REAL-TIME IMAGE WARPING FOR INTRA-OPERATIVE USE

David G. Gobbi[1,2] and Terry M. Peters[1,2]
[1]Imaging Research Laboratories, The John P. Robarts Research Institute,
London, ON N6A 5K8, Canada, and
[2]Department of Medical Biophysics, University of Western Ontario, London, Canada

INTRODUCTION

The current generation of image-guided neurosurgery systems use rigid-body transformations to register the pre-operative 3D MRI scan of the patient to the location of the patient's head once the head has been clamped to the operating table. These systems are unable to compensate for deformation of the brain, although it has been demonstrated that the brain deforms by an average of 10 mm [1] following a craniotomy (the removal of a skull flap). This error can result in incomplete resection of tumours, and also limits the applicability of image guidance for surgery near critical areas of the brain because serious harm is done if these areas are inadvertently damaged.

The potential registration accuracy of image-guided surgery is on the order of 1 mm, but in order for this accuracy to be attained for craniotomy procedures a nonlinear "warp" transformation must be applied to the MRI image to match it with the current shape of the brain. The two approaches used to measure the deformation that will be applied to the MRI are mechanical modelling of the brain [2] and intra-operative imaging [3]. Our current focus is on the use of intra-operative 3D ultrasound to measure brain deformation. We expect to see a convergence of the imaging and modelling approaches in the future as research in this area progresses.

We have developed our own prototype image-guided surgery system (displayed in Figure 1) which provides the sagittal, coronal and axial views of the MRI as well as volume rendering. We acquire 3D ultrasound image volumes through a freehand technique, and measure the deformation through identification of homologous landmarks in the intra-operative ultrasound volume and the pre-operative MRI volume. As additional landmarks are identified, the views of the deformed MRI that are displayed on the computer screen are updated in real-time.

METHODS

Our experiments were performed on a poly vinyl alcohol cryogel (PVA-C) deformable phantom that we manufactured in our laboratory [4]. This phantom was scanned on a 1.5T clinical MRI scanner, and afterwards a 3D ultrasound scan was performed in our laboratory using a neurosurgical ultrasound probe. The orientation and position of the probe are recorded by a POLARIS optical tracking system via a set of infrared light-emitting diodes.

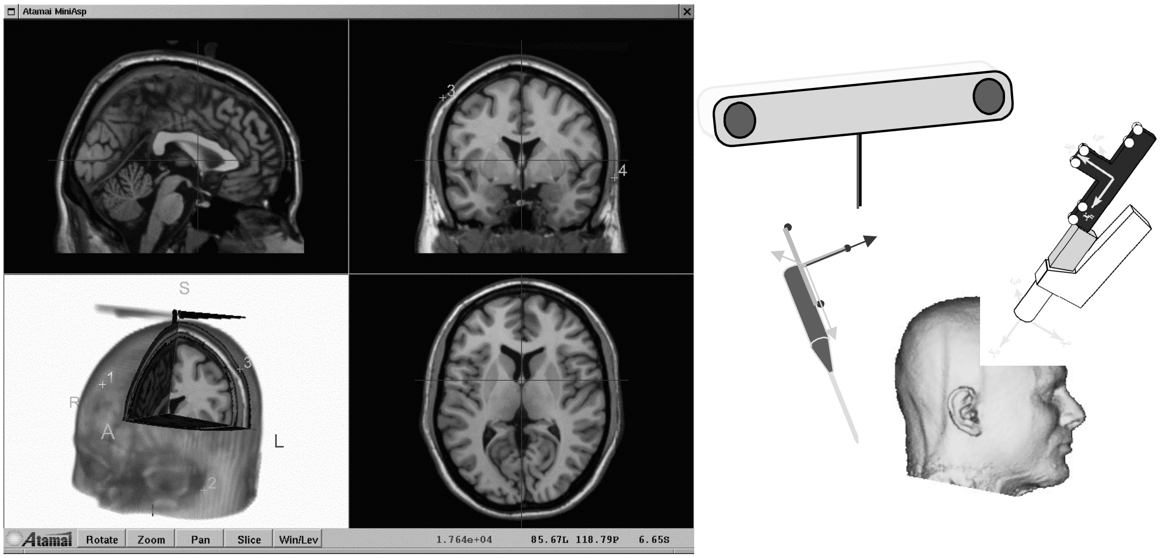

Figure 1: Our image-guided surgery platform. The surgical pointers and ultrasound probes are tracked with a POLARIS optical tracking system (Northern Digital Inc., Waterloo, ON, Canada) with an accuracy of 0.4 mm.

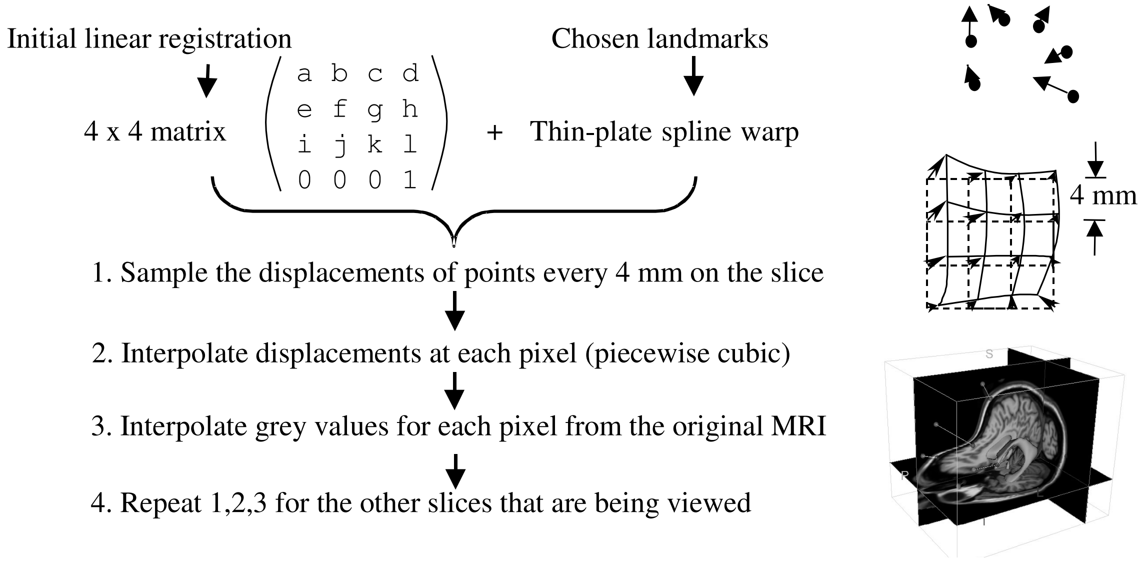

Figure 2: Flowchart for a method of displaying slices of a warped volume in real time. The pixel intensity values for each viewed slice are computed by an interpolated lookup into the original MRI volume via a piecewise cubic approximation of the warp transformation. If a new slice is chosen for one of the three views, steps 1 through 3 are repeated. Every time the warp transformation is modified through the addition of a new homologous point pair, steps 1 through 3 are repeated for of each of the three views.

Our prototype image-guided surgery system has been built using the Visualization Toolkit [5] (VTK), a free C++ library for 3D data analysis and visualization. All of our C++ code related to nonlinear transformations and image warping has been contributed to the VTK copyright holders, and is available for download for commercial and non-commercial use as part of the VTK 3.2 distribution (URL: ftp://public.kitware.com/pub/vtk/vtk3.2/). The nonlinear transform types that are currently supported include thin-plate spline transformations as well as transformations described by a regular grid of displacement vectors. A full description of the implementation and special features of the transformations will be provided in an upcoming publication [6].

The application of a nonlinear transformation to a 256 × 256 × 120 slice MRI volume typically requires several minutes on a modern computer (450 MHz Pentium-II CPU). It is not necessary to apply the transformation to the entire volume, however, because only a few slices of the volume are viewed at one time. We have developed a technique whereby only the computations necessary to display three orthogonal slices of the warped volume are performed. This subset of the warp computations provides the same user experience as warp of the entire volume (because image-guided surgery systems display only three slices of the brain at one time) but allows updates of the warp to be viewed in a fraction of a second per viewed slice. The algorithm that we use is outlined in Figure 2.

RESULTS AND DISCUSSION

Figure 3 demonstrates, for our deformable phantom, the nonlinear registration of an MRI volume to a 3D ultrasound volume. To perform the registration, we dropped circular markers on landmarks the MRI image with the mouse, and then used the mouse to drag each marker to the corresponding landmark in the ultrasound image. The markers are the small spheres arranged in a half-circle in the figure. Each time the position of a marker is adjusted, the warped view MRI automatically updated within 1.5 s. The registration that resulted from applying a thin-plate spline based on the landmarks was very good (within 1.0 mm) in the portions of the volume that lie near to the landmarks. However, near the bottom the arc-shaped cavity on the left side of the image there was a registration error of 1.5 mm because no nearby landmarks were selected with the mouse.

Our future goals include the development of a suitable protocol for the acquisition of clinical 3D ultrasound images and identification of landmarks such that the error on the registration of the MRI to the ultrasound, within a volume of interest, is known to be less than 1.0 mm. We will also evaluate other transformations that are related to the thin-plate spline to determine whether they are better suited to this application. To assist in the evaluation of different protocols and algorithms, we have created PVA-cryogel phantoms from MRI scans of a human brain that we will be using in future studies.

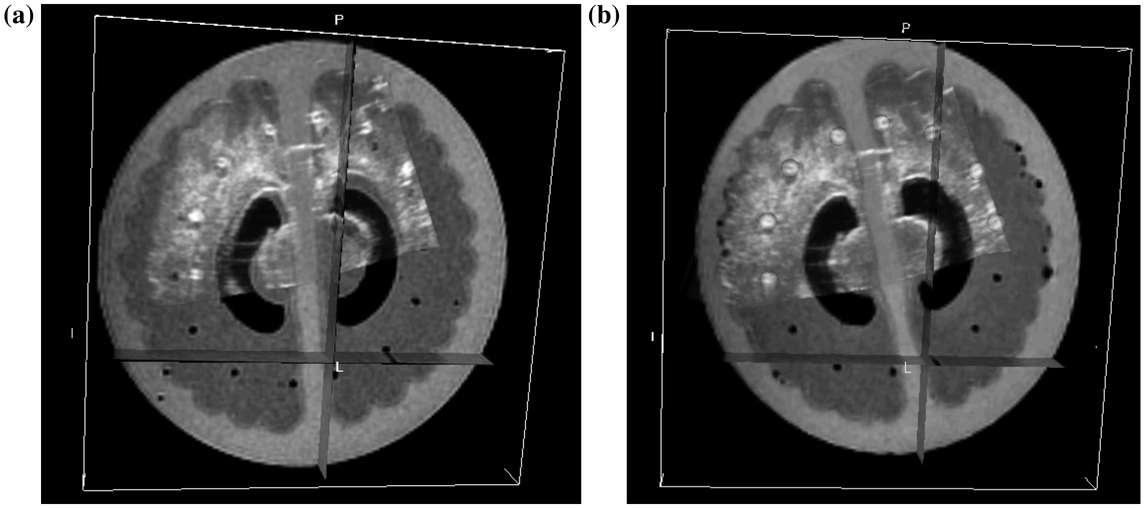

Figure 3: The overlay of the 3D ultrasound volume on the MR volume (a) before thin-plate spline registration and (b) after registration. The circles in the second image are the thin-plate spline control points. The mouse is used to drag the control points to different positions in 3D space.

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