Invited Paper

A Comprehensive Approach to Image-guided Surgery.

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ABSTRACT

Image-guided surgery has evolved over the past 15 years from stereotactic planning, where the surgeon planned approaches to intracranial targets on the basis of 2-D images presented on a simple workstation, to the use of sophisticated multi-modality three-dimensional image integration in the operating room, with guidance being provided by mechanically, optically or electro-magnetically tracked probes or microscopes. In addition, sophisticated procedures such as thalamotomies and pallidotomies to relieve the symptoms of Parkinson's disease, are performed with the aid of volumetric atlases integrated with the 3-D image data. Operations that are performed stereotactically, that is to say via a small burr-hole in the skull, are able to assume that the information contained in the pre-operative imaging study, accurately represents the brain morphology during the surgical procedure. On the other hand, performing a procedure via an open craniotomy presents a problem. Not only does tissue shift when the operation begins, even the act of opening the skull can cause significant shift of the brain tissue due to the relief of intra-cranial pressure, or the effect of drugs. Means of tracking and correcting such shifts form an important part of the work in the field of image-guided surgery today. One approach has been through the development of intra-operative MRI imaging systems. We describe an alternative approach which integrates intra-operative ultrasound with pre-operative MRI to track such changes in tissue morphology.

Keywords: Image-guided surgery, brain atlases, ultrasound, MRI, 3-D imaging, stereoscopic visualization, image distortion.

1. INTRODUCTION.

The surgeons at the Montreal Neurological Institute (MNI) have been using image-guided neuro-surgery (IGNS) techniques in the operating room for many years. The standard tasks performed by the principal IGNS system, (the ISG Viewing-Wand[i]) are the visualization of 2-D and 3-D MR images of a patient, registered with a computer tracked probe as a navigational device during surgery. While this and other commercial devices perform well as basic IGNS tools, they do not currently support the integration of images from multiple image sources, the introduction of brain atlases, stereoscopic presentation of 3-D images or the ability to integrate a real-time imaging source into the surgical imaging environment. For this reason we have developed our own platforms to complement the Viewing Wand. VIPER (Visual Integration Platform for Enhanced Reality) provides the means to achieve the first of these three objectives, while real-time intra-operative imaging via Ultrasound/MRI integration is currently handled by a separate system. Both interface to the tracking device that is part of the ISG Viewing Wand, with which they can be used concurrently in the operating room.

The core of VIPER contains the same basic functionality as most standard IGNS systems, i.e. the ability to manipulate and display image-slices selected from a volume, while navigating through the volume with a computer-tracked pointing device. MRI scans provide anatomical information relating to the patient's brain, while certain critical structures, such as the nuclei of the thalamus, cannot be visualized at all on standard MR scans. Moreover, standard MR images do not display functional characteristics, so the use of Positron Emission Tomography, (PET), Magnetic Resonance Spectroscopic Imaging (MRSI), functional Magnetic Resonance Imaging (fMRI) and other imaging modalities must be recruited to provide the appropriate functional context. VIPER was designed to address these issues by explicitly allowing the simultaneous display of multiple pre-registered data-sets, and the integration within the patient image volume of standard neuro-anatomical atlases. The surgical requirement for 3-D realism when navigating amongst three-dimensional structures is met by the provision of a stereoscopic display mode.

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[i] ISG Technologies Inc, Mississuaga, ON, Canada

2. MULTI-MODALITY IMAGE VISUALIZATION.

Functional imaging modalities usually present very poor spatial resolution, (on the order of 3-6mm), and generally do not contain anatomical information to assist in their interpretation. This context is usually provided by MRI or CT images to which the functional images have been appropriately registered. On the other hand, even MRI and CT scans do not always provide enough anatomical insight in some regions of the brain, either because of low contrast or limited resolution. This is the case for IGNS needs for thalamotomy where the specific nodes that must be identified within the thalamus cannot be differentiated on either MRI or CT. In these cases it is necessary to combine atlases with the anatomical images.

The merging of multi-modality images in VIPER is performed according to the philosophy of adding complementary information (functional, chemical, taxonomic...) to a primary structural scan (MRI or CT). VIPER classifies the volumes to be loaded into two categories, namely a primary volume to which up to seven secondary volumes can be attached. A volumetric scan consists of a 3-D array of intensity values which can be displayed on the screen in numerous ways. The most standard is to assign a gray-scale mapping with the displayed image contrast controlled by standard "window level" and "window-width" functions. This method is adequate when displaying only MRI or CT images, however, when additional imaging modalities are incorporated, we employ up to six additional color-maps, so that a number of different volumetric scans may be superimposed and visualized simultaneously. An opacity value between zero and one, assigned to each image, controls the merging of the different images that must appear simultaneously. For instance, if both PET and MRSI scans must be displayed on top of an MRI data-set, the PET scan is merged with the MRI according to the PET opacity value, then the MRSI image is merged with this composite image. In addition to the window, level and color-map, the opacity of the slices for every individual scan can be adjusted as well. An example of a multi-modality image display within VIPER is shown in Figure 1, where choline and lactate signals are displayed from MRIS, and sensory-motor activation images are displayed by PET. MRSI was crucial in this situation for separating tumor from edema, and PET allowed critical brain tissue to be avoided during the procedure.

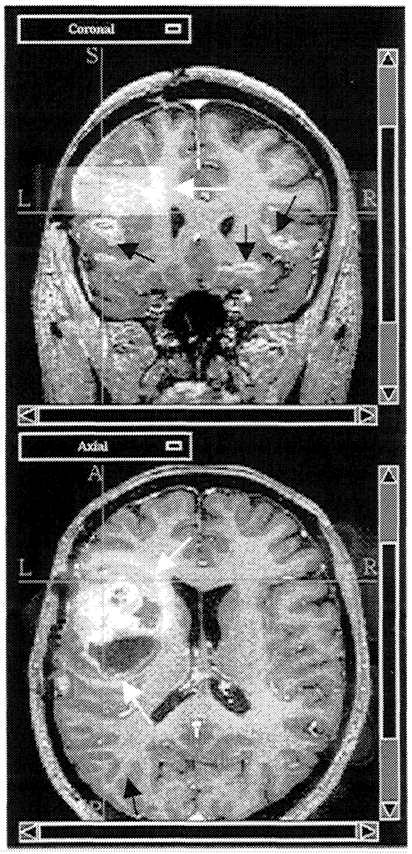

Figure 1. Coronal and Sagittal MR slices with superimposed MRS and PET images. Normally The PET and MRSI images are displayed using distinct color scales. Here the MRSI signals are identified by white arrows and PET by black arrows.

3. THALAMOTOMY AND PALLIDOTOMY

Symptomatic Parkinson's syndrome manifests itself in a variety of ways, including tremors, rigidity, bradykinesia, impaired gait, dysphasia and ocular disturbance. Most of these symptoms can be relieved by the prescription of L–dopa, except for tremors, which are more resistant to this drug. Tremors result from "rhythmic alternating contraction of opposing muscle groups usually of the distal extremities".[1] In the case of Parkinson's syndrome, the tremors are known to come from disorders in the extrapyramidal motor system. Parkinsonian patients suffering from significant tremors, and those who have gained only a mild improvement (or no improvement at all) of their general state from L–dopa, are good candidates for thalamotomy and pallidotomy. At the MNI, pallidotomies are usually performed when the patient suffers from rigidity rather than tremors. Ideally, candidates for thalamotomy and pallidotomy should be under 60 years of age, with unilateral tremors or rigidity. The probability of almost complete relief of symptoms (without side-effects) for these candidates is around 80%[2]. Currently at the MNI, the specific target for thalamotomy is the nucleus ventralis intermedius (Vim) (Fig. 2), while the target for pallidotomy is part of the globus pallidus mediale (Gpm) and sometimes a small part of the globus pallidus laterale (Gpl). This operation requires the use of a stereotactic frame, which serves as a geometrical frame of reference, as well as a support for surgical instruments, attached to the patient's head.

Unfortunately, it is difficult to identify precisely the position of the Vim and Gpm with conventional imaging tools, such as MRI, CT or X-ray ventriculography. Even though the mid-plane and antero-posterior commissural (AC-PC) plane can easily be localized on ventriculograms, the distances at which the target structure lies from these planes change from one patient to the other, mostly because of the variable size of the third ventricle.

In order to address this problem, physiological verification of the position of the target with respect to the stereotactic frame coordinate system is performed prior to the excision itself. The position of the target is confirmed through electrical stimulation of the basal ganglia, and an electrode is inserted into the patient's brain through the same burr–hole used for the injection of the ventriculography contrast agent.

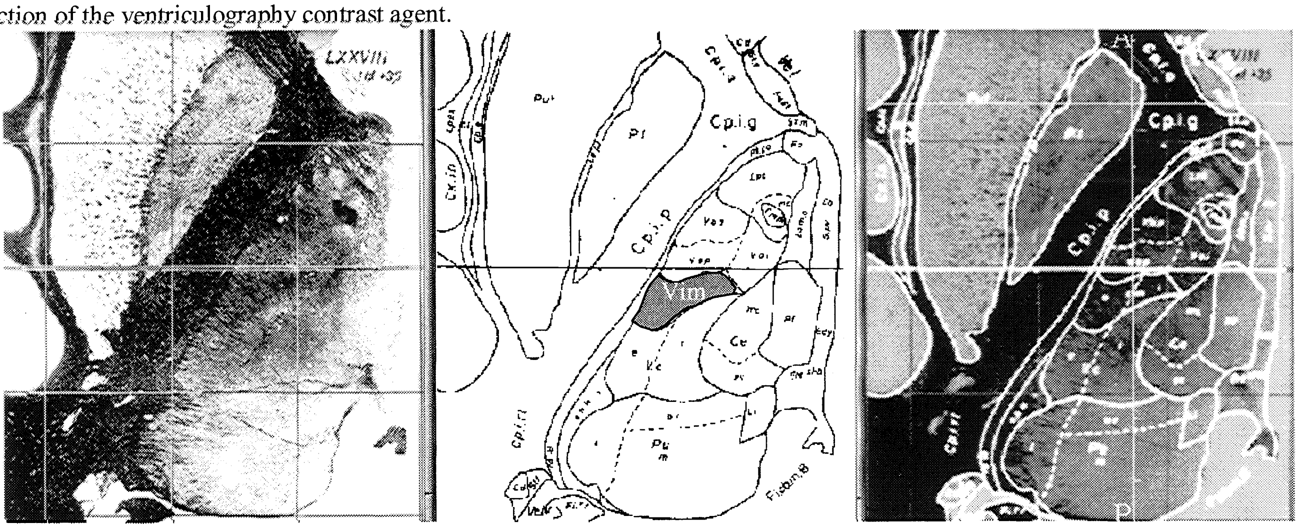

Figure 2. The anatomical slice of the thalamus (left), the derived contour altas (Shultanbrand and Bailey, centre) and the superposition of the two (right). Vim nucleus is shown shaded in the centre panel.

The neurosurgeon stimulates various regions of the deep brain structures, while at the same time noting the sensory, motor or visual responses from the patient. Motor responses come from stimulation of the internal capsule, an important white matter structure that must be spared from the excision. The internal capsule defines the border of the Vim on the lateral side, and of the Gpm on the medial side. Sensory responses, which come from Vci and Vce nuclei in the thalamus, and which both touch the posterior part of the Vim node, should be spared as well. The information obtained with the electrical stimulation allows the neuro-surgeon to determine the position of the target relative to the frame coordinate system with acceptable precision. Once the surgeon has obtained enough information about the position of the important deep brain structures of the patient, he can proceed with the therapeutic lesion through excision of the Vim or the Gpm.

4. ATLASES

The Schaltenbrand and Bailey atlas is one of the principal anatomical references for neurosurgeons performing thalamotomies and pallidotomies. This data-set consists of a series of micro-thin cryogenic slices that were stained to discriminate between structures presenting different cytoarchitectures. The slices are not equi-spaced, since each is separated from its neighbor by a distance varying between 0.5mm to 3mm. Our task consisted of creating an easily deformable volumetric atlas from the original data consisting of two-dimensional slices with varying inter-slice separations. This was achieved by creating a number of smooth-surfaced 3-D structures representing atlas components, using Hermite polynomials, to fit the atlas data[3,4]. Similar volumetric atlas construction has also been reported by Novinski *et.al.*[5]

4.1 Model MRI data set: Superbrain

The benefit of having a computerized volumetric version of some structures defined in the Schaltenbrand atlas is still somewhat limited from a neuro-surgical point of view. Anatomical variability from one patient to another prevents a simple overlap of a properly oriented version of the atlas onto the MRI. We must not only rotate, translate and scale the atlas, but also deform it locally (in a non-linear fashion) so that it can be adequately superimposed onto the patient's MRI with the different structures appearing in their correct places. Our objective was to have a nonlinear deformation that could performed on a routine basis without operator intervention. We achieved this goal through the application of the ANIMAL[ii] algorithm, a 3-D non-linear warping algorithm that uses a multi-scale cross-correlation approach to warp one 3-D MRI data-set onto another[6,7]. It first computes the best global linear transformation, and then defines the local deformations at finer and finer scalings. This kind of non-linear deformation corresponds to our needs, except that the algorithm requires MRI images as both the original and target volumes.

To use this algorithm to warp the atlas onto the patient's MRI, we must first go through an intermediate step. Thin-plate spline 3D interpolation[8] allows us to perform a smooth non-linear deformation from any volumetric data set to another, if we can identify a sufficiently large set of homologous points in the two volumes. The strategy then is to identify manually the corresponding landmarks on the atlas and within a model MRI data set one time only, so that the atlas may then be warped onto the model MRI. Subsequently, the non-linear transformation from the model MRI to any patient's MRI may be automatically computed using the ANIMAL algorithm. Since the atlas is already registered the MRI model space (from the manual registration), this last transformation can be applied to the atlas in order to register it to the patient's MRI.

The appropriate choice of the model MRI volume is fundamental to the success of this procedure, since it affects both the quality of the manual tagging of the atlas, and the performance of the ANIMAL algorithm. One of the most important criteria is the signal to noise ratio (SNR) in the image. Our model, which we call “Superbrain”, was created from an average of 27 T1-weighted 3-D MRI scans of the same individual[9]. Superbrain reveals an SNR about 5.1 times higher than that obtained with a standard volumetric acquisition under this same protocol (Fig.3).

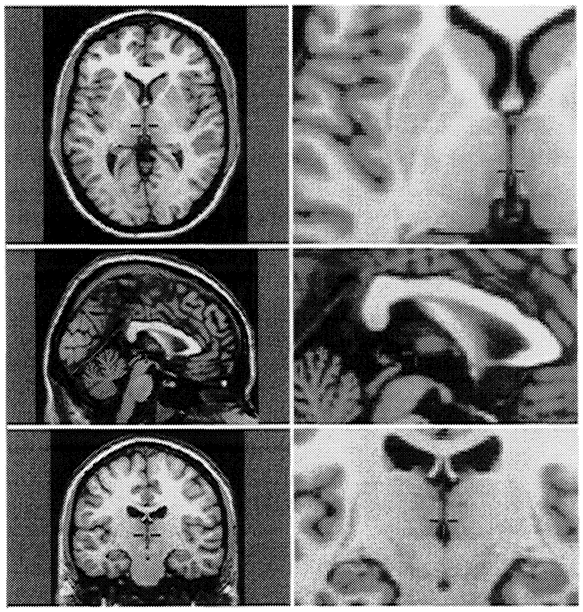

Figure 3. Orthogonal sections through “Superbrain” (left) and an enlargement of the thalamic regions (right).

The manual tagging of the atlas to the model MRI was performed by an experienced neuroanatomist, using a total of 250 homologous points (Fig. 4). The thin-plate spline interpolation forces all 250 points from the atlas to be mapped onto their corresponding points on the model MRI, while the regions in between the points are smoothly warped into the new space.

4.2 Patient space

In order to be useful in the IGNS context, both the atlas and the patient's MRI must be navigable in a coordinate system understood by the neurosurgeon. Thus they must be registered with respect to the actual physical position of the neuro-surgical tools. These tools are rigidly attached to the stereotactic frame that is fastened to the patient's skull prior to the operation. Rulers drawn on this stereotactic frame define the coordinate system that the neurosurgeon refers to when localizing the target. The atlas and the MRI of the patient's brain are transformed into frame space through the identification in the MR images of Z shaped reference markers attached to the frame.

[ii] ANIMAL – Automatic non-linear imaging matching and labelling

4.3 Imaging Tools for Thalamotomy and Pallidotomy.

As described above, VIPER employs the concept of primary and secondary volumes when dealing with multi-modality visualization. The primary volume (MRI or CT) is drawn on the screen first, followed by the secondary volumes which are “blended” on top of the original image one by one. For thalamotomy-related guidance, secondary volumes consist of atlases and virtual lesions.

The type of visual information that is available in the 2D display panels is often a limiting factor. If only slices through the volume rather than complete 3D structures are displayed, the neurosurgeon must mentally extrapolate the extent of these structures outside of the slice, which is often a difficult task. It is particularly hard to reconstruct the overall shape, size and orientation of a region of the anatomy or of a lesion site in the brain. It is also difficult to assemble different neighboring parts together in space and to imagine how they are related to one another.

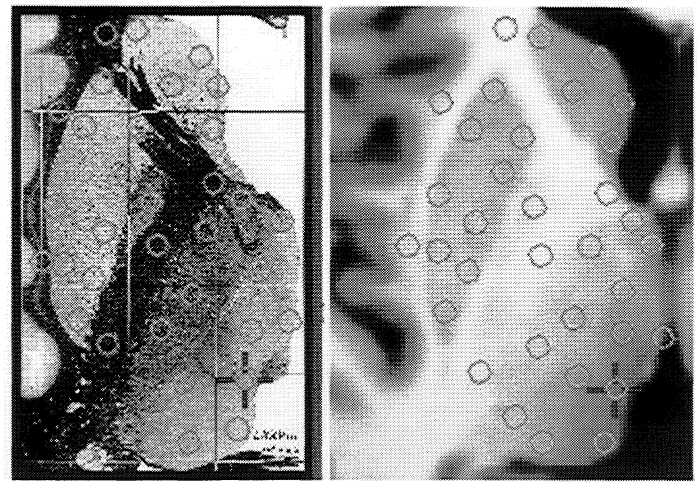

Figure 4. The tagging procedure to match the Atlas to “Superbrain”. Homologous points within the two volumes are indicated by the circles.

During a thalamotomy, two neuro-surgical tools (the electrical stimulator and the leukotome – a knife-like instrument for deep brain lesioning – see Fig. 5) are inserted deep within the patient's brain. Since no craniotomy is performed during this operation, the neurosurgeon never sees directly what is happening with the tools inside the brain. The only on-line verification is currently via radiographs (ventriculograms) that are taken during the operation. The neurosurgeon must mentally reconstruct both the geometry of the anatomy and the position of the neuro-surgical tools, which must be manipulated to create the lesion in the desired place. It is a complicated mental task to follow its position with respect to the critical structures that must be either avoided or excised. In VIPER, the geometry of the anatomy is already displayed either on 2D slices or in the 3D window as surface-rendered objects. It is also possible to display the position and geometry of the neuro-surgical tools with respect to the thalamic nodes, on the 2D slices as well as in the 3D window.

The position of the leukotome with respect to the co-registered atlas and MRI data set is fully controllable by the user. The position of the target point, (the point where the tip of the leukotome should be) and the two angles of entry (declination and azimuth) must be specified first. During the surgical procedure, the excision is performed by rotating the leukotome around its shaft axis. The neurosurgeon must be able to determine the path of the leukotome through the anatomy when performing the excision. This is difficult to show in the 2-D images, so perhaps the greatest advantage of the 3-D mode in VIPER is that the volume of interest can be inspected from arbitrary points of view via rotation, translation and zooming, (Fig. 5).

5. STEREOSCOPIC VIEWING

In the 3-D window we try to present structural features such as depth, geometry and the relative position of objects, attributes that cannot be appreciated on 2-D slices. However, the computer screen is still a 2-D plane, onto which we are limited to displaying projections of 3-D objects. The user can gain more insight into the 3-D geometry of the objects by interactively moving them in space, but is still limited to viewing a series of 2-D pictures. One of the strongest depth-perception cues, stereopsis, relies specifically on human binocular vision. We have incorporated our previous work with stereoscopic visualization[10,11,12] into VIPER, providing it with a fully interactive stereoscopic viewing mode. Stereo images are created in the standard manner by computing separate left and right-eye views of 3-D images, and viewing the images using standard active shutter devices[iii]. The program may be switched back to standard mode at will. In the stereoscopic mode, the user can still interact with the 3D objects, including the leukatome and the stimulator, (Fig. 5).

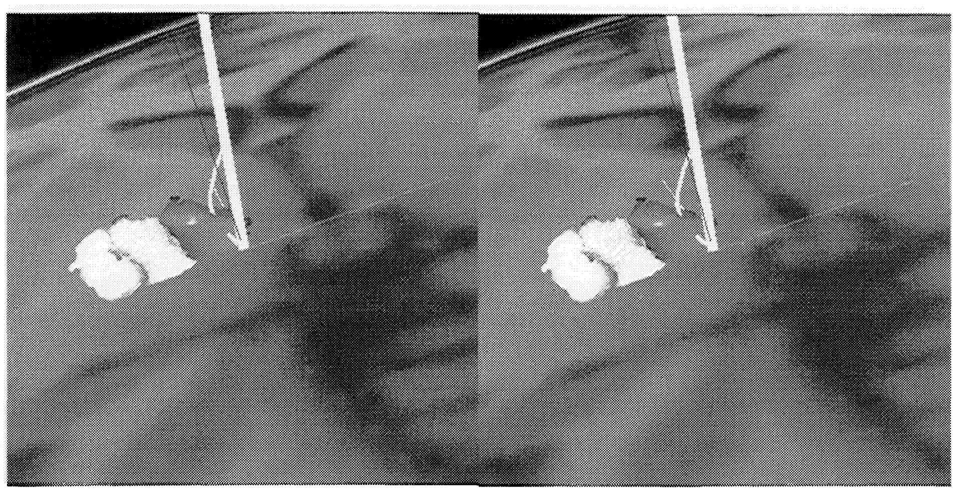

Figure 5. Stereoscopic view (set for cross-eye viewing) of nuclei within the 3-D atlas volume integrated with a texture map of MRI intensity values, and leukatome surgical tool.

6. TISSUE SHIFT DURING SURGERY

In the procedures described above, the stereotactic frame provides a common co-ordinate system where targets of interest can be defined in the image during the planning phase of an operation, and accurately reached using tools attached to the frame during the procedure itself. However, "frameless stereotaxy" permits processing and display of anatomical and functional preoperative image on the computer screen, based on the position and orientation of a pointing device, during open-craniotomy procedures. One factor affecting the overall accuracy of these systems is brain tissue deformation during the surgical procedure. By definition, an ideal IGNS system reports the exact position of a surgical probe on the preoperative images for the duration of the procedure. However, in reality, the overall position error of the frame-less navigation system is dependent on many factors, which can be attributed to failures of two basic assumptions:

1) That the equipment, registration and images are perfectly accurate, i.e. the pointer tracking device is free of positioning error, the registration between the patient and image spaces is error free, and that the images themselves are free from spatial distortion.

2) That the coordinate frames of the equipment and volume of interest are rigidly connected. This implies that, for the duration of the procedure, the base of the tracking device remains rigid with respect to the patient skull, that the structures of interest within brain remain in the same position with respect to each other, and to the external fiducial points used for patient-image registration.

6.1 Target Volume Rigidity

During the operation, particularly during open craniotomies, the skull and brain are invaded, causing the brain tissue to distort with respect to the skull and the external registration points used to map the patient to the images. Distortions may be due to drug-induced intra-cerebral pressure changes, impact of surgical instruments, and resection. IGNS systems based solely on preoperative information, cannot account for such movements and therefore may erroneously display the position of the pointing device with respect to the internal brain structures.

[iii] CrystalEyes, Stereographics Corp, San Raphael, California.

6.2 Intra-operative Ultrasound

In order to detect brain shift during surgery, an intra-operative imaging technique is required. Several systems using "open" MRI magnets, which allow MR images to be acquired in real time during surgery, are currently employed in operating rooms[15]. Here we propose an alternative approach using intra-operative Ultrasound (US) which has the advantages of being simple to setup, inexpensive, has a track record in qualitative surgical guidance and is readily available[16,17]. Because the skull attenuates the US beam, US images can only be acquired after craniotomy, or through a large burr hole.

Our ultrasound based IGNS system consists of a PowerMacintosh workstation[iv] with a built-in video frame grabber, interfaced via the video input to an ATL Ultramark 9 ultrasound system[v] with a P7-4 multi frequency (4-7 MHz) phased array transducer. The transducer is tracked by interfacing it to a six degree-of-freedom tracking arm[vi], which measures its position and orientation and reports these parameters to the computer workstation via a serial port.

After a calibration procedure, the position and orientation of the US image are known in both the patient and image coordinate systems. The image can therefore be considered to be rigidly attached to the transducer itself and to sweep through space as the transducer is moved.

6.3 System errors

Our approach combines the high-quality imaging capabilities of MRI, with the real-time imaging ability of US. For MRI (the most common image modality used in IGNS), image distortion is highly dependent on acquisition parameters and is the subject of investigation in our laboratory[13]. Typically MR image distortion can be in the order of 2mm if no precautions are taken to avoid it.[14] US images are formed by measuring the echo times of sound waves emitted by a transducer, and converting them to distances using the sound propagation speed in that tissue. The speed of sound in tissue is generally accepted to be 1540 m/sec while the average speed of sound in brain tissue is 1510 m/sec[7]. If the ultrasound machine assumes an average velocity of sound in tissue of 1540 m/sec, there will be a 2% error in the direction of the sound beam (direction of increasing depth), representing a 1mm error at 5cm depth. This error may be reduced by re-calibrating the US system for a sound velocity of 1510 m/sec. During a neuro-surgical procedure, the patient's head is rigidly fixed to the operating table using a clamp that attaches directly to the skull using compressed pins. The base of the tracking device is attached directly to the head fixation clamp forming a rigid system consisting of the patient skull, the fixation device and the tracker.

6.4 Clinical Application

After extensive testing in phantoms[18], this IGNS system was evaluated clinically by using it during a neurosurgical procedure where intra-operative US had already been indicated. In this case, US guidance was required to monitor the IV ventricle and brain stem in a patient with syringobulbia and syringomyelia in the upper cervical spine. A syringo-subarachnoid shunt was inserted, and the collapse of the syrinx was monitored by ultrasound.

The US transducer was placed in a sterile drape and attached to the Faro tracker using the sterilized adapter. The intra-cranial cavity was filled with saline solution to form a good acoustic interface between the transducer and the patient. Images were acquired and examined at several points during the course of the operation.

During this procedure, the MRI-US IGNS system proved useful in several ways. First, it assisted the surgeon in establishing the orientation of the US image being displayed, by providing an oblique MRI view that coincided with the live US image. The superior anatomical MRI display assisted the surgeon in the interpretation of the corresponding US images. In addition, the US-MRI display enabled the surgeon to evaluate the patient-MR image registration by delineating structures on the MR image and observing where they lay on the live US images, as well as using an MRI-US overlay tool to visualize the discrepancies. This was particularly useful here, since the surgeon needed to determine whether the upper cervical spine had moved between the MRI scan and the operation. The IGNS system did in fact show a discrepancy between the intra-operative US and the preoperative MRI. Figure 6 demonstrates a deformation and lateral shift of the fourth ventricle, which may have been due to movement of the upper cervical spine after the MRI, error in the patient-image registration or a shift in the internal structures during surgery. Here, the surgeon was able to immediately appreciate the discrepancy and consider it in evaluating the overall usefulness of the IGNS system for this case.

[iv] Apple Computer Inc, Cupertino, CA.

[v] ATL Laboratories Inc, Bothwell, WA.

[vi] Faro Medical Technologies, Lake Mary, FL.

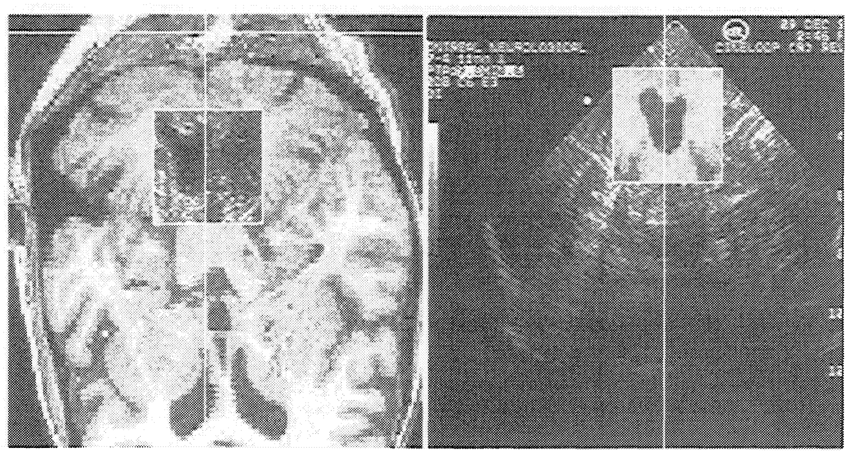

Figure 6. Slice through pre-operative MR image (left) corresponding to the live ultrasound image (right). Note the ROI windows on each image corresponding to information from the other modality.

7. CONCLUSIONS

We have presented a number of enhancements to standard image-guided neuro-surgery procedures, which we believe will become integral components of such systems in the future. We have demonstrated that inexpensive technological solutions exist to dramatically enhance the information that can be delivered to the surgeon in the operating room. Future developments with MRI-US integration will allow pre-operative MR images to be warped to match intra-operative US. However, if such systems are going to be commonplace in the future, there are many logistical and human factors issues still to be addressed. These systems must be seamlessly integrated with radiology and hospital information systems, and their user interfaces must be enhanced so that surgeons may interact with these tools with the same facility with which they use their traditional surgical instruments.

8. ACKNOWLEDGEMENTS.

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9. REFERENCES

1. A. E. Walker. *Classification of movement disorders,* chapter 31, pp 503-509. Georg Thieme Verlag, 1982.
2. *ibid.* chapter 33, pages 515-521. Georg Thieme Verlag, 1982.

3. St-Jean P, Clonda D, Sadikot A, and Peters TM. Computer guidance for thalamotomy and pallidotomy. *Medical Physics* 24, 1205, July 1997.

4. Clonda D, St-Jean P, Sadikot A, Evans AC, Peters TM. Automatic labeling of the patient's brain with a sub-cortical atlas in image-guided neurosurgery. *Medical Physics* 24, 1211, July 1997.

5. W. L. Nowinski, A. Fang, B. T. Nguyen, J. K. Raphel, L. Jagannathan, R. Raghavan, R. N. Bryan, G. A. Miller.
6. Multiple brain atlas database and atlas-based neuroimaging system. *Computer Aided Surgery*. 2; 42-66. 1997.

7. D. L. Collins, W. Dai, T. S. Marrett, and A. C. Evans. Three-dimensional non-linear warping of a computerized volume-of-interest brain atlas for morphometric analysis. In *Information Processing in Medical Imaging,* Wye, UK, July 1991.

8. D. L Collins, T. M. Peters, and A. C. Evans. An automated 3D non-linear image deformation procedure for determination of gross morphometric variability in human brain. In *Proceedings of Third Conference on Visualization in Biomedical Computing, volume 2359*, pages 180--190. **SPIE**, Oct. 4-7 1994.

9. Fred L. Bookstein. Principal warps: Thin-plate splines and the decomposition of deformations. *IEEE Transactions on Pattern Analysis and Machine Intelligence,* ***PAMI-11***(6):567--585, 1989.

10. C.J. Holmes, R. Hoge, D.L. Collins, and A.C. Evans. Posthoc enhancement of MR signal using automatic registration. In: *2nd International Conference on Functional Mapping of the Human Brain*, page 240, Boston, June 1996.

11. Bruce Davey, Roch Comeau, Chris Gabe, Andre Olivier, and Terence Peters. Interactive stereoscopic image-guided neurosurgery. *In Proceedings of SPIE Medical Imaging*, *volume 2164*, pages 167--176. **SPIE**, 1994.

12. Bruce Davey, Patrice Munger, Roch Comeau, Laura Pisani, Andre Olivier, and Terry Peters. Applying stereoscopic image visualization to neurosurgical image guidance. In: Anthony M. DiGioia, Takeo Kanade, and Russell Taylor, editors, *Proceedings of the First International Symposium on Medical Robotics and Computer Assisted Surgery*, pages 264--271, Pittsburgh, Penn., September 22-24 1994.

13. Bruce L. K. Davey, Roch M. Comeau, Patrice Munger, Laura J. Pisani, Daniel Lacerte, Andre Olivier, and Terence M. Peters. Multi-modality interactive stereoscopic image-guided neuro-surgery. In Richard A. Robb, editor*, Proceedings of Third Conference on Visualization in Biomedical Computing*, volume 2359, pages 526--536. **SPIE**, Oct. 4-7 1994.

14. Munger P, "Accuracy considerations in MR image-guided neuro-surgery", Masters thesis, Dept. of Physics, McGill University, 1994.

15. Sumanaweera T, Glover G, Song S, Adler J, Napel S. Quantifying MRI geometrical distortion in Tissue. *Magnetic Resonance in Medicine* 31:40-47, 1994.

16. Moriarty TM, Kikinis R, Jolesz FA, Black PM, Alexander E III. Magnetic resonance imaging therapy. Intra-operative MR imaging. *Neurosurgery Clinics of North America.* vol 7(2) pp 323-331, 1996.

17. Tong S, Downey DB, Cardinal HN and Fenster A. A three-dimensional ultrasound prostate imaging system. *Ultrasound Med Biol*, vol 22, pp 735-746, 1996.

18. Trobaugh JW, Richard WD, Smith KR and Bucholz RD. Frameless Stereotactic Ultrasonography: Methods and Applications, *Computerized Medical Imaging and Graphics*, vol 18:4, pp 235-246, 1994.

19. R. M. Comeau, A Fenster, and T. M. Peters. Intra-operative Ultrasound Imaging in Interactive Image Guided Neurosurgery. *Radiographics*, in press, 1998.