Intraoperative US in Interactive Image-guided Neurosurgery[1]

Roch M. Comeau, MSc • Aaron Fenster, PhD • Terry M. Peters, PhD

Tissue movement can be a significant source of error in image-guided neurosurgery. A surgical guidance system that incorporates preoperative image information (eg, magnetic resonance [MR] imaging data) and intraoperative ultrasound (US) allows detection of tissue deformation during neurosurgery. An interactive image overlay tool allows a region of interest (ROI) to be defined and permits the operator to move the ROI over the MR or US image and overlay the associated image (US or MR) within the ROI. The system can be validated with a deformable multimodality phantom. Before deformation of the phantom, good agreement between the MR imaging and US data overlay confirms proper registration of the MR and US images; after deformation, the overlay demonstrates significant distortion of ventricles and movement of simulated blood vessels. Intraoperatively, this information helps establish the orientation of the US image being displayed by providing an oblique MR image that coincides with the live US view. The superior anatomic display of MR imaging also helps the surgeon interpret the corresponding US images. Finally, the system enables the surgeon to evaluate the patient–MR image registration by comparing structures on the MR images and live US images and using the overlay tool to visualize discrepancies.

■ INTRODUCTION

Many implementations of interactive image-guided neurosurgery systems involve use of preoperative image information (eg, magnetic resonance [MR] imaging or computed tomographic [CT] data) as a guide to the surgeons throughout the procedure. However, tissue movement during an operation can be a significant source of error in image-guided neurosurgery. This article presents a surgical guidance system that incorporates preoperative image information (eg, MR imaging or CT data) and intraoperative ultrasound (US) to detect deformation of brain tissue during neurosurgery. We begin by discussing the effect of brain tissue deformation on image-guided neurosurgery. We then describe a multimodality validation phantom and demonstrate use of the phantom in verifying the ability to detect tissue deformation with the US-based image-guided neurosurgery system by means of constant reference to the undistorted MR

Abbreviation: ROI = region of interest

Index terms: Brain, surgery, 10.45 • Computers, examination control, 10.12987 • Stereotaxis, 10.1267 • Ultrasound (US), guidance, 10.12986 • Ultrasound (US), intraoperative, 10.12982

RadioGraphics 1998; 18:1019-1027

[1] From the McConnell Brain Imaging Center, Montreal Neurological Institute, WB3B (MRI), 3801 University St, Montreal, Quebec, Canada H3A 2B4 (R.M.C., T.M.P.); the Department of Biomedical Engineering, McGill University, Montreal (R.M.C., T.M.P.); and the Robarts Research Institute, University of Western Ontario, London, Canada (A.F., T.M.P.). Presented as an *info*RAD exhibit at the 1996 RSNA scientific assembly. Received September 2, 1997; revision requested November 10 and received December 31; accepted February 19, 1998. Supported by Collaborative Research Grant CPG 0164606 from the National Sciences and Engineering Research Council of Canada and Operating Grant MT 11540 from the Medical Research Council of Canada. **Address reprint requests to** R.M.C.

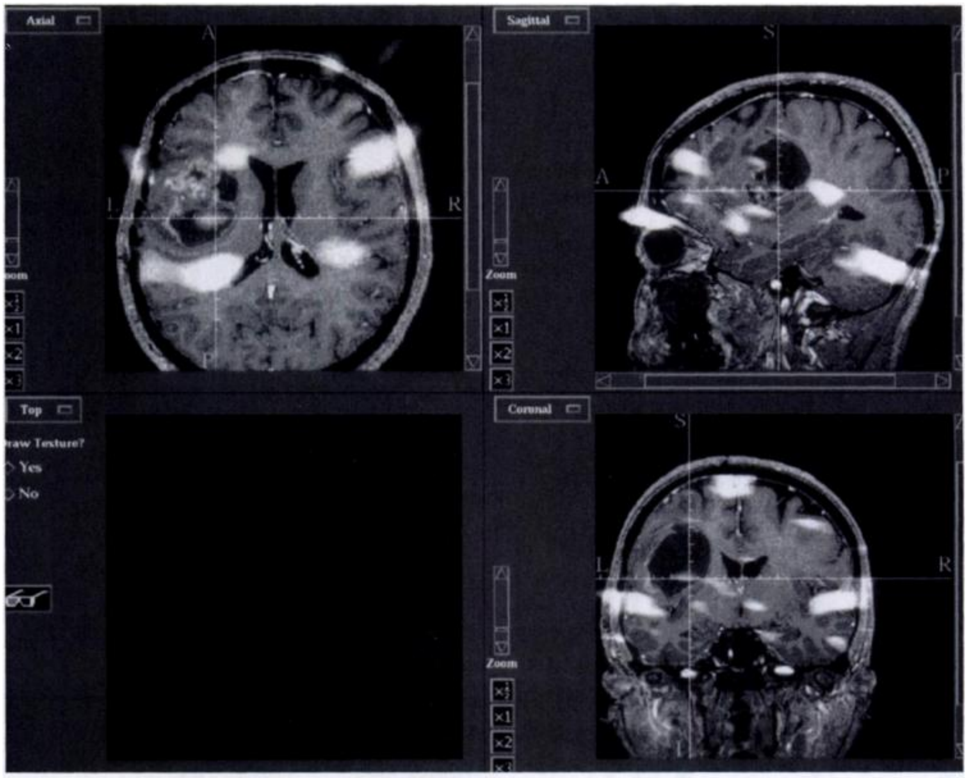

Figure 1. Images obtained with a multimodality image-guided neurosurgery system show data from a functional positron emission tomographic activation study superimposed on anatomic MR images used during lesion resection.

imaging representation of the phantom. Finally, we present an example of use of this combined MR imaging–US system during a neurosurgical procedure.

■ IMAGE-GUIDED NEUROSURGERY

Soon after the advent of radiography, the usefulness of radiographs as a guidance tool for neurosurgery became obvious. The stereotactic frame was subsequently developed (1) and has been used for many years to help the surgeon relate structures observed on a radiograph to the head of a patient. The frame provides a common coordinate system in which targets of interest can be defined on the image during the planning phase of an operation and accurately reached during the procedure by using tools attached to the frame.

A recent improvement has been the introduction of "frameless stereotactic" (ie, interactive image-guided neurosurgery) systems (2,3). Such tools permit processing and display of anatomic and functional preoperative image information (eg, from MR imaging, CT, MR angiography, positron emission tomography, or functional MR imaging) on the computer screen on the basis of the position and orientation of a pointing device tracked by the computer. The tracking device and display software map the operating room coordinate system (the patient) to the image coordinate system by means of a registration procedure. This process allows the surgeon to use the pointer as an interactive navigational tool during surgery.

Figure 1 shows images obtained with an image-guided neurosurgery system during a surgical procedure. The information displayed consists of anatomic MR imaging data overlaid with data from a positron emission tomographic activation study. This anatomic and functional information permits the surgeon to navigate to, and resect, a lesion near the sensorimotor area of the brain.

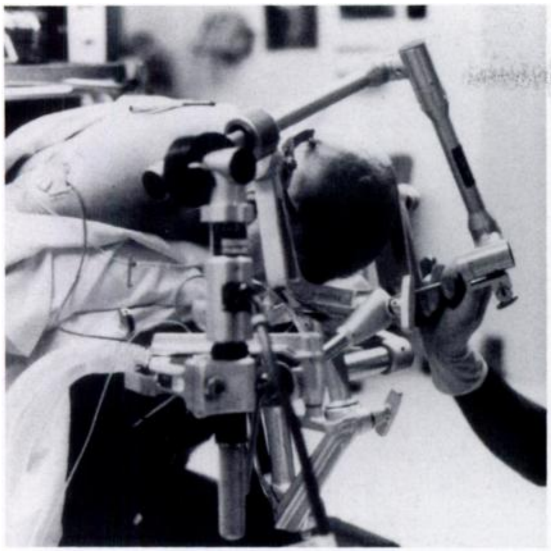

Figure 2. Photograph shows the patient's head fixed to the operating table with a Mayfield clamp.

One factor that affects the overall accuracy of interactive image-guided neurosurgery systems is brain tissue deformation during surgery. We discuss this issue later in the article and present an intraoperative US-based multimodality image-guided neurosurgery system that can help the surgeon assess the extent of this problem during surgery and hence improve the overall accuracy of the surgical navigation.

• Possible Sources of Error in Image-guided Neurosurgery

By definition, an ideal image-guided neurosurgery system reports the exact position of a surgical probe on the preoperative images for the duration of the procedure. In reality, however, the overall position error of the frameless navigation system depends on many factors and can be attributed to the failure of two basic assumptions: (*a*) that the equipment, registration, and images are perfectly accurate (ie, that the pointer tracking device is free of positioning error, the registration between the patient and image spaces is error free, and the images themselves are free from spatial distortion) and (*b*) that the coordinate frames of the equipment and the volume of interest are rigidly connected. According to the second assumption, the base of the tracking device remains rigid with respect to the patient's skull and the structures of interest within the brain remain in the same positions with respect to each other and to the external fiducial points used for patient-image registration for the duration of the procedure.

Tracking Device Error.—Commercially available tracking devices can provide position and orientation information with submillimeter position accuracy (4). Simple homologous point matching with a least squares distance minimization technique can yield patient-image coordinate mapping with an accuracy on the order of 2-3 mm.

Image Distortion.—For MR imaging (the most common image modality used in image-guided neurosurgery), image distortion is highly dependent on acquisition parameters and is the subject of investigation in our laboratory (5). Typically, MR image distortion is on the order of 2 mm if no precautions are taken to avoid it (6). US images are formed by measuring the echo times of sound waves emitted by a transducer and converting the echo times to distances by using the sound propagation speed in that tissue. The speed of sound in tissue is generally accepted to be 1,540 m/sec (7), whereas the average speed of sound in the brain is 1,510 m/sec (7). If the US machine is calibrated to the average velocity of sound in tissue of 1,540 m/sec, there will be a 2% error in the direction of the sound beam (the direction of increasing depth). The result would be an error of 1 mm at a depth of 5 cm. This error may be reduced by recalibrating the US system for a sound velocity of 1,510 m/sec.

Rigidity of Patient and Tracking Device.—During a neurosurgical procedure, the patient's head is rigidly fixed to the operating table by means of a clamp that is attached directly to the skull with compressed pins (Fig 2). The base of the tracking device is attached directly to the head fixation clamp. The result is a rigid system that consists of the patient's skull, the fixation device, and the tracking device.

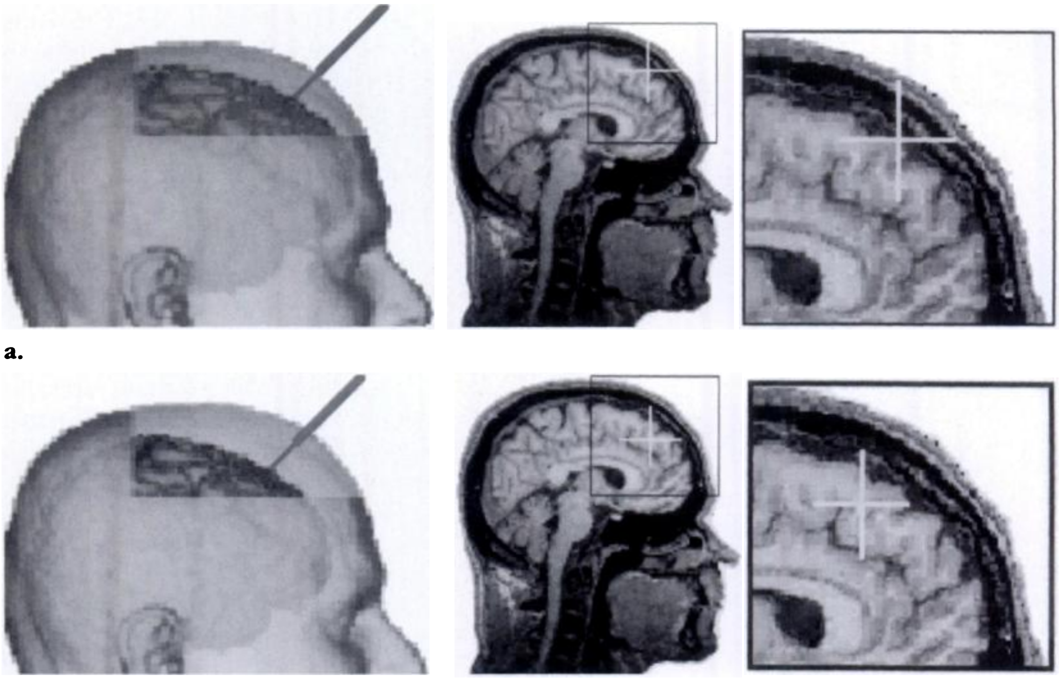

Figure 3. **(a)** Illustration shows that, after craniotomy and with the assumption that there is no shift in the brain tissue or any static error, the position of the pointer is accurately represented on the preoperative image. **(b)** Illustration shows that, if there is some shift or distortion in the tissue, the system is not able to account for such changes because the basic assumption of rigidity is violated. In this example, the brain has compressed, requiring the pointer to move further into the head to encounter the cortical surface than the image would indicate. This situation is erroneously reported by the computer as the probe having actually entered the cortex.

Rigidity of Target Volume.—During the operation, particularly during open craniotomy, the skull and brain tissue within are invaded. Such invasion causes 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 intracerebral pressure changes, the presence of surgical instruments, and resection. Image-guided neurosurgery systems based solely on preoperative information cannot account for such movements and therefore may display the position of the pointing device erroneously with respect to the internal brain structures (Fig 3).

• Intraoperative US

To detect brain shift during surgery, an intraoperative imaging technique is required. Several systems that use open MR imagers are currently employed in operating rooms. Such systems allow MR images to be acquired in real time during surgery (8). We propose an alternative approach that uses intraoperative US, which has the advantages of simple setup, low cost, a track record in qualitative surgical guidance (9,10), and easy availability. Because the skull attenuates the ultrasound beam, US images can be acquired only after craniotomy or through a large burr hole.

■ US-BASED IMAGE-GUIDED NEUROSURGERY SYSTEM

Our US-based image-guided neurosurgery system consists of a PowerMacintosh workstation (Apple Computer, Cupertino, Calif) with a built-in video frame grabber and an Ultramark 9 US system (Advanced Technology Laboratories, Bothell, Wash) with a P7-4 multifrequency (4–7-MHz) phased-array transducer. The workstation and the US system are interfaced via the video input. The transducer is tracked by interfacing it to a tracking arm with 6° of freedom (Faro Medical Technologies, Lake Mary, Fla). The tracking arm measures the position and

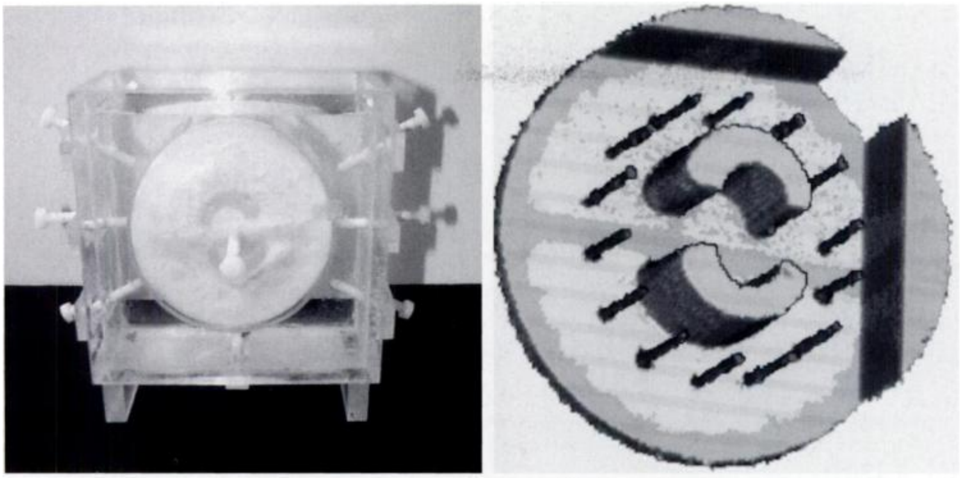

Figure 5. **(a)** Photograph shows the phantom in the container. **(b)** Three-dimensional reconstruction image from MR imaging data shows the simulated cortical surfaces, ventricles, and blood vessels.

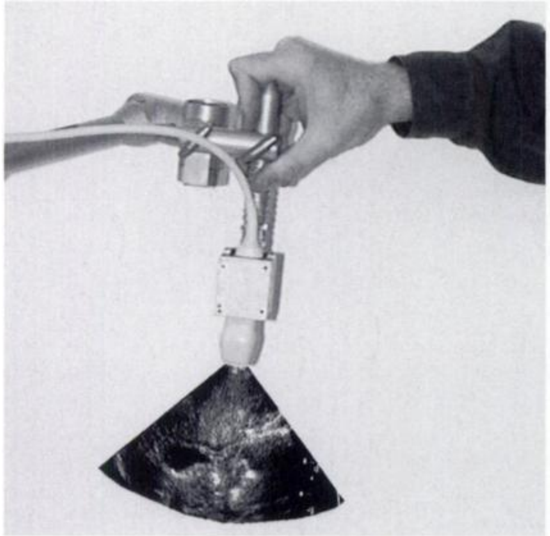

Figure 4. Photograph shows a US image conceptually "attached" to the probe after calibration.

orientation of the transducer and reports these parameters to the 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 (Fig 4) and to sweep through space as the transducer is moved.

• Validation

Deformable Phantom.—To evaluate the ability of the system to demonstrate intraoperative brain shift, a multimodality deformable phantom was constructed. It consists of a polyvinyl alcohol cryogel disk designed to simulate the brain for imaging purposes (11). The phantom has two hemispheres; ridges on the outer surface simulate the cortical surfaces, and internal fluid chambers simulate the ventricles and blood vessels (no flow). The hemispheres were recast with more polyvinyl alcohol gel to form a disk. Properties of the polyvinyl alcohol were altered between the hemispheres and the surrounding gel to provide T1- and T2-weighted MR imaging contrast and boundaries visible at US. The disk is supported by a Plexiglas collar, and both are immersed in a Plexiglas container filled with a 5% glycerin solution to match the speed of sound in tissue. Seven screws protrude into the container to support the collar and allow controlled deformation of the disk. Figure 5a shows the phantom in the container; Figure 5b is a cutaway view of a volume-rendered three-dimensional reconstruction of the disk that shows the hemispheres, ventricles, and simulated blood vessels.

Validation Experiment.—The phantom was initially set up with no deformation (representative of the preoperative situation). It was first imaged with MR imaging (T1-weighted gradient-echo sequence, 14-msec repetition time, 6-msec echo time, 23° flip angle, four signals acquired), and the resulting volume information was transferred to the image-guided neurosurgery system. The phantom was then fixed to a table along with the image-guided neurosurgery tracking arm. The image-guided neurosurgery software was used to register the internal polyvinyl alcohol disk to the MR imaging data by identifying homologous points on the MR image (with a mouse-driven cursor) and on the phantom (with the pointer attached to the tracking arm).

After registration, the US transducer was substituted for the pointer. The phantom was imaged with the US scanner, and the image orientation information was fed into the image-guided neurosurgery system. Simulated blood vessels and ventricles that intersected the image plane were visualized and were compared by means of an interactive image overlay tool. This tool allows a region of interest (ROI) to be defined and permits the operator to interactively move the ROI over the MR or US image and overlay the associated image (US or MR) within the ROI. The user may also drop markers and draw spline-based curves within either image and visualize them on the associated image.

Figure 6a shows the US and MR images of the phantom along with the ROIs but without the overlays. Figure 6b shows the two images with the associated images overlaid within the ROIs. The good agreement between the MR imaging and US data confirms that proper registration of the MR and US images is possible before deformation of the phantom.

Without moving the disk, the phantom was deformed by tightening some of the screws on the support structure. The result was a deformation from the upper right to the lower left of the phantom. The phantom was again imaged with US, and the simulated blood vessels and ventricle were again visualized with the interactive overlay tool. Figure 6c shows MR images with the ROI and without and with the US overlay. The overlay clearly demonstrates significant deformation of the ventricle and movement of the simulated blood vessels.

• Illustrative Surgical Case

The image-guided neurosurgery system was evaluated by using it during a neurosurgical procedure for which intraoperative US had already been requested. In this case, US guidance was required to monitor the fourth ventricle and brain stem in a patient with syringobulbia and syringomyelia in the upper cervical spine. A syringosubarachnoid shunt was created, and the collapse of the syrinx was monitored with US.

The US transducer was placed in a sterile drape and attached to the tracking arm with the sterilized adapter. The intracranial 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 operation.

The image-guided neurosurgery 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 MR image that coincided with the live US view (Fig 7a). The superior anatomic display of MR imaging helped the surgeon interpret the corresponding US images. Second, the

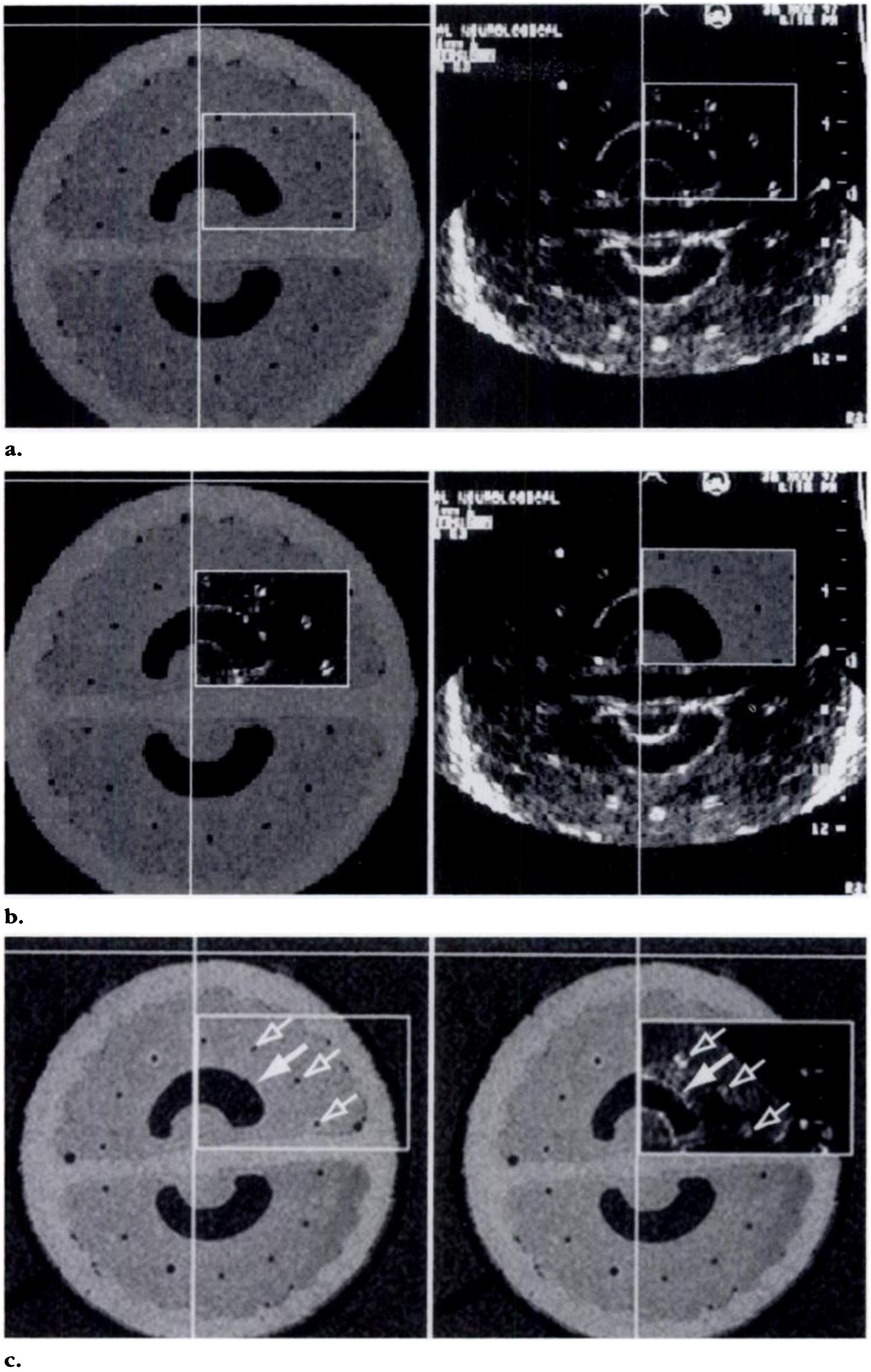

Figure 6. **(a)** MR (left) and US (right) images show the simulated ventricle and blood vessels in the upper right quadrant highlighted with the ROI to facilitate comparison. **(b)** MR (left) and US (right) images with ROI overlays (US data on MR image and vice versa) show agreement between MR imaging and US before distortion of the phantom. **(c)** MR images without (left) and with (right) the US overlay show discrepancies in the shape of the ventricle (solid arrow) and in the positions of simulated blood vessels (open arrows) after distortion of the phantom.

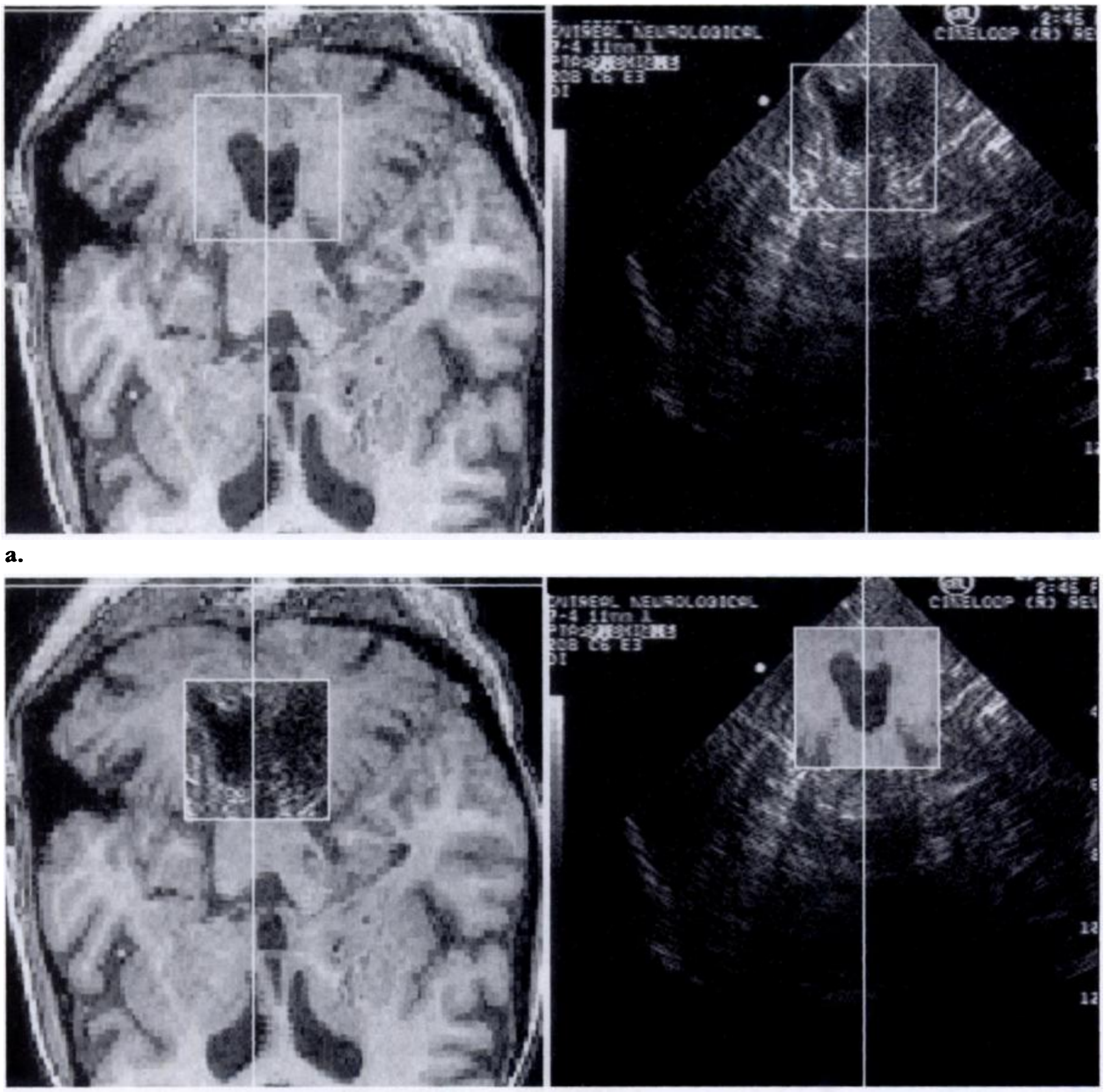

Figure 7. **(a)** Preoperative MR (left) and intraoperative US (right) images show the ROI with no overlay and a trajectory centerline for reference. **(b)** Preoperative MR (left) and intraoperative US (right) images show the US image superimposed within the ROI on the MR image and vice versa. The ROI and centerline permit visualization of distortion and lateral shift of the fourth ventricle.

US-MR imaging display enabled the surgeon to evaluate the patient-MR image registration by identifying structures on the MR image and observing where they occurred on the live US images and by using the MR imaging-US overlay tool to visualize any discrepancies. Such evaluation was particularly useful in this case because the surgeon wanted to determine whether the upper cervical spine had moved between the MR imaging study and the operation. The image-guided neurosurgery system did demonstrate a discrepancy between the intraoperative US images and the preoperative MR images. Figure 7b shows deformation and lateral shift of the fourth ventricle. This discrepancy may have been due to movement of the upper cervical spine after MR imaging, an error in the patient-image registration, or a shift in the internal structures during surgery. The surgeon was able to appreciate the discrepancy immediately and consider it in evaluating the overall usefulness of the image-guided neurosurgery system in this case.

■ CONCLUSIONS

A multimodality interactive image-guided neurosurgery system can map intraoperative US information to preoperative MR imaging or CT data. The system can be validated with a deformable multimodality phantom that can be imaged with both MR imaging and US. This intraoperative information is valuable to the surgeon in monitoring the progress of the surgical procedure and in evaluating the effects of registration error and tissue movement on the overall accuracy of the image-guided neurosurgery procedure. In the future, we hope to use features extracted from both the preoperative and intraoperative images to perform nonlinear warping of the preoperative images for correction of intraoperative tissue movement, further increasing the value of such images in the operating room.

Acknowledgments: We gratefully acknowledge the support of Abbas Sadikot, MD, PhD, and André Olivier, MD, PhD, for their insight and access to the operating room, as well as the support from the technical staff at the Montreal Neurological Institute and Robarts Research Institute. We also thank Advanced Technology Laboratories for supplying the US probe.

■ REFERENCES

1. Picard C, Olivier A, Bertrand G. The first human stereotaxic apparatus: the contribution of Aubrey Mussen to the field of stereotaxis. J Neurosurg 1983; 59:673-676.
2. Peters TM, Davey B, Munger P, Comeau R, Evans A, Olivier A. Three-dimensional multimodal image-guidance for neurosurgery. IEEE Trans Med Imaging 1996; 15:121-128.
3. Maciunas RJ, Galloway RL Jr, Fitzpatrick JM, Mandava VR, Edwards CA, Allen GS. A universal system for interactive image-directed neurosurgery. Stereotact Funct Neurosurg 1992; 58:108-113.
4. Rohling R, Munger P, Hollerbach JM, Peters T. Comparison of relative accuracy between a mechanical and a position tracker for image-guided neurosurgery. J Image Guided Surg 1995; 1:30-34.
5. Munger P. Accuracy considerations in MR image-guided neurosurgery. Thesis. Department of Physics, McGill University, Montreal, Quebec, Canada, 1994.
6. Sumanaweera T, Glover G, Song S, Adler J, Napel S. Quantifying MRI geometrical distortion in tissue. Magn Reson Med 1994; 31:40-47.
7. Kremkay FW. Diagnostic ultrasound: principles and instruments. 4th ed. Philadelphia, Pa: Saunders, 1993; 18.
8. Moriarty TM, Kikinis R, Jolesz FA, Black PM, Alexander E III. Magnetic resonance imaging therapy: intraoperative MR imaging. Neurosurg Clin North Am 1996; 7:323-331.
9. Tong S, Downey DB, Cardinal HN, Fenster A. A three-dimensional ultrasound prostate imaging system. Ultrasound Med Biol 1996; 22:735-746.
10. Trobaugh JW, Richard WD, Smith KR, Bucholz RD. Frameless stereotactic ultrasonography: methods and applications. Comput Med Imaging Graph 1994; 18:235-246.
11. Chu KC, Rutt BK. Polyvinyl alcohol cryogel: an ideal phantom material for MR studies of arterial flow and elasticity. Magn Reson Med 1997; 37:314-319.