

Review

Image-guided surgery: From X-rays to Virtual Reality

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Since the discovery of X-rays, medical imaging has played a major role in the guidance of surgical procedures. While medical imaging began with simple x-ray plates to indicate the presence of foreign objects within the human body, the advent of the computer has been a major factor in the recent development of this field. Imaging techniques have grown greatly in their sophistication and can now provide the surgeon with high quality three-dimensional images depicting not only the normal anatomy and pathology, but also vascularity and function. One key factor in the advances in Image-Guided Surgery (IGS) is the ability to register images derived from the various imaging modalities amongst themselves, but also to register them to the patient. The other crucial aspect of IGS is the ability to track instruments in real time during the procedure, and to portray them as part of a realistic model of the operative volume. Stereoscopic and virtual-reality techniques can usefully enhance the visualization process. IGS nevertheless relies heavily on the assumption that the images acquired prior to surgery, and upon which the surgical guidance is based, accurately represent the morphology of the tissue during the surgical procedure. In many instances this assumption is invalid, and intra-operative real-time imaging, using interventional MRI, Ultrasound, and electrophysiological recordings are often employed to overcome this limitation. Although now in extensive clinical use, IGS is often currently perceived as an intrusion into the operating room. It must evolve towards becoming a routine surgical tool, but this will only happen if natural and intuitive human interfaces are developed for these systems.

INTRODUCTION

Throughout its evolution, the computer has played a major role in shaping image-guided surgery (IGS). The intent of this paper is both to provide an overview of the field, with a focus on its applications in neurosurgery, and to introduce the reader to the various roles of the computer in this rapidly expanding discipline. Not only is the computer indispensable in displaying and manipulating medical images, it is ubiquitous in this branch of biomedical engineering, controlling the imaging equipment, reconstructing the images from raw data, tracking instruments, modeling tissue, evaluating surgical performance, controlling robots and facilitating the human-machine interface.

We begin with a brief history of imaging techniques as they apply to the planning and guidance of surgical procedures, and follows with a discussion of the various aspects that must be considered when implementing a computerized image- guided surgery system. The scope of IGS includes both the preoperative planning of surgical procedures based on images, as well as the guidance of the procedure itself. For a full treatment of many of the contemporary problems related to image-guided surgery and therapy, the reader is referred to the recent special issue of the IEEE Transactions on Medical Imaging [1].

HISTORY

With the discovery of x-rays at the end of the 18th century, and their rapid deployment in medicine, surgeons could finally begin to see inside the human body, and be guided to the surgical site by what they could see in the image. Simple x-ray projections remained the mainstay of medical imaging until relatively recently with the advent of the computer and the development of true 3-D imaging such as Computed Tomography (CT) and Magnetic resonance Imaging (MRI). The discussion below briefly traces the evolution of medical imaging techniques over the past century.

X-ray imaging

The first medical images used for image-guided surgery were x-rays. Their first reported use as an adjunct to surgery was by John Cox, Professor of Physics at McGill University in Montreal [2], within three months of the announcement of their discovery by Roentgen in 1895. Professor Cox localized the position of a bullet in the leg of a gunshot victim. Not only was the bullet successfully removed on the basis of the radiograph, it was later used in evidence during a suit against the man who had shot the victim!

While a bullet could be easily visualized and differentiated with respect to bone and soft-tissue in a limb, by far the most important organ to visualize was the human brain. The major difficulty however, was that a radiograph of the human head shows only a projection of the bone structure of the skull, and the minimal contrast in transmitted X-ray intensity provided by the brain tissue does not register on film. Unless an intracranial abnormality caused some deformation of the bone-structure, standard radiographs were of practically no use for detecting the presence of anomalies. Two techniques, pneumoenchephalography and ventriculography, both of which involved the introduction of a contrast medium into the cerebral ventricles were subsequently introduced to provide some intra-cranial visualization. In the former case the contrast medium was air [3], while for ventriculography an iodinated radiographic contrast material was injected. Even then, the visualization of tumors was by the indirect inference of changes in the normal appearance of the ventricles by the space-occupying lesions. Nevertheless, conventional radiography played an important role. At one time conventional skull radiography was the best means of detecting a displaced pineal gland, or bone erosion/hypertrophy, and ventriculography is still occasionally used for accurately outlining the ventricles, usually in conjunction with other imaging modalities, during image-guided surgery [4].

There are two limitations associated with standard radiography. In the first instance, it provides projections through the body and the three-dimensional information contained within is reduced to two, information about the dimension parallel to the beam having been lost. Secondly, x-rays diverge from a point source, so there is a differential magnification factor that distorts the perceived sizes of imaged structures. This makes it difficult to directly measure structure size, or to determine distances to, or between, target points by direct examination of the x-ray image. This latter objection can be overcome to some extent by constructing a very long focal-length (~9m) X-ray system, such that the rays are effectively parallel as they pass through the patient. This arrangement was employed by a number of centers to make intra-operative orthogonal anterior-posterior (AP) and lateral radiographs, and allowed the 3-D coordinates of structures that could be located uniquely in the two images to be measured with a ruler. Even so, three-dimensional perception of the imaged structure was not possible with such a pair of radiographs.

X-ray Angiography

Angiography involves the injection of a contrast agent into the blood vessels to increase their conspicuity against the surrounding tissue, and has been in use since the early days of X-rays [5]. However, it was not until 1963 when Ziedses des Plantes [6] described the principle of subtraction angiography, which allowed obscuring bony structures to be removed from the image by adding the post-contrast image to a negative of the pre-contrast radiograph. The ability to see only the vessels within the brain was a tremendous advance on what had previously been available. This technique received another enormous boost about 15 years later with a development by Mistretta and Ovitt and their colleagues [7,8] when they demonstrated automatic subtraction of the contrast and non-contrast images using electronic and digital means. Digital Subtraction Angiography (DSA) quickly gained favor amongst radiologists, and today intra-arterial DSA has become the standard angiographic imaging modality.

Stereoscopic Radiography

The means of acquiring 3-D images from radiographs were outlined by Elihu Thompson in 1897 [9], when he proposed acquiring a pair of views that could be visualized in a standard stereoscope (used for viewing stereoscopic photographs). Stereoscopic radiography gained tremendously in popularity during the first half of the century, but its use diminished with the heightened concern about dose hazards and with the increasing availability of modern 3-D imaging modalities. Nevertheless, many institutions continued to use stereoscopic radiography on a routine basis [10]. However, even stereoscopic radiography could only deliver a qualitative impression of three-dimensionality, and it was not until the introduction of DSA that image pairs were integrated within a computer-based stereoscopic display system to become useful adjuncts to surgical image planning and guidance [11-13] as well as diagnosis.

3-D imaging modalities

True 3-D imaging had to await the introduction of Computed Tomography (CT) and later Magnetic Resonance Imaging (MRI). Early versions of these imaging systems were rather more "multi-slice" than truly 3-D, but current examples from both modalities can produce images that display an effective resolution of better than 1mm in all three orthogonal dimensions. In MRI three-dimensional data acquisition is facilitated by true 3-D imaging pulse sequences that acquire data from all parts of the image volume simultaneously. This has the dual advantage that not only is the intensity and geometry of the individual slices consistent with one-another, but also the signal-to-noise of the reconstructed image is improved by a factor of $\sqrt{N}$ for an N-slice acquisition, compared with a "multi-slice" acquisition of the same data.

Computed Tomography (CT)

Conventional CT systems of the third (fan-beam detector) or fourth (ring detector) generation variety are generally considered to be geometrically accurate imaging devices. To produce 3-D images the individual slice images are stacked together and treated as a cohesive three-dimensional volume. During the scanning process, the patient is moved at a constant velocity through the scanning gantry in a direction perpendicular to the plane of the slices [14-16]. Note that it is important to ensure that the gantry "tilt" (i.e. the plane of the scanned slice relative to the axis of the patient), is set at 90º. Failure to ensure this condition results in skewed slices that are interpreted as being perpendicular to the patient axis, with the result that the image appears distorted [17]. Unfortunately, the disadvantage of using CT in the brain is that the contrast between gray and white matter is in the order of 0.5%, and depending on the x-ray beam intensity, or image filtering employed as part of the reconstruction algorithm, typical noise levels in CT are in the 0.2% to 1% range. Therefore many of the subtle anatomical details that indicate the presence of abnormalities are difficult to see on CT although they may be readily visualized with MRI.

Standard CT scanners still acquire the three-dimensional images on a slice-by-slice basis and therefore patient movement during the scanning sequence can also result in artifacts in the individual image slices, as well as distortions in the three-dimensional volume. In addition to the standard CT scanners, there are some available that acquire multiple slices simultaneously, as well as some that utilize two-dimensional image receptors (usually x-ray image intensifiers) to record the images [18]. Because the x-ray receptors on these devices are not ideal (the input window of the image intensifier is usually curved, and the earth's magnetic field may affect the electron optics of the image intensifier) they can potentially present problems of image distortion. Flat-panel solid-state detecting systems that solve some of these problems are gradually becoming available. However, even using a traditional image-intensifier, it has been demonstrated that with the appropriate calibration, these devices are capable of achieving accurate three-dimensional reconstructions of the human head at resolutions approaching 500μm. It is predicted that this modality will be extremely valuable for the planning and guidance of intracranial vascular procedures [18].

Magnetic Resonance Imaging (MRI)

MRI has become the principle imaging modality for image-guided surgery, particularly in the brain, because of its superb ability to discriminate between subtle differences in tissue characteristics. It is nevertheless not without its drawbacks. Since the spatial mapping in an MR image is performed by detecting small changes in the frequency and phase of the radio-frequency signals emitted by resonant hydrogen nuclei, any deviation from the ideal homogeneous main magnetic field, or from linearity in magnetic field gradients, will cause geometrical distortions in the images [19,20]. Even if the magnetic fields are ideal as far as the instrument is concerned, the object being scanned can have a dramatic effect on the local homogeneity of the magnetic field. This is particularly serious if the subject contains ferrous or other material of a high magnetic susceptibility (for example dental work or surgical clips). Even small changes in susceptibility inherent within the various tissue types within the human body resulting in distortions that go unnoticed on a qualitative basis, can nevertheless be serious if one is relying on the image to guide a delicate surgical procedure [20].

Techniques that measure field inhomogeneity as part of the image acquisition process have been described by Sumanaweera [19] and Dean [21]. In each case two images acquired at slightly different echo times allowed the additional phase accumulation caused by the field inhomogeneity to be measured, and the field inhomogeneity itself to be deduced. A useful method that can be employed to visualize the extent of spatial distortion directly, as it relates to an individual subject, is the SPAMM (SPAtial Modulation of Magnetization) sequence [22]. This imaging sequence superimposes an approximately distortion-free lattice of saturated spins (spins whose net magnetization is zero and which show up as black regions in the reconstructed image), within the volume to be imaged. The subsequent imaging sequence records the saturated spins as the absence of signal in a grid-like pattern, and the deviation of the geometry of the grid from the ideal is a direct measure of the amount of distortion in the image itself. This approach not only provides a distortion map that may subsequently be used to provide a geometrical correction to the images, but also allows for optimum sequences to be chosen, since the perceived distortion is highly pulse-sequence dependent.

MR Angiography (MRA)

While DSA has become the standard means of acquiring neuro-vascular information for surgery planning, it nevertheless has certain limitations. These include the radiation dose delivered to the patient, the intravenous contrast injection, and potential image image-intensifier distortions. In addition to these problems, DSA does not deliver a truly 3-D image. MR angiography, on the other hand, is fully 3D, and inherently registered with the 3-D anatomical MR image.

An MR imaging system is a remarkable device that can image many different attributes of tissue within the human body. In addition to the proton density, relaxation times, spectroscopic and functional information, it may also be “tuned” to respond only to moving material. There are two broad classes of MRA techniques that can be considered according to the component of magnetization utilized for motion sensitization: time-of-flight (TOF) and phase contrast (PC). It is beyond the scope of this discussion to give a full treatment of these angiographic techniques, however, a comprehensive discussion, along with the application of these techniques, is given by Peters and Pike [23].

The MRA image is inherently registered with the anatomy as defined by MRI, and in some cases the MRA information may be an integral part of the same dataset (as in PC MRA). So unlike DSA, it is often not necessary to perform an additional registration step required to register an MR angiogram with the anatomical MR image. A common means of displaying MRA images is the Maximum Intensity Projection (MIP) approach where the maximum value that each ray encounters, while traversing from one side of the volume to the other, is projected onto the imaging plane. This view resembles that of a projection X-ray angiogram. On the other hand, if a truly three-dimensional model is desired, the data must first be segmented from the background. This operation results in a series of slices that contain only the contours of the vessel cross-sections. These contours are subsequently tiled by a rendering algorithm to format them so that they may be displayed as surface objects.

The resolution of MRA is approximately that of MR imaging itself, i.e. 0.5 – 1.0mm, and so it is incapable of resolving individual capillaries, and may have difficulties representing the smaller vessels. The MRA techniques outlined above may also have trouble imaging flow in the plane of the imaging slice (TOF), slow and turbulent flow (PC), and neither can unambiguously differentiate between blood flow in arteries and veins. Recently, contrast-enhanced MRA, which involves the injection of MR contrast agent into the vascular system, has become popular. While this form of MRA is invasive, and more expensive than the non-contrast studies, it is a more reliable means of visualizing the vessels in the brain.

Ultrasound

The history of ultrasound imaging in surgical guidance is quite recent. Originally introduced in the 1950s, the medical use of ultrasound progressed slowly from A-mode systems producing oscilloscope traces of

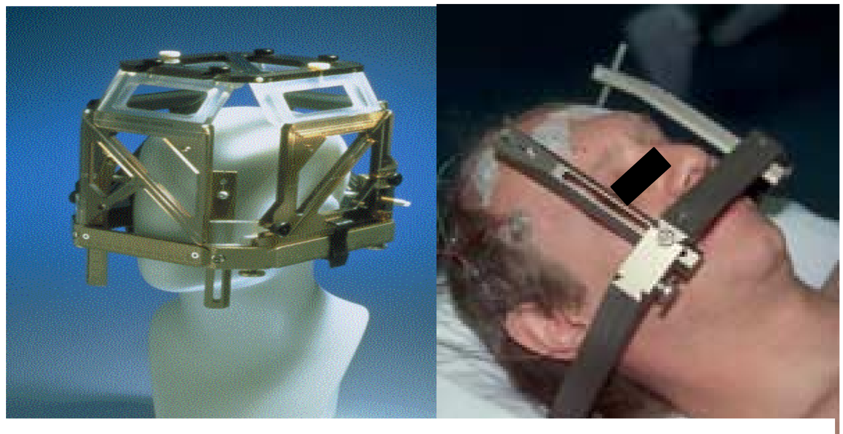

Figure 1
a) A typical stereotactic head-frame, equipped with fiducial marker plates, showing the means of mounting to the head, (Courtesy Elekta Instruments AB); b) the same frame, without the plates, mounted on a patient.

acoustic reflections, to B-mode gray-scale images of the anatomy, to those producing real-time tomographic images of the anatomy and blood flow. The image quality of medical ultrasound has advanced from low resolution bi-stable images to those with sufficient detail to make ultrasound an important, and at times, indispensable modality in obstetrics and in the diagnosis and management of a large number of diseases.

Its major advantages are that an ultrasound probe is small and can be manipulated with great flexibility allowing the generation of real-time tomographic images at user controlled orientations and positions, effectively delineating structures within the body with a resolution ranging from about 0.2 mm to 2.0 mm. Also, ultrasound instrumentation is inexpensive, compact and mobile , and does not require special facilities to permit intermittent use in the operating or therapy suite. Ultrasound can also provide real-time images of blood velocity and flow allowing the physician to map vascular structures ranging in size from arteries to neovasculature in tumors.

In spite of these attributes, ultrasound still suffers from a number of disadvantages such as the specular nature of the images, shadowing, multiple reflection artifacts and variable contrast. In addition, the quality of the ultrasound image is highly operator dependent, and as such its full potential in surgical planning and guidance has not yet been realized. Nevertheless, the real-time capabilities of US outweigh many of the deficiencies, and for this reason, US has been used for many years to guide therapy and surgery in the breast [24,25], prostate [26,27], liver [28], and brain [29]. The explicit use of US in image-guided neurosurgery applications is discussed later.

STEREOTACTIC SURGICAL FRAMES

Image-guided surgery has its roots in the field of stereotaxis or stereotactic surgery, even if the techniques at the time were not specifically “image-guided” in the sense that we understand this term today. Horsley and Clarke [30] first used Cartesian co-ordinates to establish the positions of target points within a monkey brain relative to fixed landmarks on the skull itself. Although the idea behind this technique led to a patent for a human apparatus [31], it was not developed further for this use at that time. Aubrey Mussen, an engineer who had worked with the Horsley-Clarke system, later designed a similar device for the human skull [32]. This apparatus was not discovered until some 30 years after his death, wrapped in a newspaper bearing the presumed date of its manufacture. Spiegel and Wycis in 1947 [33] further developed the technique of stereotaxy, using the concept of a three-dimensional Cartesian co-ordinate system for the human brain. These co-ordinates were defined with respect both to the brain, and to radiographic images acquired while the patient was fitted with a reference (or stereotactic) frame attached to the head.

The stereotactic frame forms the gold-standard of reference systems for image-guided surgery. While there are many different designs, they perform basically the same tasks, i.e. provide a rigid reference that establishes a coordinate system relative to the patient, furnish recognizable landmarks in the images, and provide a stable mounting and guide for instruments used to perform the procedure (Figure 1). Their differences lie in the manner in which they are applied to the head, and the way the trajectory to the target point is specified. For example, some frames are constructed in such a manner that the target may be positioned at the isocenter of an arc used to hold the instruments. When employing the arc, with the target at the isocenter, the target may be approached with any azimuthal or declination angle, with the assurance that the trajectory will always intersect the target. Prior to the advent of real-time tracking of surgical probes, the scales on the frame also provided the means to track the position of the instrument by reference to the scale-markings on the frame supports.

Early use of stereotactic frames in conjunction with computed tomography was reported by Bergstrom and

Greitz in 1976 [34] and this was followed by many other reports describing the adaptation of CT to stereotactic surgery. In 1979, Brown [35] formally outlined a configuration of fiducial markers attached to the frame that could be imaged with CT and which could be recognized in the images. Brown described the methodology of establishing the frame coordinates of any pixel in the image, based on the locations in the images of the markers. The configuration of parallel and oblique rods surrounding the frame form the basis of frame-mounted stereotactic fiducial markers in use today.

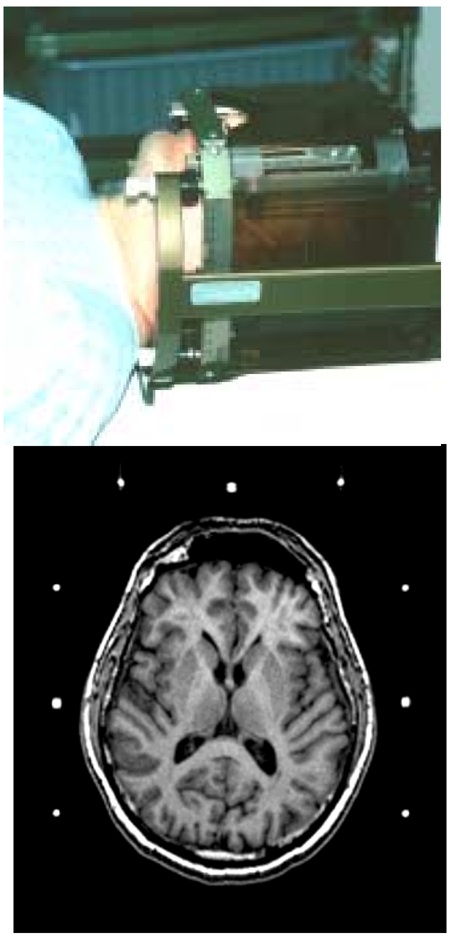

Figure 2
a) The stereotactic frame, including fiducial marker plates, attached to the patient's head prior to an imaging study. The plates only present during the imaging study, and are removed during surgery. b) Image-space representation of the fiducial markers embedded in the plates, as seen in a transverse image.

Stereotactic frames come equipped with two types of fiducial marker. The first is designed for imaging by a tomographic imaging system and generally consists of a series of 'Z', 'N' or 'V' bars mounted on the frame. They are configured in such a way that they are intersected by the image slices, and their configuration as they appear in the images is unique for any slice (Figure 2). Indeed, if a slice intersects at least three sets of markers, its oblique position within the volume may be determined uniquely. Of course, registration of a three-dimensional volume can be achieved with a minimum of just three points within the volume. However, the stereotactic frame predated the true 3-D imaging modalities and planning was performed manually on a slice-by slice basis – a procedure that is still followed today in many centers.

The other fiducial marker configuration is designed specifically for use in X-ray projection images, and consists of two sets of (usually four) point objects on plates, mounted perpendicular to the center-line of the X-ray beam, and placed on each side of the subject. Each of these points shows up clearly in the projection image, and their positions can be used to reconstruct the imaging geometry exactly. With this approach, orthogonal or stereoscopic image pairs can be employed to perform three-dimensional localization of structures within the body.

From the earliest days of stereotactic surgery, images were used as guidance to the targets. Initially, projection radiographs (ventriculograms) were the only available means of identifying landmarks, and charts overlaid on the radiographic images were used to calculate the frame coordinates of the target. However, unless the precise depth of the structure was known a-priori, precise measurements were difficult. With the introduction of tomographic imaging modalities, this procedure was simplified considerably, since all structures in a slice could be localized within a well-defined plane, and the coordinates within that plane could be read off directly. This technique is still often employed directly using a transparent overlay on the video monitor of a CT or MR system, or indirectly using the measurement tools built into standard diagnostic imaging devices. Many authors have demonstrated that through the use of a stereotactic frame, the registration between the images and patient can be achieved with an in-plane accuracy and precision in the order of 1mm [36]. This result can be achieved also in the longitudinal direction with modern isotropically resolved imaging systems.

Initially, CT was the tomographic modality used with the frame to create images for stereotactic surgical guidance. When MRI arrived on the scene, the same methodology was employed. However, certain metallic components of the stereotactic frame that could be imaged with CT without creating serious image artifacts, had to be replaced in the MRI compatible frame to avoid undesirable distortions caused by their influence on the magnetic field. Whereas the fiducial markers for CT were typically manufactured from copper or aluminum, for MRI they had to be replaced by channels in plastic plates filled with a fluid, typically a copper sulfate solution, that produce a bright signal in an MR image [37]. In fact, such channels filled with the solution of a water-based x-ray contrast agent can produce images of the fiducial markers with both MRI and CT. It must be pointed out that while the use of oil in MR fiducial markers has been reported, this is to be discouraged since the resonance frequency of oil-based protons is slightly different from free water. Depending of the imaging pulse sequence employed, this can result in the fiducial markers being shifted from their proper positions by several pixels!

COMPUTERIZED PLANNING AND GUIDANCE SYSTEMS

Even before the advent of CT there were reports of computers being used in the operating room to assist in the planning of stereotactic neurosurgery. The imaging modalities available at the time were x-ray ventriculography and brain atlases. Pioneering work in the early 1970s by Bertrand, Olivier and Thompson [38,39], set the stage for many of the computer-assisted surgery applications that have subsequently evolved.

Frame-based systems

Computerized planning systems made their debut in the early 1980’s with simple programs that allowed for the establishment of coordinate systems based on fiducial markers that were recognized in the image planes. Initially these systems were based on software running on the computers controlling the imaging devices themselves At that time CT and MR scanners were typically controlled by 16-bit mini-computers with 128-256KB memory, and special-purpose planning programs were developed for each computer operating system, and each modality [40]. These systems were soon followed by independent computer stations that could read images from different modalities, and from images acquired from scanners from a variety of scanner manufacturers [12]. The next logical step was to modify the software so that the images from multiple modalities could be combined and surgical planning could take place using information from a combination of MRI, CT and angiographic images. Such multi-modality imaging was considered important for certain procedures, such as the insertion of probes or electrodes into the brain. The ability to simultaneously visualize the trajectory with respect to images of the blood vessels (either using orthogonal projections or stereoscopic image pairs), enabled the pathway to be planned with the confidence that it traversed the brain within a vascular-free zone [11,12].

Frameless systems

In spite of the advantages of the stereotactic frame, most involve a minimally invasive surgical procedure to fasten them to the patient, (either through the creation of 3-4 1-2 mm deep holes in the skull with a twist-drill to accept blunt pins, or sharp pins applied to the skull under pressure). The frame (or in some cases just the base-ring which can be separated from other structures) may also present unnecessary clutter in the operating room. Hence, there is a general desire to be rid of the frame if possible. However, without the frame to provide the fiducial markers, other means must be employed to register the patient to the image(s).

REGISTRATION AND TRACKING

If images are to be useful in planning or guiding surgical procedures, they must be registered to the patient, such that image-space (defined by the 3-D array of voxels) can be related directly to the real-world coordinate system of the patient. This process occurs in two steps. First the homologous points or surfaces must be identified in the image and on the patient, and then this information must be used to compute the transformation matrix that relates the coordinate system of the image to that of the patient.

Point-matching

A great deal of effort has been expended on the development of techniques to register pre-operative images to the patient, without the use of frames. The most common involve the identification of structures and surface patches in both the images and the patient. One approach is to use a computer tracked pointer to identify landmarks such as the outer canthi of the eyes, the tragus of the ears, and the nasion. These same structures are then identified using a cursor driven mouse within the three-dimensional images of the patient. This approach is not entirely satisfactory for two reasons. Firstly, there is bound to be some variation in the identified locations of the landmark points on the patient, as well as a problem of identifying exactly the same locations within the patient’s three-dimensional image. Therefore the best one can hope to do is to perform a least-squares approximation to the correct registration. Of course, the accuracy of the least-squares match will be improved as the number of homologous points is increased. Secondly, if all of the registration points are clustered in one area, a small inaccuracy in the registration in the region containing the homologous points, can result in a magnified error at points remote from this region. This phenomenon was originally

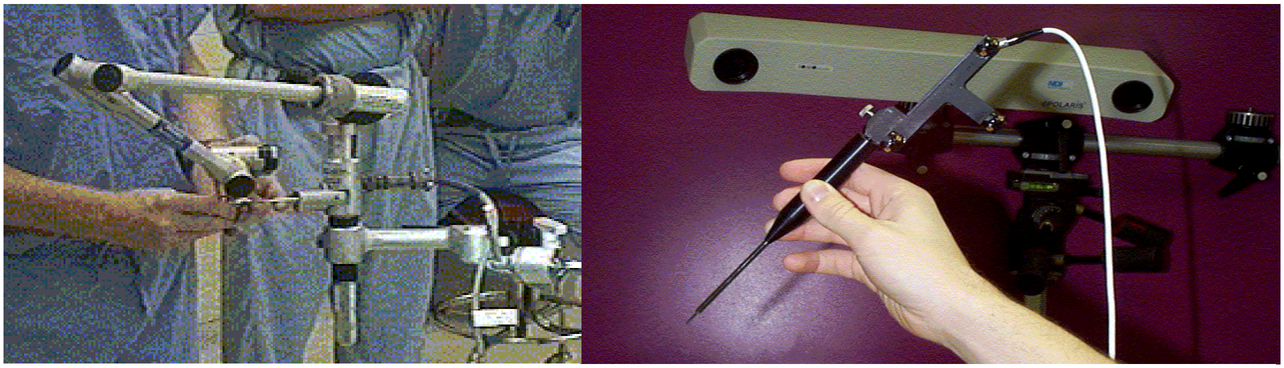

Figure 3.
Tracking devices.
a) Six degree-of-freedom mechanical tracking device (the FARO arm, Faro Industries Inc, FL).
b) b) Polaris optical tracking device (Northern Digital Inc, Waterloo ON)

pointed out by Peters *et al.*[13] and put on a rigorous footing recently by Maurer [41].

Surface matching

Some systems employ a surface-matching approach to complement the point matching method. The use of surfaces extracted from 3-D patient images acquired from multiple imaging modalities, to match these images in a common space, was described by Pelizzari and Chen in 1989 [42]. While their approach required two digital images, the same procedure can be followed to match the surface extracted an image with physical samples of the same surface of the object. This technique involves using the probe to sample points on the surface of the patient, and then determining the best match of this point-cloud to an extracted surface from the 3-D patient image. This combined approach using both points and surfaces is described by Maurer [43] and is incorporated in at least one commonly used commercial image guided neurosurgical system. Under ideal conditions, (i.e. in phantom tests where homologous structures are easily identified and there is no movement of the markers with respect to the object), accuracy approaching that of stereotactic frames can be achieved [44]. However, under clinical conditions, where natural features on the patient's skin are identified, the accuracy obtainable from his type of approach decreases due to the subjective identification of homologous point-pairs on the patient and in the images. While this may be adequate for many neurosurgical purposes, it is not appropriate for procedures requiring high precision.

Bone-mounted markers

The accuracy and precision of point matching procedures can be improved somewhat by using surface markers glued to the patient's skin. In this case their location can be more precisely determined using the pointer, and they can be automatically identified within the patient's three-dimensional images. While this improves the precision of the matching, there remains the problem that skin mounted markers can move with respect to the underlying bony anatomy, and therefore add additional error. Maurer *et al.* [45] have demonstrated convincingly that the only way to achieve patient-to-image registration with the same accuracy as can be obtained with a stereotactic frame, is through the use of bone-mounted fiducial markers. Even though the implantation of these reference markers constitutes a procedure that is approximately as invasive as the installation of a stereotactic frame, it can nevertheless be performed under local anesthetic and represents a level of invasiveness that is minor compared to the surgical procedure that is to follow. Maurer shows that the targeting accuracy obtainable with properly designed markers can be in the order of 0.5 to 1.5mm.

Tracking Systems

The move away from frame-based to frame-less stereotactic surgical procedures was made possible both by the availability of computers sufficiently powerful to manipulate and display three-dimensional medical images, and by the development of systems that allow probes and instruments to be tracked by the computer during the procedure. Many approaches exist to perform intra-operative surgical guidance. Some employ a probe that is physically linked by a multi-jointed arm to the apparatus restraining the patient's head, and the position of the probe is determined by sensing the angles at each joint [44,46]. Others use ultrasonic, optical [47], or magnetic [48], methods to determine the probe's position. Examples of optical and mechanical trackers are illustrated by Figure 3. Each system has its advantages and drawbacks. The mechanical system is always in communication with the computer and does not need to maintain sight- lines between the probe and some signal transducer. On the other hand, it is bulky and intrusive in the operating room.

Ultrasonic, electromagnetic and optical methods allow the mechanical system to be dispensed with, but in the case of ultrasound and optical systems, there must always be an unobstructed line of sight between sensors on the probe and some transmitting device. Electromagnetic systems do not suffer from this disadvantage, but their performance is often limited by the presence of metallic devices in the vicinity of the position sensors. Some efforts have been made to combine optical and electromagnetic devices to build instruments that use an optically tracked system to re-calibrate a magnetically based tracker in real time so that the magnetic tracker can "take over" from the optical system when sight-lines are broken [48]. Currently both mechanical and optical tracking systems are employed routinely in image guided surgery procedures, and these devices generally achieve a tracking accuracy and precision of better than ±1mm [49].

Some of the more recent optical systems use passive tracking technology where reflecting objects attached to the probe are recognized by a multi-receptor optical imaging system. Currently, there are no satisfactory solutions to the problem of tracking an object (for example a catheter tip or a flexible endoscope) within the body unless one employs a real-time imaging system like MRI [50-52] ultrasound [53] or fluoroscopy. The size of the standard magnetic receptor precludes it from being incorporated within a probe-tip or catheter, forcing the tracking of an intra-cavity point to be achieved through tracking of a sensor connected to the probe-tip by a rigid rod. The use of a miniaturized magnetic probe that can be inserted within a catheter, has been reported, [54] but details of its performance are sketchy at this time. One technology that shows promise for use in tracking situations in general, and in flexible endoscopes and catheters in particular, employs fiber-optics. Specially treated optical fibers can be manufactured in such a way that the light transmission is a function of the radius of curvature of the fiber. This concept has been incorporated into a commercial shape-measuring device "Shape-tape" [55,56] which allows the shape of a stiff ribbon to be constantly monitored in 3-dimensional space.

Most of the devices in routine use currently report accuracies and precisions of better than 1mm, which is generally accepted as a target performance standard for tracking systems.

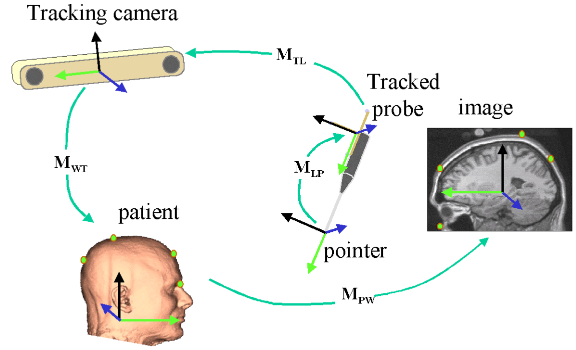

Figure 4.
Spatial transforms.
Several transformation matrices must be computed to enable the tracking of an intra-operative tool, and relate its position to the patient's image: M_{TL} represents the transform between the coordinate system of the optical tracking device, and the LED infra-red emitters on the tracked probe. M_{LP} related these emitters to the probe tip (or other instrument being tracked); M_{WT} is the transform between the tracking device and "world" coordinates (the physical patient), and M_{PW} the transform that maps the patient coordinate system to the image. Each of these transformation matrices must be carefully determined from calibration procedures.

Coordinate Transformations

Central to the success of image-guided surgery systems, is the ability to relate the measurements made in one coordinate system (or frame of reference) to that of another. When dealing with the head (and image) in a stereotactic frame, this is relatively straightforward, since the coordinate system of the images is aligned with that of the frame. However, when the orientation of the frame coordinate system bears no relationship to the imaging reference frame, it is necessary to transform one coordinate system to the other. In a typical application, the tracking of an ultrasound transducer for example, there may be several coordinate transformations that must be computed before the images recorded by the transducer can be properly registered with the image of the patient. As shown in Figure 4, the process of arriving at the final composite transform, involves a concatenation of the individual transformations. In addition, occasionally one is concerned with displaying the image of the tracking device within both a 2-D image (DSA for example) and a 3-D image (MRI). In the former case, where the image is a diverging-ray projection through the 3-D object, this process involves the use of a homogeneous transformation matrix [17] which performs a many-to-one transform when moving between the 3-D and 2-D images (i.e. multiple points in the 3-D image map into a single point in the 2-D image). Conversely, a one-to-many transform is computed when moving in the reverse direction (i.e. mapping a point in the 2-D image maps into a line in 3-D).

MULTI-MODALITY IMAGING

Although CT was the primary imaging modality originally employed for stereotactic surgical planning, it soon became evident that other imaging modalities could complement the information from CT. In particular, MRI and digital subtraction angiography provide information not available in the CT scan. For this reason, multi-modality image display and analysis systems were developed to allow the planning and guidance operation to take place using multiple image data sets simultaneously [17,40,57]. In addition to making use of anatomical information from different imaging modalities, (bone from CT and soft tissue from MR for example), the practice of integrating functional images from functional magnetic resonance imaging (fMRI), positron emission tomography (PET) or single-photon emission tomography (SPECT) is possible. The integration of functional and anatomical information is generally performed to identify functionally active regions in the brain which should be avoided during surgery [58]. A large body of literature on image registration exists, and a comprehensive review has recently been published by Maintz and Viergever [59].

Even though MRI and CT are three-dimensional imaging systems and digital subtraction angiography is two-dimensional, it is nevertheless possible to display the projection of a point identified in the 3-D modality within the 2-D angiogram. Likewise a point within the planar angiogram, can be represented as a line within the MRI or CT volume. By employing orthogonal or stereoscopic angiogram pairs, three-dimensional localization can be achieved within the angiogram-defined space. This multi-modality approach was particularly important when planning trajectories to deep brain structures, or when implanting EEG recording electrodes during a procedure to localize the foci of epileptic seizures [60].

Video

Video imaging can add a great deal of information to an image guided surgical procedure. Often the integration of video images, which are acquired live during the surgical procedure, with the 3-D preoperative images, is referred to as enhanced or augmented reality (see discussion below). Kikinis [61] demonstrated the efficacy of integrating a video image of the operative site with the graphical representation of a tumor or other target structure. This approach enables the surgeon to "see-through" the patient's skin into the operative site, and can facilitate the planning of the optimal craniotomy required to approach the target.

Microscopes

Since many surgical procedures involve the use of an operating microscope, the integration of the microscope images with those obtained from preoperative MRI and CT scans has obvious advantages. In this case the images from a tracked microscope can be integrated electronically in a digital or analog fashion, with the combined images being displayed on a video or computer monitor. Alternatively, the images from the preoperative scans can be projected into the visual field of the microscope [62] (Figure 5). This technology has also been incorporated into head-mounted displays where the position of the surgeon wearing such a display is tracked and the preoperative image corresponding to the viewpoint of the surgeon is projected on his field of view [63].

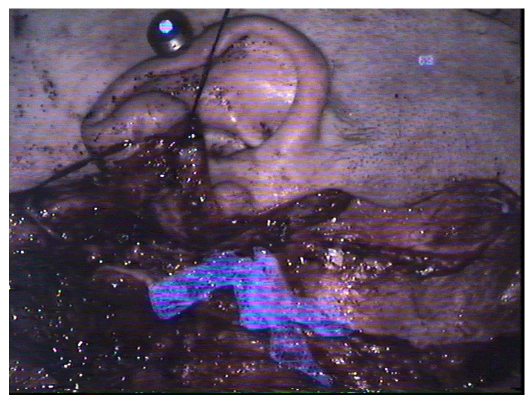

Figure 5.
Example of an anatomical structure (zygomatic arch in blue) extracted from a CT scan, and overlaid on the microscope image of a patient undergoing surgery to remove a petrous apex cyst. The cyst was very close to the carotid artery, making the procedure potentially hazardous. Registration was performed using bone-implanted fiducials, and overlays of various structures such as the zygomatic arch, the carotid artery and the cyst provided guidance throughout the procedure. (Image courtesy Prof David Hawkes, Guy's Hospital, London UK).

Endoscopes

While an operating microscope can "see" into an operating site through a skull opening greater than about 2 cm in diameter, when we seek to directly visualize the target through smaller openings, other means must be employed. The endoscope has been used for many years to gain visual access to body cavities, and has been employed in neurosurgery to assist in surgical procedures within the ventricles. A technique that has become popular recently is "virtual endoscopy" [64,65]. This approach uses computer graphics to simulate the view seen by an endoscope placed in a particular body cavity, based on the representation of the cavity derived from preoperative MRI or CT images. With the increase in emphasis on minimally invasive surgery, there has recently been an active interest in combining the video images from standard endoscopy with the computer-generated images of virtual endoscopy. The aim here is to place the endoscope image that one observes during a minimally invasive surgical procedure, into its proper context by merging it with the equivalent surface extracted from the preoperative images. Several authors have recently reported experience with the clinical application of combining images of the ventricle wall obtained from a tracked endoscope, with the equivalent images from CT or MRI [66-68]. The simplest mode of operation of this approach is to simply track the tip of the endoscope, and display the tracked point on orthogonal image slices from the original data. The most convenient means of achieving this is to display the three slices that intersect at the tracked point. While useful, this approach does not present the endoscopic image in its proper context. To achieve this, a surface representation of the cavity must be constructed, and the video image mapped to this surface.

TISSUE MOVEMENT DURING IMAGE-GUIDED PROCEDURES

Image guided surgery procedures divide naturally into two distinct categories, namely traditional stereotactic approaches where targets deep within the brain are approached via a probe inserted through a twist drill or burr hole; and open craniotomies. In the former case very little tissue movement is expected within the cranium between the imaging and surgical procedures, and the registration accuracy between the patient and the images at the time of imaging is maintained during the operation. Nevertheless, even in these stereotactic procedures, loss of cerebral-spinal fluid (CSF) can cause tissue shifts and must be guarded against. In the latter case however, as soon as a craniotomy is performed, brain tissue can change shape significantly.

All conventional image-guided surgery systems operate on the assumption that the tissue being operated on can be treated as a rigid body that does not change its form between the time the pre-operative images are acquired and the surgery is performed. Thus, the concerns about image resolution, registration and tracking accuracy, along with geometrical distortion of images is only half the battle. Prior to surgery, various procedures are commonly employed to reduce intra-cranial pressure, including the administration of steroids to reduce inflammation; control of cerebral blood volume by hyperventilation, or tilting the bed to increase venous drainage and the reduction of water content by administering the drug mannitol. In addition, during an open craniotomy, tissue can move under the influence of gravity, causing the brain as presented to the surgeon to differ morphologically from that represented by the pre-operative images. All of these factors present the surgeon with an additional source of error that is not accounted for during the image-to-patient registration step.

Brain-shift measurement

It is perhaps surprising, given the potential impact of brain shift during image-guided neurosurgery, that until now so little emphasis has been placed on quantifying its effect. However, in recent years, attention has turned to this problem and various investigators [69-71] have indicated that, depending on the procedure, tissue shifts of up to 20 mm or more can be observed under routine surgical conditions.

Maurer *et al.* [41] employed intra-operative MR images to assess the extent of tissue motion by comparing pre-operative images with those obtained during surgery. They reported measuring brain shifts of up to 5mm, and ventricular volume changes of up to 44%. A similar study performed by Hill *et al.* [69] observed the displacement of reference points on the cortex after craniotomy with a tracked surgical microscope. They reported median tissue displacements on a sample of five patients of up to 7.4mm. Roberts *et al.* [71] recently published an extensive study of 28 patients where they used a tracked surgical microscope to measure the shift of identifiable points on the cortex from their ideal positions as defined by the pre-operative MR scans. For this population of patients, they reported a mean displacement of 10mm, and shifts of up to 24mm.

Correction for intra-surgical brain shift

Concurrent with the efforts to measure tissue shift, an increasing number of authors has begun investigating means of correcting for the distortion that takes place during open-craniotomy surgical procedures. Three distinct directions have emerged. Two attempt to predict volumetric changes on the basis of the anticipated effect of drugs on one hand, and the observed changes to cortical surface on the other. The first of these models the effect of drugs on the drainage of fluid from the brain (resulting in a volume decrease) and the effect of tissue movement under the influence of gravity, respectively. Miga [72] employed finite element models of brain tissue that incorporate both its mechanical properties and the manner in which it retains fluid. At the other end of the spectrum is an approach that scans the surface of the cortex with a laser range-finder, and matches the measured surface map with the cortical surface as determined from a pre-operative MR image [73]. As surgery proceeds, and the shape of the cortex is observed to change, the MRI-derived cortical model is deformed to match the surface as measured by the laser scanner. The effect of this deformed surface is propagated through a volumetric, finite-element representation of brain structure in order to predict the extent of the tissue distortion throughout the volume.

The third approach to this problem centers of the acquisition of real-time images of the brain during a surgical procedure, and either use these images directly to guide surgery, or employing them to correct pre-operative images to conform to the intra-operative observations.

These methods, employing both intra-operative ultrasound and MRI, more accurately fall under the category of "Interventional Imaging", which is discussed later.

DISPLAY SYSTEMS

As discussed above, the essence of an image-guided surgical system is its tracking device. However, an equally important component is the display; it must be capable of displaying the image in three-dimensions, either by presenting the observer with orthogonal planes that follow position of the tracked probe, or else display a representation of the probe within a three-dimensional rendition of the patient image. To be an effective tool in the operating room (OR), it must be truly interactive.

Multi-planar image display

Multi-planar image representation is as old as tomographic image acquisition, and even when the resolution of early systems was far from isotropic, vertical "slices" through the data were demonstrated to be useful in navigating through the images in three dimensions. Even though the voxel dimensions of early imaging systems were of the order of 1.5 x 1.5 x 13 mm^3, multi-planar representations were popular. Today, with both CT and MRI, it is possible to acquire the data isotropically with $1mm^3$ voxels. While early computer systems were only capable of slicing the data along orthogonal axes (along rows and columns), today's systems enable arbitrary slices to be extracted from the images interactively. Most image-guided surgery systems also allow the user to select views that are perpendicular to the direction, or in the plane, of the probe.

Three-dimensional Displays

The traditional means of displaying medical images was an x-ray film on a view-box, and even with the advent of 3-D imaging techniques like CT and MRI, until recently images were still viewed slice-by slice, either as films on a standard light-box, or on a computer display. However, as the volume of data in a typical image increased, it became increasingly necessary to display the volume as a single entity, rather than as a collection of slices. The display of these data (volumes of voxels) is achieved either by projecting surfaces within the volume [74], or the entire volume itself, onto a viewing screen [75,76]. These two approaches are known as surface rendering and volume rendering respectively.

Surface rendering demands that the surfaces of structures of interest be identified within the volume (by automatic or manual contouring methods) and is the most common method of displaying 3-D medical images. It is often used in association with texture-mapping to "paint" characteristics of the original data (voxels) onto selected surfaces within the 3-D images. An example of a combined surface/texture-map display is given by Figure 6. Volume-rendering, on the other hand, employs a ray-tracing technique where the intensity of each point in the projected image is a function (line integral, maximum intensity etc) of the structure traversed by each ray. In either case, contemporary computer workstations are sufficiently powerful to allow interactive manipulation of either kind of image.

Stereoscopic Displays

Stereoscopic imaging is no longer in general use in diagnostic radiology, although there are some centers that still employ it routinely, while others use it on a sporadic basis. However, since the advent of three-dimensional imaging modalities, there is an increased need to navigate through large data spaces quickly. The use of 3-D visualization in surgical planning and guidance, and the availability of inexpensive computational power that permits the interactive manipulation of three-dimensional volumes, has once again made stereoscopic visualization a compelling adjunct to surgical navigation. Stereoscopic visualization has been employed in the operating room by the surgical team at the Montreal Neurological Institute (MNI) for a number of years, initially for the

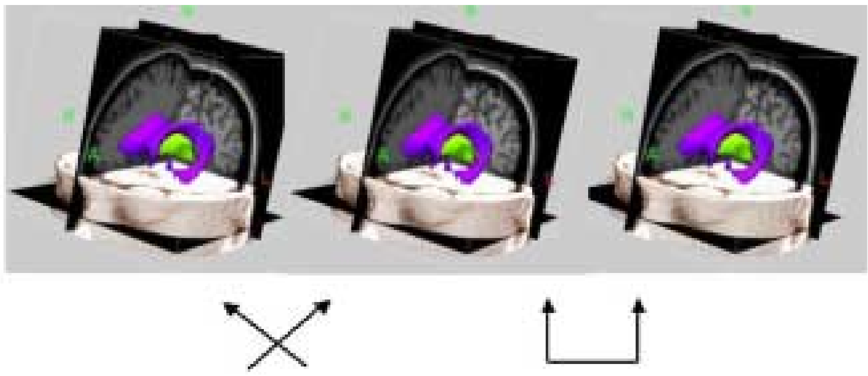

Figure 6.
Typical display from an image-guided neurosurgery system showing texture-mapped MR data on orthogonal planes, along with surface-rendered ventricles.

This trio of images, and those in Figure 7, are intended to be viewed stereoscopically. The left two of the pair are designed for "cross-eye" viewing, while the right pair are appropriate for direct viewing. To view an image with crossed-eye stereo, focus the eyes on a finger placed above the left-hand image pair, and move the finger until a third image appears between the other two. To view with parallel geometry, either converge the eyes at a point far beyond the right-hand image pair, or hold the images close to the eyes and gradually move the page away, until a fused stereoscopic image appears.

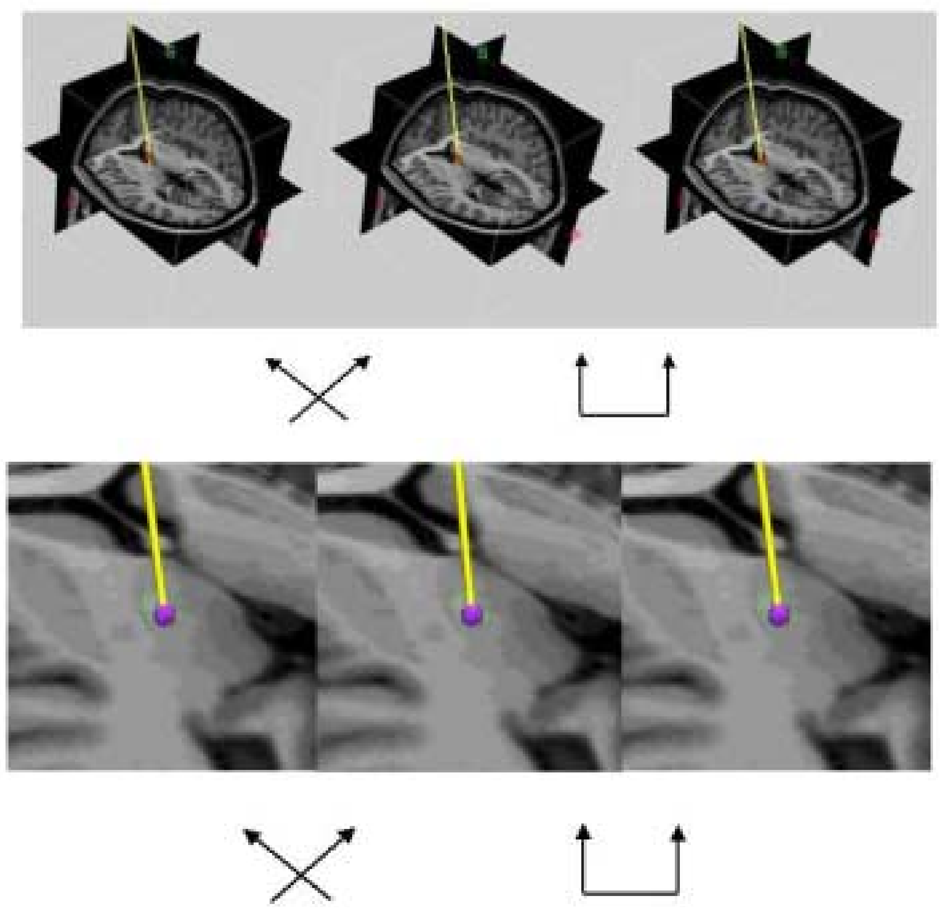

Figure 7.
a) Probe positioning a model lesion in the thalamus during an image-guided neurosurgical procedure.

b) Close-up of probe and lesion being placed within the thalamus.

integration of digital subtraction angiograms with MRI and CT for surgical planning purposes [12,13,77]. Subsequently, this group incorporated stereoscopic visualization into the three-dimensional views presented to the surgeon during surgical guidance, displaying dynamic representations of the surgical probe along with surfaces and MRI slice data [13,78]. More recently they have extended the use of stereoscopic visualization to the planning of thalamotomies and pallidotomies for the treatment of Parkinson's disease [4]. Here, stereoscopic visualization has proved to be a major advantage when planning the shape and position of a three-dimensional lesion that must be placed precisely with respect to a functional atlas that has been merged with the data (Figure 7).

Virtual reality

By virtue of the fact that image-guided surgery relies on the use of a "virtual" representation of the body or specific organ in the form of medical images, all of the techniques discussed above contain elements of "virtual reality" (VR). While the popular vision of virtual reality is centered mainly on the entertainment industry, it has significant applications in many other fields, including remote sensing and robotics. A full treatment of virtual reality in medicine is given by Englmeier [79].

In image-guided surgery, VR techniques are beginning to play increasingly important roles as the emphasis grows towards the use of minimally invasive procedures. Since direct visualization of the surgical field becomes more and more difficult as the surgical opening becomes smaller, the natural direction of research is towards the model-based realization of the structures being operated. Computer graphics technology can provide realistic 3-D images of structures derived from MRI or CT, and the emerging field of virtual endoscopy [64,65,80,81] provides a prime example of this. Both surface- and volume-rendering approaches are commonly employed in this field, and stereoscopic visualization is often employed to maximize the advantage of the 3-D image representation.

A useful adjunct to VR is "Augmented Reality" where real-time information from the "real-world" is integrated with that from the 3-D model. This approach has seen extensive application in remote sensing and has been demonstrated in the context of image-guided surgery by Gleason *et al.*[61] who superimposed video images with the 3-D computed representation of the brain during neurosurgery as a means of combining real (patient) and virtual (3-D MRI) images. Other authors [62] were concerned with the integration of microscope images with digital representations of the operative field, while Davey [78], and Pisani [82] combined live stereoscopic video images with the 3-D MRI.

Head-mounted displays

The advances over the last few years in computer and imaging technology have naturally led to the application of sophisticated (and formerly excessively expensive) approaches to image-guided surgery. One such device is the head-mounted display (HMD) [63], which can incorporate stereo visualization with merged direct-view and computer model images. It allows the surgeon a direct view of the operation, while at the same time projecting a registered virtual image (from 3-D MRI for example) onto his visual field. As long as the head is accurately tracked during the procedure, the images will remain superimposed regardless of the surgeons head position. One of the greatest advantages of this approach is that it keeps the images that would normally be displayed on an inconveniently placed computer monitor, within the surgeon's field-of-view at all times. With this system, images from the pre-operative studies are always in view. In addition, the real-time, video-based endoscopic images, being used for navigation within the ventricles for example, are always available to the surgeon without the need to turn away from the site of the operation.

ROBOTICS

There have recently been a number of examples of the use of robotics in surgical planning and guidance. Perhaps the most effective and widespread is its use in the surgical implantation of hip prostheses into the femoral canal, as introduced by Paul *et al.* [83] The major breakthrough in this field was that the prosthesis shaft could be constructed precisely according the geometry of the femur as determined from 3D CT scans, and the robot was then employed to manipulate the implant into its final position. They suggest that the in-growth process is facilitated when the surgeon achieves a satisfactory fit for the prosthesis. Their objective was to use image-directed surgical robotics to precisely ream the femoral canal to obtain an exact match with the prosthesis.

Sawaya *et al.* [84] speculate on the possibilities of using robotically assisted procedures and virtual reality to improve tumor surgery, while Benabid [85] recently published an excellent review on (neuro)surgical robotics, discussing many of the early contributions, including the pioneering work of Kwoh and Young [86,87], who demonstrated the use of robots to position instruments in a stereotactic setting. He also describes later implementations of this concept in other centers using commercially available robotic arms. One extension to these ideas that has been implemented commercially, is the neurosurgical operative microscope as described by Giorgi [88]. Here a robotic arm is connected to a microscope, while force feedback sensors drive the motors of the arm in response to the positioning of the microscope by the surgeon. The coordinates of the microscope focal point (and potentially the images acquired by the microscope optics), are integrated with the standard multi-planar and 3D MRI views of the patient on the surgical workstation. Chan [89] describes a prototype system that incorporates preoperative image-guided planning, minimal-invasive and non-invasive calibration between the virtual (patient images) and actual environment, and a surgical robot for location and remote-controlled procedures based on the preoperative strategy.

While much of the discussion of robotics in surgery has centered around the brain, Stephenson *et al.* [90] propose the use of robotics to overcome many of the limitations have become apparent in cardiac surgery, particularly the lack of adequate precision with standard endoscopic instruments. They demonstrate the use of an endoscope, controlled by a microsurgical robotic system, to perform coronary anastomoses on an isolated porcine heart. The fundamental contribution made by this system is that the computer control system eliminates tremor, and motion scaling allows for precise maneuvering of the endoscope and instrumentation. The first application of this approach to coronary by-pass surgery, of a beating human heart in an intact chest was performed in October 1999 at the London Health Science Centre, London, Canada [136].

MINIMALLY INVASIVE TECHNIQUES

Standard IGNS techniques yield superb visualization of the anatomy (both in 2-D planes and 3-D reconstructed images) relating to the lesions and target. However, the systems used for tracking intra-operative probes vary in their accuracy due to inherent limitations in the hardware and technologies employed [44], and to the techniques used to register the images to the patient [45]. Also, as discussed earlier, they do not account for the fact that brain tissue almost always shifts during open-craniotomy surgery, due to a combination of brain retraction, tissue dissection and removal, drug-induced pressure changes, and the changing influence of gravity. These issues have been addressed by others through the use of intra-operative MRI systems [91], and through the integration of intra-operative ultrasound [53] and laser-scanning of the cortical surface during surgery [73].

While frame-based stereotactic procedures have been considered by some to have been superceded by “frameless” approaches, both techniques are in fact part of the spectrum of image-guided surgical techniques. Originally frames were discarded in favor of the frameless approach because the application of the frame was itself considered overly invasive. Nevertheless, frameless techniques generally require a craniotomy. An equivalent minimally-invasive approach may require a frame, but the procedure itself would be performed through a single burr-hole in the skull. The trauma of frame fixation is often outweighed by the removal of the risk associated with a craniotomy. In focussing on minimally invasive techniques, much of the concern about intra-operative tissue movement is removed, but the need to monitor the procedure intra-operatively must still be addressed.

Use of Atlases

Some minimally invasive surgical procedures require lesions to be made at target points which cannot be seen directly on the MRI or CT image. The only means of ensuring that the lesion is made in the correct position, is to perform electro-physiological measurements prior to creating the lesion. To aid in the placement of the stimulating or recording electrodes, surgeons have traditionally relied on standard atlases of the thalamus and globus pallidus. These atlases have been published by several authors [92-96], and are the basis of the standard practice that involves the manually correlation of the atlas structures with the position of a stimulating or recording electrode. Since the atlases are constructed on the basis of sections made from an individual cadaver brain, they obviously can not exactly match the brain on an individual patient. The practice in the past has been to manually scale individual cross-sectional images from the atlas to match gross features seen in the diagnostic images. Some systems allow standard atlas images to be integrated directly with the 3-D MRI or CT data sets using piece-wise linear transforms [95,96].

Recently St. Jean *et al.* [4] reported the use of a three-dimensional atlas that could be automatically warped to fit the 3-D MRI of an individual patient. They created a three-dimensional version of the Schaltenbrand and Wahren atlas which they accurately registered to a “standard” 3-D MRI brain dataset. This was subsequently matched, using a non-linear spatial warping algorithm [97] to the patient’s MR images. Using the non-linear spatial transform calculated by this operation, the three-dimensional atlas was subsequently mapped to the patient. Using their system, target structures within the atlas that define the lesion site may be visualized in three dimensions along with anatomical landmarks defined by the 3-D MRI, a model of the proposed lesion and the probe used to create it. While such atlases offers a great deal of assistance to the surgeon, it is still necessary to verify that the lesion is being created in exactly the right place through physiological stimulation and recording. Mis-

Figure 8.
Examples of interventional MR magnets
a) General Electric Signa-SP “Double-donut” interventional system. (courtesy GE Medical Systems)
Siemens “Magnetom Open” interventional magnet. (courtesy Siemens Medical Systems)

placing such a lesion by as little as one or two millimeters could have a disastrous effect on the patient’s outcome. The purpose of the atlas therefore, is not so much to guide the surgeon to exactly the correct target, but to make it easier for him to approach the desired region quickly and to verify the target position with a minimum of electro-physiological tests.

Lesioning Devices

When performing minimally invasive surgery, it is necessary to create well-defined lesions that are precisely placed at the appropriate target points. The most common means of creating such a lesion is through the use of a probe that injects a radio-frequency current into the brain tissue. This process causes a region close to the probe tip to be heated to a temperature in excess of 50º C, sufficient to destroy the tissue [98,99]. The alternative approach, is to use a cutting device called a leukotome [100], which consists of a wire loop that can be extended from the end of a probe, which cuts the tissue while the probe is rotated. Each of these two devices has its advantages and its disadvantages. While the leukotome is able to create a very well-defined lesion, and has a number of degrees of freedom in terms of building a shape that conforms to the shape of the desired target, there is a potential risk of severing blood vessels and causing bleeding. It should be noted however that this risk is present to some extent also during the introduction of either probe to the target site. While the RF Lesioning tool does not present such a risk of bleeding, the temperature gradient produced around the probe-tip creates the lesion with “fuzzy” boundaries, and may not be able to create lesions that conform to the shape of target structures.

INTERVENTIONAL IMAGING SYSTEMS

Most of the discussion to this point has concentrated on the use of a pre-operative image (CT or MRI) as the primary guidance image, perhaps in conjunction with real-time augmentation with endoscopy, microscopy of video imaging.. Since tissue morphology can change between imaging and surgery, and during the surgical procedure itself, efforts are in place to develop systems that can image the state of the tissue while the operation proceeds.

MRI

One approach towards the provision of intra-operative imaging for real-time verification of the surgical site has been the adaptation of MR imagers for use during a surgical procedure. In general, there have been three approaches to this problem:

- Special purpose horizontal-bore 0.5 Tesla superconducting magnet systems designed exclusively for dedicated use in the operating room, have an open design that allows the surgeon minimally restricted access to the patient, and which permit imaging to take place during a surgical procedure.
- General-purpose 0.2 to 1.0 Tesla permanent- or electro-magnet systems, with open access to the patient from all sides.
- Conventional self-shielded 1.5 Tesla horizontal bore systems with rapid fall-off of the fringe field.

All of these systems employ adaptations of conventional intra-operative tracking systems to follow the positions of probes and instruments in the operating field.

The first design listed above, employs a "double-donut" superconducting magnet geometry (Figure 8a) that permits access to the patient between the two toroidal magnets [101,102], with sufficient space between the magnet sections for a surgeon to perform a procedure on the patient. There are currently (2000) about twelve of these units in operation worldwide, but most of the pioneering development in their use has taken place at the Brigham and Women's Hospital in Boston [91,101].

The second approach is based on permanent or electro-magnets which may be set up with their magnetic fields in either a horizontal [103] or a vertical [104,105] configuration (Figure 8b). These systems are usually general purpose low-medium field machines, whose inherent design makes them particularly suitable for use in the operating room. Because of their design, they have almost no fringe field and are therefore much more amenable to the operating room environment than an air-cored superconducting magnet. While access is possible from all sides in the horizontal plane, the proximity of the pole-pieces restricts the access vertically. Steinmeier *et al.* [105] have reported on their initial experiences with this system in conjunction with the surgical treatment of 31 patients with supratentorial brain tumors, 18 for the guidance of trans-sphenoidal operations, and six for temporal lobe epilepsy procedures. They conclude that the intra-operative MR images obtained during these procedures provided considerable additional information that optimized the resection of tissue in these operations.

The third means of using MRI intra-operatively, is to adapt standard clinical MR imagers for operating room use by increasing the diameter of the bore and shortening the length of the magnet to improve access to the patient. At the same time the magnetic shielding must be sufficiently effective to permit the use of standard surgical tools and instruments (including x-ray image-intensfiers) in close proximity to the scanner [106]. A similar approach has been explored by Saunders and colleagues [107,108] who have developed a surgical magnet that is "parked" adjacent to the operating room, but which is moved into the OR on ceiling mounted rails as required. Since these approaches are based on clinical magnets, access to the patient is not as convenient as in the previous two examples, and imaging is performed while the patient is placed inside the magnet between stages of the surgical procedure. On the other hand, they are in all respects standard clinical machines, and as such, exhibit the image quality that one obtains in a standard diagnostic imaging system.

Most of these systems permit imaging to occur during the surgical procedure itself, in some cases in real-time using dynamic sequences. So-called MR-fluoroscopy imaging sequences can image, reconstruct and display images at better than 10 frames per second, depending on the resolution of the image. Apart from the systems that employ standard 1.5 Tesla magnets, most interventional MR magnets operate at fields from 0.2 to 1.0 Tesla, and the design compromises that must be made to ensure increased access to the patient often degrade the linearity of the gradient fields across the field of view. For this reason, the geometrical integrity and resolution of images from such systems is often not what one would expect from a conventional imaging device. Nevertheless, the ability to visualize surgical tools, the target structures, and the surrounding anatomy in real-time offers a unique opportunity for image guided surgery. Perhaps the most attractive feature of these devices is that the progress of certain therapies, e.g. cryo-surgery [109,110] and thermal ablations [111-113], can be monitored in real-time through the measurable changes in MR characteristics that are undergone by the tissues during freezing or heating. The major disadvantage of these devices, is that even in the most "open" of systems, access to the patient is awkward at best. Also, the presence of the magnetic fields associated with these systems in the operating room (particularly the "double-donut" design) creates a potentially unsafe environment where instruments must be modified for compatibility with the magnetic field.

Intra-operative MR systems have now been in evaluative clinical use for approximately five years, and there is still much debate about the efficacy of these techniques. Questions that arise include the cost-effectiveness of the technology; the effect on patient outcome; the quality of the intra-operative images and the advantages over standard neuro-navigational techniques. Consequently, the evaluation of intra-operative MRI remains the subject of intense research.

Ultrasound

Ultrasound, has been employed for many years as an interventional imaging modality [29]. It is generally conceded that the image quality available from ultrasound is inferior to that attainable from even low-field intra-operative MRI [105]. However, image-quality issues notwithstanding, there is mounting evidence that the ability of ultrasound to acquire images in real time during surgery, combined with the high quality of MR images acquired prior to the procedure, can achieve a result similar to intra-operative MRI. In order to combine intra-operative

US with MRI, the images must be acquired in such a way that they can be mapped to the pre-operative image data. In 1994, Trobaugh [114] introduced the concept of correlating intra-operative US with preoperative CT or MRI, and others have developed various techniques to either compare 2D ultrasound images to other modalities [115,53], or to accumulate a series of ultrasound images to create volumetric data sets [116,117,118]. Many techniques for acquiring and displaying 3-D ultrasound data have recently been described in the literature and many of these have direct relevance to the acquisition of intra-operative 3-D images. Given the expense of intra-operative MRI, intra-operative US is being seriously considered as an alternative means to interventional MRI for acquiring information about the state of tissue during surgery [119,120]. Besides being used to detect the changes that occur during surgery, US is also being investigated as part of a procedure to correct the pre-operative images, so that they match the intra-operative brain morphology.

Comeau *et al.* [121,122], propose the identification of homologous landmarks within MRI and Ultrasound images in order to construct a set of distortion vectors that would allow the image of the pre-operative MR image to be warped to match the intra-operative MRI. They have demonstrated [123] the clinical utility of combining ultrasound images with pre-operative MRI's and the feasibility of deforming the pre-operative MR slices to match the intra-operative US images (Figure 9). Their current work uses tracked 2-D ultrasound data acquisition, but they propose to extend this work to embrace the use of true 3-D ultrasound image acquisition. While this approach has some disadvantages with respect to intra-operative MRI (the ultrasound transducer must be in contact with the brain tissue during imaging, necessitating a craniotomy), most of the procedures for which intra-operative MRI is employed also involve open craniotomy procedures. The presence of an ultrasound transducer and its associated tracking mechanisms does present an intrusion into the operating field, but no more so than a surgical probe being employed to permit guidance during the procedure.

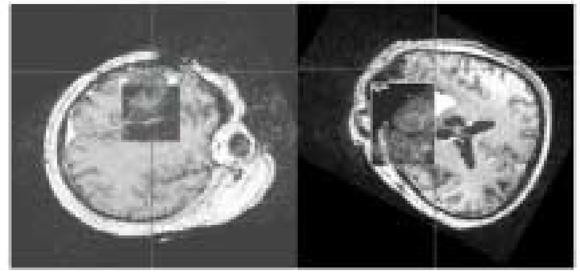

Figure 9a,b.
Two examples of intra-operative tracked ultrasound images, merged with pre-operative MRI, for the detection of tissue shift during open-craniotomy neurosurgery.

NON-NEUROSURGICAL APPLICATIONS

By far the majority of image-guided surgical procedures have to date been performed in the brain. However, there is increasing activity for image-guided surgery in other parts of the body as well.

Breast

Using the GE "Double-donut" interventional magnet discussed above, Gould et al. [124] have demonstrated that interventional MRI is an effective tool for accurately identifying palpable breast tumors and guiding their surgical excision. Also, the use of radiographic and ultrasonic techniques for image-guided breast surgery is increasing, but is currently restricted to fine needle aspiration and core biopsy, as discussed by Staren *et al.* [125,126]. They comment that future care of patients with diseases of the breast will continue to be increasingly dependent on image-guided breast biopsy techniques. This approach should avoid many unnecessary open biopsies for benign lesions and facilitate therapeutic planning for malignant lesions. The increasing use of computer assisted stereotactic radiographic and ultrasound breast imaging is also reviewed by Burns [127], who suggests that such techniques to guide percutaneous core sampling of the abnormal area are less invasive, less painful, highly accurate, and less expensive than incisional breast biopsy with preoperative needle localization.

Prostate

Image guidance is common in treatment of prostate cancer, but its application is focussed on therapy (via cryosurgery or brachytherapy) rather than surgical removal. The most common modality in use for image-guided therapy of the prostate is ultrasound, and the recent development of 3-D trans-rectal ultrasound (3-D TRUS) imaging techniques [27,128,129] has had a positive impact on this field. When employing 3-D TRUS for therapy, a series of 2-D US images, uniformly spaced in angle, are reconstructed to form a 3-D volume. This is registered with a therapeutic probe (radio-active seed implant device or cryosurgery probe) and manipulated in much the same manner as a lesioning device is introduced into the brain. Intra-operative 3-D TRUS provides unique views of the prostate, which facilitates the optimal placement of the probes.

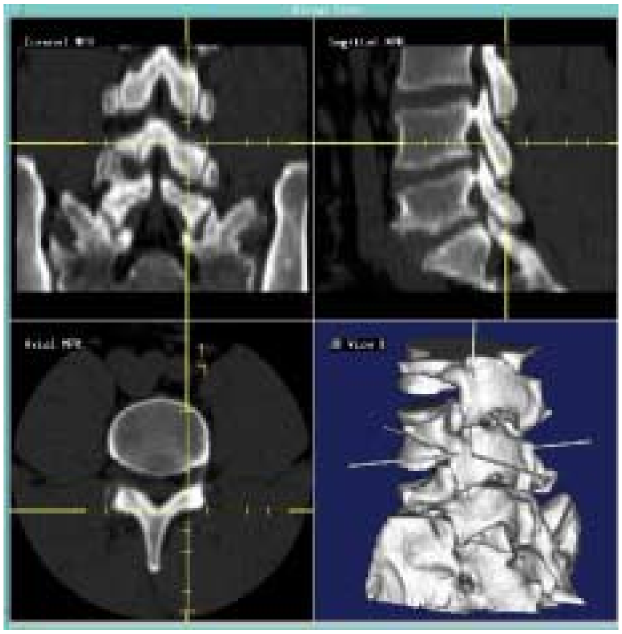

Figure 10.
Display on a commercial image-guided surgery workstation of a spinal surgery procedure. The image shows orthogonal slices through the spine, the cross-hair cursor shows the position of the pointer tip on each slice, while the 3-D view illustrates the position of the pointer with respect to the spinal process due to be operated.

Spinal and Orthopedic applications

Computerized image-guidance of spinal and orthopedic procedures has lagged somewhat behind its application in neurosurgery, but it is rapidly growing in popularity. Lavalee *et al.* [130] discuss a computer-assisted surgical system for navigation during spinal surgery, using CT images as guidance (Figure 10). Their approach follows closely that outlined for IGNS. This same group [131] describes a similar approach for anterior cruciate ligament reconstruction. This system uses a workstation and a three-dimensional optical localizer to create images that represent knee kinematics. In the future, it is expected that standard arthroscopy will be combined with a tracking system and "virtual arthroscopy" based on high quality MR and/or CT images of the joint to assist in such procedures. Foley *et al.* [132] point out the limitations of conventional 2-D imaging techniques for spinal navigation, and describe a three-dimensional image-space representation of the surgical volume, using a specially designed referencing system and computer workstation, while Nolte [133] discusses the use of interactive navigation of surgical instruments for the fixation of spinal implants.

The use of rapid 3-D MRI was investigated by Martel *et al.*, [134] to acquire images suitable for image guided surgery of the spine. They employed a fast 3-D MRI sequence (117 secs for a 256 x 256 x 96 point image volume), with a wide bandwidth and short echo time (TE) to minimize susceptibility distortions, along with MRI/CT compatible fiducial landmarks. This permitted the validation of the MR approach against CT, and demonstrated that the registration can be undertaken with an accuracy of 0.4 mm using 3-D MRI. They demonstrated that MR was effectively as accurate as CT for spinal imaging in this context, and conclude that MRI shows promise for use in computer assisted surgery of the spine.

OPERATING ROOM ENVIRONMENT

Given the huge range of technology available for image-guided surgery, it is easy to lose sight of the fact that it must all fit physically, and be usable, in the operating room. Figure 11 shows a view of the surgical team employing a tracked probe, during an IGNS procedure. One of the first points to note is how far the image monitor is located from the surgeon. This is typical, and is dictated by the bulk of the equipment, and the (usually) confined space in the OR. Unfortunately, this does not provide the ideal ergonomic environment for the surgeon who must constantly move his head, and re-accommodate his vision, each time he looks between the monitor and the surgical field. This situation can be improved by the provision of a flat-panel display that could be placed close to the operating field or through the use of head-mounted displays. The tracking system shown in Figure 11 is optical, and though it is light-weight, it suffers from "line-of-sight" constraints, and will cease to operate when the LED emitters lose sight of the tracking device. Even tighter constraints apply to the use of interventional MRI systems in the OR. In addition, software on typical systems tends to be inflexible in that it constrains the user to specific built-in functionality. In the future systems must be developed with sufficient flexibility to adapt to the requirements of the surgeon.

FUTURE DIRECTIONS

Perhaps the greatest challenge to the successful adoption of image-guided surgery systems in the operating room of the future, relates not to technology, but to the user interface. Over the last two years, affordable high-speed computing hardware has become available. Such hardware (e.g a 500MHz Pentium III-based computer, equipped with a sophisticated but inexpensive video processor) now allows interactive surgical image-guidance to be performed in real time (images updated in fractions of a second), with the

Figure 11.
Typical operating room image-guided neurosurgery environment.

images presented to the surgeon stereoscopically. It is interesting to note that most of the improvements in computer and visualization hardware are currently being driven by the consumer video gaming industry! It is nevertheless unfortunate that to date little effort has been expended to ensure that the human factors issues relating to the use of such equipment in the OR have been adequately addressed. The key will be to design these systems so they are unobtrusive and simple to operate in the OR without the need for a computer specialist on hand. The logistics of data management (acquisition of images, merging them with scans from other sources, segmenting relevant structures) must be handled either automatically or with a minimum of intervention. It must ultimately be possible to operate computer systems in the OR without the need to use the keyboard, or complicated switching devices. The IGS system must not add complexity to the OR environment.

It might be argued that IGNS will have a significant impact on "difficult" procedures that could not be performed satisfactorily, if at all, without computerized image-guidance. However, it is probable it will have the greatest overall effect when it becomes a standard operating-room tool for routine procedures, reducing OR time, decreasing patient trauma and streamlining data management. A well designed universal multi-dimensional data structure (such as the MINC approach described by Neelin [135]) will allow images from multiple sources to be integrated and manipulated effortlessly. The effective integration of this technology with hospital information systems will play a significant role in its widespread acceptance.

The psychophysical issues of manipulating objects within a virtual 3-D environment have barely been addressed to date. Today's technology provides us with a wealth of tools (VR displays, high-speed interaction, haptic feedback, robotics etc), but without a detailed understanding of how the human visual/sensory system interacts with these tools, their advantages may be lost to us.

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