are cut into several fragments and relocated to give the patient a more normal facial appearance. There is a significant synergy between better presurgical planning methods and the ability to execute the plans precisely and efficiently. The planning component of our system will transform CT images into a 3D geometric model of the patient's skull and assists the surgeon in planning an optimal procedure based on an analysis of the patient's anatomy compared to a database of normal anatomy. The surgical component will use realtime sensing to register the model-based surgical plan with the reality in the operating room. It will employ a variety of man-machine interface modalities (graphics, synthesized speech, etc.) together with passive manipulation aids to assist the surgeon in precise execution of his plan. This paper describes the overall system architecture, the proposed surgical procedure, implementation status, and some early experiments that we have performed.

Introduction and Background

Recent advances in medical imaging technology (CT, MRI, PET, etc.), coupled with advances in computer-based image processing and modelling capabilities have given physicians an unprecedented ability to model and visualize anatomical structures in live patients, and to use this information quantitatively in diagnosis and treatment planning.

The precision of image-based pre-surgical planning often greatly exceeds the precision of surgical execution. Typically, precise surgical execution has been limited to procedures (such as brain biopsies) for which a suitable stereotactic frame is available. The inconvenience and restricted applicability of these devices has led many researchers to explore the use of robotic devices to augment a surgeon's ability to perform geometrically precise tasks planned from computed tomography (CT) or other image data (e.g., [1], [2], [3], [4], [5]). The ultimate goal of this research is a partnership between a man (the surgeon) and machines (computers and robots) that seeks to exploit the capabilities of both to do a task *better* than either can do alone. Machines are very precise and untiring and can be equipped with any number of sensory feedback devices. Numerically controlled, robots can move a surgical instrument through an exactly defined trajectory with precisely controlled forces. On the other hand, the surgeon is very dexterous. He is also quite strong, fast, and is highly trained to exploit a variety of tactile, visual, and other cues. "Judgementally" controlled, he understands what is going on in the surgery and uses his dexterity, senses, and experience to execute the procedure. By nature, he wants to be in control of everything that goes on. However, he *must* rely on the machines to provide precision. How can he trust them not to harm the patient?

Augmentation with Simple Passive Aids

The most obvious way to prevent a robotic device from making an undesired motion is to make it incapable of moving of its own accord. Motor-less manipulators have been implemented, in which joint encoders are used to provide feedback to the surgeon on where his instruments are relative to his image-based surgical plan (e.g., [4], [6]). One important limitation of this approach is that it is often very difficult for a person to align a tool accurately in six degrees of freedom with only positional feedback. Passive manipulators, permitting free motion until locked, have been implemented for limb positioning, tissue retraction, instrument holding, and other applications in which accuracy is not important [7] [8]. In another case, Davies [9] implemented a passive manipulation aid with restricted degrees-of-freedom for prostate surgery, which was used clinically, after prototyping the necessary motions on an active robot.

Within the planning system, initial programs for image processing, model construction, and feature extractions have been completed. Substantial progress has been made on the interactive graphics and surgical simulation components. Independent work is underway in to construct the anatomical data base. Work on a full-blown surgical plan optimizer using point, line, and surface features is still in very early stages. However, an earlier plan optimizer based on point features is already in clinical use ([21], [15]) and could be used as an alternative.

Within the surgical system, we have developed interfaces to the Optotrak, an adequate pointing system, beacon mounting methods and means of tracking and displaying bone fragment motions relative to each other. A variety of PC-based graphics, voice, and tonal cues are available for feedback to the human, and we are in the process of debugging the service routines needed to provide a connection to the RS6000 for more sophisticated graphics and other online processing.

We have gone through several generations of the passive manipulator. The first version, shown in Figure 5, used a modified SCARA structure with a counterbalanced Z-arm for coarse (and fine Z) positioning, with a 5 DOF distal fine positioner. It had electric particle brakes with zero-backlash gearboxes for the four proximal joints and manual clamps distally. Our experience with this version was mixed. The distal fine positioning mechanism worked very well. However, the proximal (coarse) structure proved too compliant and the Z-arm had too much inertia to be effective as a fine positioning mechanism. Consequently, we modified the structure to provide a fine "Z" stage counterbalanced by a constant force spring and replaced the coarse positioning joints with a simpler and more rigid cartesian structure (Figure 6).

Several different versions of the end-of-arm tooling were developed. We eventually settled on an Elmed "Retract-Robot" (tm) surgical instrument holder [8]. Our initial implementation placed a Lord Corp. 6DOF force/torque sensor between the goniometer cradles and the instrument holder. In practice, we found that this arrangement did not leave sufficient clear working space, so we removed the force sensor. Subsequent versions will have a larger "standoff" distance, thus leaving more room for the force sensor and other tooling.

Experiments and Early Experience with the Surgical System

Pointing System

Our prototype pointer is shown in Figure 7, and has four LED beacons placed approximately at the corners of a 75mm by 150mm rectangle. The pointer tip is a needle approximately 90mm from the plane of the rectangle. We calibrate the pointer by placing it in a

Next, one or more "views" of the patient are obtained to measure the *relative* position of the beacons, which are all rigidly affixed to the skull. At this point, it is necessary to "register" the beacons to the CT coordinate system used to plan the procedure. The most straightforward technique for doing this is to use a calibrated pointer to identify known landmarks while simultaneously recording the positions of the beacons mounted to the patient's skull. These landmarks can either be anatomical features or fiducial markers rigidly affixed to the patient when he is CT'ed and replaced in the same place at the time of surgery.[3] Once this has been done, it is possible to compute the coordinates T_{ci} of each bone fragment B_i relative to the camera. The coordinates of fragment B_j relative to fragment B_i are, of course, $T_{ij} = T_{ci}^{-1} \bullet T_{cj}$

Next, the surgeon cuts free a bone fragment B_j. He now needs to manipulate this fragment so that it is in a desired pose ${}_{des}T_{ij}$ relative to another bone fragment B_i.[4] To do this, he places the center-of-rotation of the passive manipulator's goniometer cradle stages over the desired center-of-rotation of the bone fragment. He uses standard surgical tools to grasp the bone fragment firmly and uses the adjustable end-of-arm tooling clamp to grasp the instrument. At this point the bone fragment is rigidly affixed to the manipulation aid. He now uses the passive manipulation aid to realign the bone fragment so that $T_{cj} = T_{ci} \bullet {}_{des}T_{ij}$. In a typical alignment strategy, this will be done by first unlocking all "fine motion" degrees of freedom and manipulating the fragment into its approximate desired position, with the surgeon relying on his own tactile feedback and the force sensor information to verify that there is no undesired obstruction. Then, each degree-of-freedom will be successively brought into alignment and locked. Once the B_j has been aligned, standard surgical screw and plate methods are then used to affix it to B_i. Bone grafts are used as necessary to fill in any gaps between fragments.

This process is repeated until all bone fragments have been repositioned. Surgery then procedes normally.

Implementation Status

As of May 1991, many components of the total system architecture have been implemented, at least in preliminary form. However, system integration is just beginning. In particular, we have yet to integrate the planning system with the surgical system and more integration has still to be done integrating sub-components as well.

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scheme is that the beacon carrier could be used to as a handle to assist in surgical manipulation. The disadvantage is that it is bulky.

[3] One possibility is to use a "mouth guard" customized to the patient's teeth to hold calibration phantoms and (possibly) LED beacons to eliminate the use of the pointer.

[4] For simplicity, this discussion will assume that B_i is rigidly affixed to the skull base, which is held fixed relative to the manipulator base during this part of the procedure. There are a number of options for relaxing this constraint.

such aids is that the surgeon provides all the motive force. Generally, the manipulation aid should interfere as little as possible with the surgeon's tactile "feel" for what is happening to the patient, while preserving the desired alignment once it is achieved. Six DOF manipulation aids with manually [8] and semi-automatically [7] actuated brakes have been developed for tissue retraction, instrument placement, and similar applications. One serious drawback of these systems is that they provide little assistance in actually achieving the desired alignment. Even without the additional inertia of a mechanical linkage, most people find it extremely difficult to achieve an accurate six degree-of-freedom alignment.

Our approach is to develop manipulation aids with computer controlled (or manually actuated) brakes to provide selective locking of *orthogonally decoupled* degrees-of-freedom resolved in a tool frame centered at a point reasonably far removed from the mechanism. This permits implementation of a variety of manipulation strategies in which the surgeon only needs to work on aligning a few (often, one) degrees-of-freedom at a time.

The basic structure of our present implementation is illustrated in Figure 4. There are three basic components. A 6 DOF *fine positioning system* consists of three counterbalanced linear stages carrying a conventional k_z axis and crossed goniometer cradle k_x and k_y axes with a rotation center about 150 mm from the mechanism. A standard 6 DOF surgical *adjustable tooling clamp* [8] permits a surgical instrument to be grapsed at any desired position relative to the fine positioning stage. A three degree-of-freedom *coarse positioning system* is used to position the fine positioner's work volume at any desired location relative to the patient. One advantage of the coarse-fine structure is that it permits relatively large work volumes while limiting the inertia that the surgeon must cope with. The modularity is similarly very useful for experimentation.

Surgical Procedure

It is anticipated that surgery will procede normally up to the point where the surgeon is ready to perform the first planned osteotomy. At this point, the surgeon will affix at least three (and, usually, four or five) LED beacons to each (future) bone fragment. In our present planned implementation, the surgeon will affix each beacon by inserting a standard 1.5 mm surgical "K-wire" into the patient's bones, trimming the end at a convenient length, and then fitting a beacon carrier over the "stump." Depending on the location, the K-wires may be inserted percutaneously or beneath a skin flap but in a position where the beacons can be exposed to the camera. The beacon carriers are constructed so that the LED center is mounted coaxially with the K-wires and always "bottom out" against the end of the K-wires. This means that beacons may be removed and replaced for convenience during surgery, once all the initial locations have been measured.[2]

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[2] One alternative implementation might mount several beacons on a single rigid carrier that would be secured to the bones by standard orthopaedic bone screws or similar means. The advantage of this

The Sensing Subsystem provides information needed to register the reality on the operating table with the models from the surgical plan, for tracking motion of bone fragments, surgical instruments, etc. The principal geometric sensor in the present system is an Optotrak 3D digitizer manufactured by Northern Digital. This device uses three CCD linescan cameras to track active LED beacons. This system is fast and accurate, is much less readily confused by stray light than similar lateral-cell based devices, and (unlike electromagnetic field 6D sensors) is unaffected by metal in the operating theatre. The model in our laboratory is capable of producing 1000 3D positions/sec to an accuracy of about ±0.1 mm and of returning up to eight 6D positions with an additional overhead of about 10 ms beyond the 3D sampling time. Beacons are mounted on the patient for bone tracking (see below), on the manipulation aid for positional feedback, and on a pointer used as a designator by the surgeon. Other potential uses might include head tracking for a helmet-mounted display, tracking of additional surgical instruments, location of mouth or head-mounted CT fiducial markers on the patient, etc. A force sensor mounted on the passive manipulation aid (below) will be used to provide additional safety monitoring of manipulation forces. In the future, we anticipate incorporating a a number of additional sensing modalities, including normal computer vision, realtime radiography, and (possibly) redundant kinesthetic sensors in the manipulation aids to provide continuity when visual endpoint sensing is temporarily blocked.[1]

The Surgeon Interface uses a variety of modalities (graphics, synthesized voice, tonal cues. programmable impedence of manipulator joints, etc.) to provide online, realtime "advice" to the surgeon, based on the sensed relationship between the surgical plan and surgical execution. Eventually, we anticipate a quite sophisticated, "intelligent" system that uses its model of the task plan to automatically customize displays, select appropriate sensor tracking modes, and help interpret inputs from the surgeon. In this ultimate system, a helmet-mounted stereographic display might be used to project the surgical advice directly onto the surgeon's visual field, and the surgeon would use voice input to tell the system what he wants. Our initial plans are much more modest, and are intended to provide a framework for later growth. Initially, very simple realtime graphics and auditory cues will be provided for alignment. An online 3D model display will provide somewhat more detailed "snapshots" of bone fragment positions relative to the surgical plan. The surgeon will have a limited ability to modify the sequence interactively through standard menus, sterilizable computer input devices, and the pointing system. For instance, he could designate where, exactly, he has cut (or proposes to cut) an bone. The computer will be able to simulate this cutting action "online" and allow the surgeon to compare it with the cut proposed when the surgery was planned.

Passive Manipulation Aids are provided to assist the surgeon in precisely aligning bone fragments or in aligning his instruments relative to the patient. The defining charactistic of

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[1] Preliminary experience with redundant joint encoders on one version of the passive manipulator has been mixed. Whether the extra design complexity is justified will necessarily depend on the particular application and implementation details.

representation ([19]) of each connected set of tissue classified as "bone" is constructed by a variation of Baker's "weaving wall" algorithm. In the third step (model simplification), coplanar faces are merged to reduce the size of the model to about 1/3 of the original number of faces. A rendering of a typical reconstruction can be seen in Figure 3. Although the reconstructed models are of very high quality, they are still characterized by very large data structures (300,000 faces for a typical skull). We are presently developing approximation methods that should very significantly reduce the model size for most applications.

The anatomical feature extractor identifies anatomical features from the models. These features include standard morphometric landmarks, ridge curves, and surface patches bounded by ridge curves and geodesics between landmarks [20]. The present implementation is semi-automatic. A technician "seeds" the search by identifying points on or near ridge curves, and the computer then locates and follows the ridge curves. We anticipate implementing a more automatic procedure some time in the future.

The surgical simulator permits a surgeon to interactively specify where he wishes to cut the bones apart and to manipulate the pieces graphically. It also permits him to display the bone fragment motions computed by the plan optimizer (described below) and to modify the plan as he chooses.

The anatomical data base summarizes anatomical feature information for "normal" individuals, and is being constructed in a parallel research activity at NYU.

The surgical plan optimizer uses information from the anatomical data base to compute optimal motions of each bone fragment to most closely approximate the corresponding anatomy of a "normal" individual of the same age, race, sex, and size as the patient. This component, like the anatomical data base, is based on the work of one of the authors (Cutting), Grayson, Bookstein, and McCarthy [21].

The surgical plan produced consists of the model data, the location and sequence of the cuts, the location of key anatomical features to be used in registering the patient to the model data. and the planned optimal motion of each bone fragment.

Surgical System

The surgical system assists the surgeon in carrying out his surgical plan. "Real time" (i.e., predictable latency) computation is provided by an enhanced PC/AT DOS system with a 33 MHz Intel 80386/387 processor. This machine supports all sensor and manipulator interfaces and provides a limited online graphics capability. It is connected via a local area network to an IBM RS6000 workstation which provides online 3D graphics and higher level functions.

Earlier work at NYU led to the development of optimal planning methods based on analysis of morphometric landmarks obtained from radiographs, together with an innovative surgical technique based on inter-occlusal splints to help with surgical execution ([14], [15]). Although helpful, these techniques are far from perfect. Better planning techniques, based on CT-derived 3D models of the bone surfaces, are needed. Even with the inter-occlusal splints, surgical execution is still significantly less precise than planning. Furthermore, the procedure is very time consuming, and the use of splints (necessary for accuracy) forces compromises in the surgical plan. In one recent case, for example, optimal positioning of the cheek bone would have required the soft palate to be stretched (temporarily) more than was possible [15]. There is significant synergy between better planning methods and better means of executing the plans that are developed.

Our joint research is in relatively early stages and is aimed at an in-vitro demonstration on plastic skull models, and does not directly address clinical qualification of in-vivo surgical devices. The goal is an integrated system based on 3D models derived from CT data [16]. These models will be used both for optimal planning and interactive presurgical simulation of the planned procedure and for realtime online "advice" to the surgeon during execution.

Subsequent sections will describe the overall system architecture, the proposed surgical procedure, implementation status, and some early experiments we have performed.

System Architecture

The overall architecture is illustrated in Figure 2 and may be broken down, roughly, into a presurgical modelling and planning subsystem, and a surgical subsystem.

Presurgical Planning System

The presurgical planning system uses models derived from CT images to assist the surgeon in planning precise surgical procedures. It runs on an IBM RS6000 workstation with advanced graphics hardware. The principal components of this system are discussed below.

The medical image database and display system supports archival, retrieval, low-level processing, and 2D display of CT, MRI, and other images. The system we are using (QSH) was developed by one of the authors (Noz) and Maguire at NYU Medical Center, where it is in clinical use [17].

The anatomical model builder transforms CT images into 3D solid models of the patient's anatomy [16] [18]. The process proceeds in three steps. In the first step (segmentation), each voxel in the CT data set is assigned a tissue classification label, based on an adaptive thresholding technique. In the second step (model reconstruction), a winged-edge boundary

Augmentation with a Stationary Robot: Stereotactic Surgery

In cases where only a single motion axis is required during the "in contact" phase of the surgery, the robot may be used essentially as a motorized stereotactic frame (e.g., [1], [2], [10]) A passive tool guide is placed at the desired position and orientation relative to the patient; brakes are applied; and robot power is turned off before any instrument touches the patient. The surgeon provides whatever motive force is needed for the surgical instruments themselves and relies on his own tactile senses for further feedback in performing the operation. This approach ameliorates, but does not entirely eliminate, the safety issues raised by the presence of an actively powered robot in close proximity to the patient and operating room personnel. Furthermore, maintaining accurate positioning is not always easy, since many robots tend to "sag" a bit when they are turned off or to "jump" when brakes are applied. Leaving power turned on and relying on the robot's servocontroller to maintain position introduces further safety exposures. Finally, the approach is limited to cases where a passive guide suffices. The surgeon cannot execute a complex pre-computed trajectory.

Augmentation with an Active Robot: Hip Replacement Surgery

Over the past several years, researchers at IBM and the University of California at Davis developed an image-directed robotic system to augment the performance of human surgeons in precise bone machining procedures in orthopaedic surgery, with cementless total hip replacement surgery as an initial application [5]. This application inherently requires computer controlled motion of the robot's end-effector while it is in contact with the patient. Thus, considerable attention had to be paid to safety checking mechanisms [11]. In-vitro experiments conducted with this system demonstrated an order-of-magnitude improvement in implant fit and placement accuracy, compared to standard manual preparation techniques ([12], [13]). A clinical trial on dogs needing hip replacement operations is underway, and the veterinary surgeon (Dr. H. A. Paul) has founded a startup venture (ISS, Inc.) to develop and market a version for use on humans.

Augmentation with Intelligent Passive Aids: Craniofacial Osteotomies

Researchers at IBM and NYU Medical Center have recently begun research on computer-integrated methods for optimal planning and augmented execution of precise osteotomies to correct cranio-facial malformations. In these procedures, the facial bones are cut into several fragments and relocated to give the patient a more normal facial appearance. Bone grafts are applied, together with metal plates to hold the fragments in place while the patient heals. Typical pre-operative and post-operative results may be seen in Figure 1. Although these results are dramatic, considerable improvement is often still possible.

number of different orientations with the tip at the same spot, $\mathbf{c}_{tip}$ relative to the camera and measuring the pose, F_{ptr}, of the pointer rigid body relative to the camera. Then $\mathbf{c}_{tip}$ and (more importantly) the displacement $\mathbf{p}_{tip}$ relative to F_{ptr} are found by linear regression on the relation

$$F_{ptr,i} \bullet \mathbf{p}_{tip} - \mathbf{c}_{tip} = \mathbf{0}$$

for multiple pointer poses $F_{ptr,i}$. Typical residual errors when this is done for 9 data points are on the order of 0.2 mm or less. When the calibrated pointer is then used to measure known dimensions on a steel rule placed at an arbitrary spatial pose, the values returned were verified to be consistently within 0.3 mm of the ruler values. An undetermined amount of this variation may be attributed in the difficulty of placing and holding the pointer tip accurately on a flexible ruler in free space. We are presently designing a more pencil-like pointer that should be rather more convenient to use.

Positioning with Active Brakes and Endpoint Sensing

We have experimented with various strategies for exploiting the computer settable brakes on the proximal joints of our initial manipulator. Using only endpoint sensing and a simple strategy of automatically setting brakes when each successive degree-of-freedom was aligned, and then iterating once, it was possible to place the center-of-rotation of the end effector to within about 0.5 cm of a desired target within a couple minutes. Providing a variable pitch auditory signal and/or computer graphics feedback speeded this process up considerably. However, the relatively large inertia and (more importantly) the structural compliance of the initial implementation made it difficult to improve significantly on the positioning without using the fine motion joints. The whole process felt clumsier than it needed to. This lead us to replace these joints with a more rigid cartesian structure (Figure 6), which we have yet to equip with active brakes.

In order to factor out the effects of manipulator structure on active braking, we have also experimented with the much simpler "direct drive" structure shown in Figure 8. Using this structure, we found that a simple strategy of setting the brake to 10 or 20 percent of its full torque when the manipulator is within 1 mm of its target, and setting full torque when it is within its final goal (0.2 mm of target) works quite well. With practice, a person can achieve and hold a desired alignment (in two degrees-of-freedom) in an average time of 40 seconds. The broader band prevents one from overshooting the mark, and then one can hunt around for the "detent" at the target. Performance and consistency is very significantly improved by the addition of a simple tonal indication of which error band the manipulator is in. Indeed, with tonal feedback and a single narrow locking band, we are consistently able to position the manipulator to ±0.15 mm in an average time of 13 seconds.

Positioning of Simulated Bone Fragments

We have also begun experiments to verify our ability to move and place bone fragments accurately using the method outlined in "Surgical Procedure," above. We mount two sets of four beacon carriers onto eight K-wires driven into a piece of simulated bony material, which is mounted on an XYZ stage (Figure 9). We point at simple landmarks on the simulated bone, measure the beacon positions, perform an "osteotomy," reposition one fragment, "graft" it with hot melt glue, and then point at the landmarks again to measure the result. Using this process, we are able to coarse-position the manipulator to within about 0.75 mm in about 1-2 minutes. After the fragment is cut free, we can consistently fine-position it in about 2-3 minutes so that $T_{c,moved}^{-1} \bullet T_{c,fixed}$, as measured by the beacons, is within 0.3 mm and 0.5° of the desired value. We believe that these values can be improved somewhat with experience and further development. Fragment positioning, as measured from the landmarks, is within about 0.8 mm and about 1.5° of the desired value, which is rather better than a surgeon can do unaided. A significant part of this larger error is believed to be due to the particular method chosen for pointing at a few hand-drawn landmarks on somewhat flexible foam "bones," combined with the rather poor ergonomics of our existing pointer design. Beyond this, there is also a certain circularity in using the Optotrak to measure the performance of a system that it is a part of, even though it has been independently calibrated. We are planning a more careful series of experiments using better (and more) landmarks for the initial registration and a coordinate measuring machine to verify accuracy of bone fragment motion after repositioning.[5]

Simulated Surgery on a Plastic Skull

We have also performed the following "bottom line" experiment, in which a number of markers were implanted in a plastic skull model (Figure 10 and Figure 11). The skull was located by pointing to three anatomical landmark points. A fragment was then cut free using an oscillating surgical saw and the skull base was moved to a new position and orientation. The fragment was then returned to its original position relative to the skull and secured with hot melt glue. Finally, the positions of the fragment and skull base were remeasured by pointing. The measurements were repeated three times and averaged. The results were similar to those in the previous experiment. As measured by the beacons, the relative displacement of the fragment relative to the skull base, was 0.4 mm and 0.4°. With further practice, we might reasonably expect to halve these values for in-vitro tests without significant changes to the apparatus. On the other hand, actual surgery may add additional factors limiting the ultimate accuracy that is obtained.

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[5] It should be noted that errors in determining the initial positions of beacons relative to the landmarks will be largely cancelled out in tracking *relative* bone fragment motion. For example, if the fragment-to-beacon misregistration is 1 mm and 1° and the fragment is relocated by 5° and 10 mm, the additional positioning error introduced will be about 0.26 mm.

The pointer measurements showed a displacement of 0.7 mm and 0.45°. These values are comparable to values of 0.4 mm and 0.7° obtained from a similar set of measurements taken before the skull was cut apart. A significant part of the measured displacement is doubtless due to the double use of a rather clumsy pointer and only a few landmarks. A more careful experiment is being planned. Nevertheless, these initial results are quite encouraging and (again) are rather better than a surgeon can do with existing techniques.

Future work

The primary goal for the immediate future is to complete integration of the entire system. Once this is accomplished, we anticipate a period of laboratory experimentation to help us determine (a) how accurately we can, in fact, plan and execute these procedures, and (b) what forms of man-machine interface are most useful in this environment.

At the same time, we are beginning to act on some of the lessons already learned. Experience with the setups used to construct the beacon holders showed the value of micrometer-style fine adjustments after the structure has been locked. We are therefore incorporating such adjustments into the manipulation aid. At the same time, we are enlarging the effective radius of the goniometer cradles to provide room for a force sensor and additional free workspace near the patient. This will permit extremely precise alignment of the tool endpoint. Since the surgeon will have little or no direct tactile feedback while using the micrometer screws, we anticipate carefully monitoring and displaying the force sensor values during this phase of the procedure, after the surgeon first tactilely verifies a free range of motion about the target pose.

At a future date, we also anticipate motorizing the micrometer fine adjustments. This will permit us to implement a number of "shared autonomy" strategies for instrument and bone fragment positioning. It will also provide an automatic readjustment capability to compensate for small perturbations in patient positioning, as well as an extremely precise micromachining capability that could be useful in a number of surgical applications. There are, of course, a number of crucial safety issues that must be solved before an active device is used in surgery [11]. Although this research is targeted at *in-vitro* feasibility demonstrations, and does not address qualification for *in-vivo* clinical trials, it is reasonable to expect that the passive version will be used first. In any case, the modularity of both the system architecture and passive manipulation system should permit ready experimentation and a natural evolution of capabilities.

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Figures

Figure 1. Preoperative and postoperative views. showing results of manual execution.

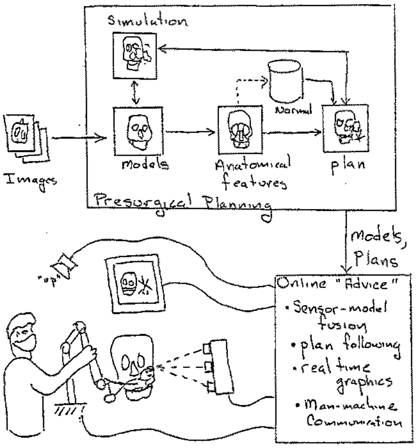

Figure 2. Overall Architecture

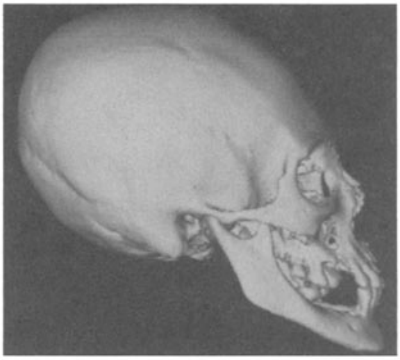

Figure 3. Rendering of typical skull model.

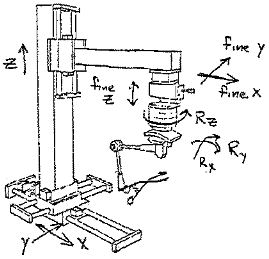

Figure 4. Passive Manipulator Structure

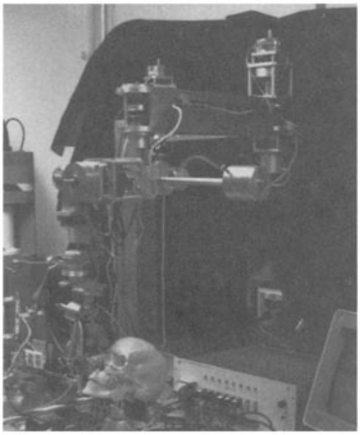

Figure 5. Initial Passive Manipulator

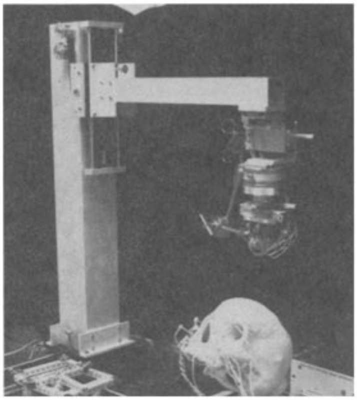

Figure 6. Refined Passive Manipulator

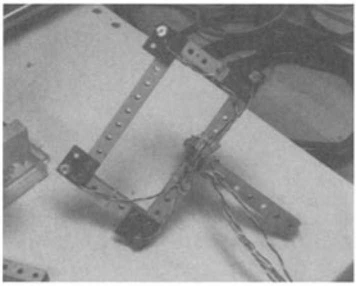

Figure 7. Pointer

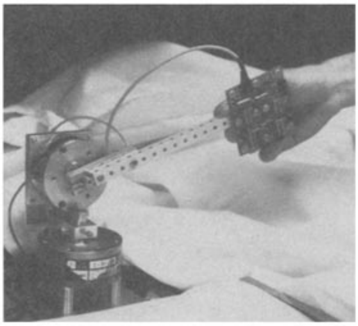

Figure 8. Computer braking experiment

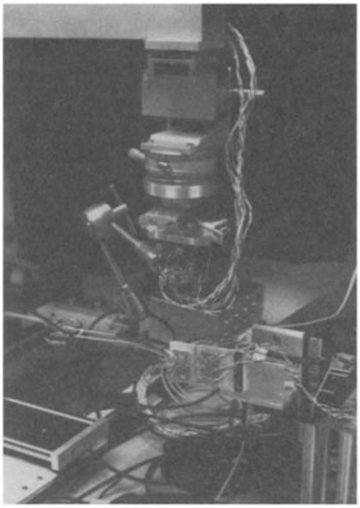

Figure 9. Simulated bone fragment experiment

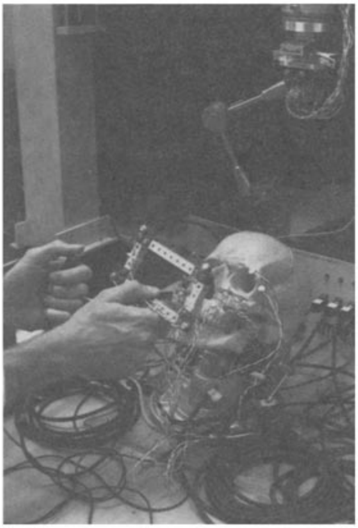

Figure 10. Locating landmarks on skull

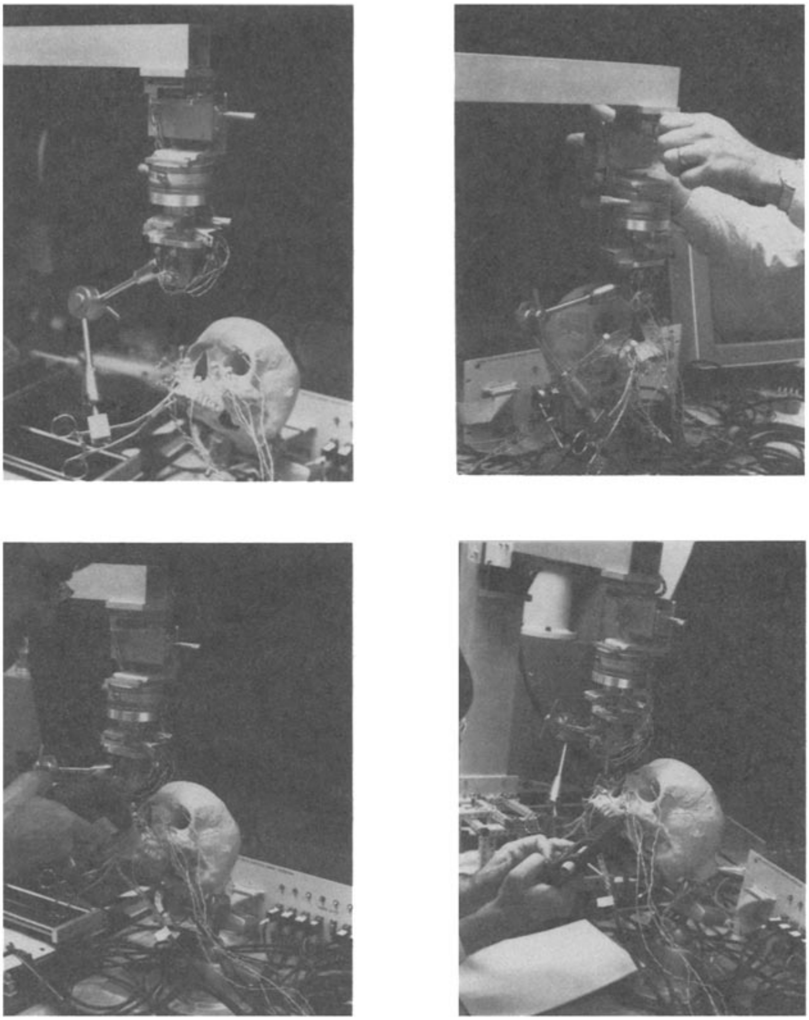

Figure 11. Simulated skull surgery