Abstract – We consider the problem of multimodality image fusion, with emphasis on planning, navigation and control in neurosurgery. Given that images to be fused must be registered, we explore the issue of image registration in neurosurgery. While rigid registration is appropriate for registering images that are either all pre-op(ervative) or post-op, nonrigid registration is necessary to register, say, pre-op and intra-op images. This is a consequence of the “brain-shift” occurring when the skull is opened for surgery and of issue retraction and resection, e.g., for tumor removal. We describe our finite-element-based nonrigid registration and fusion systems. Our approach to fusion is inspired from the human visual system; it was initially developed for remote sensing.

Keywords: Medical imaging, neurosurgery, neuro-navigation, brain-shift, segmentation, SPM, ITK, level sets, surface mesh, volume mesh, finite elements, nonrigid registration, multimodality image fusion.

1 Introduction

Neurosurgical planning, navigation and control increasingly rely on multiple 3D imaging modalities such as CT, structural MRI (sMRI), functional MRI (fMRI), PET, SPECT, and US. Such images are taken in various combinations before, during and after surgery. They are respectively referred to as pre-op(ervative), intra-op and post-op images. Surgeons are thus potentially faced with a large number of types of images, such as pre-op PET and intra-op MRI.

Given the number and variety of potentially available image types, the question of “fusing” these images into a smaller number of image products naturally arises. We carefully distinguish between “registration” and “fusion.” Registration refers to the 3D voxel-wise (or 2D pixel-wise) alignment of images in space. Registered images can then readily be compared. Fusion implies the combination of these images into a single new (fused) image, which is also registered with the individual images.

It is a rare instance in medical imaging that different imaging modalities, even taken with the same scanner, would be registered right from the start. Thus, some form of registration is usually called for. Registration techniques for medical imagery (as well as for other applications) are generally divided into rigid transformations (consisting of translations, rotations, and perhaps scalings) and nonrigid transformations.

Rigid registration is sufficient when the imaged object (here, the brain or cortex) does not deform between the times the images to be registered are taken. For example, if we successively collect pre-op MRI-T1 and MRI-T2 images of the same patient, even with the same scanner, registration will probably be necessary since the patient is likely to have moved between the scans. However, rigid registration will generally be sufficient since the same object is being imaged. This is also the case if we want to register pre-op images taken with different scanners, such as MRI and PET.

Nonrigid registration is necessary whenever the imaged object deforms between the times the images are taken. This situation occurs almost inevitably in surgery, when an object, such as the brain, is imaged first pre-op and then intra-op. One source of cortex deformation is the phenomenon of “brain shift” that occurs following opening of the skull. Another source is the retraction and resection of brain tissue, e.g., during tumor removal. There are several reasons for wanting to register pre-op and intra-op images. On the one hand, surgical planning is commonly performed on (a) high-resolution structural modalities, such as CT and sMRI, for identifying landmarks and for accurate positioning, and on (b) functional modalities, such as PET and SPECT, for identifying the general location, extent and nature of tumors. On the other
hand, during surgery, image guidance can currently be provided by lower-resolution interventional MRI or US, if available; furthermore no interventional PET is currently available. Examples of interventional-MRI systems are shown in Figs. 1 and 2. Each of the following important tasks requires that one be able to register pre-op and intra-op modalities: (a) To transfer planning from pre-op imagery to intra-op imagery; since intra-op imagery can be related to “patient coordinates,” surgical tools can be accurately directed to target areas, while avoiding critical ones; (b) To display the current position of surgical tools in pre-op imagery, whether structural or functional; (c) To artificially enhance low-resolution intra-op MRI with high-resolution pre-op MRI; (d) To fuse, or otherwise combine, pre-op and intra-op modalities, such as PET and MRI, for enhanced visualization.

This paper describes our work to date in nonrigid registration and multimodality fusion. Our 3D nonrigid registration system relies on tracking of cortex surface deformations and subsequent finite-element calculations of volume deformations. Our 3D multimodality fusion system is based on a 2D fusion system based on the human visual system that was developed at MIT Lincoln Laboratory. The reader should consult [17] for useful related information on earlier versions of our systems.

2 Combining registration and fusion

Figure 3 shows two useful ways in which (1) nonrigid registration and (2) fusion can be combined to great benefits in neurosurgery. The 3 common subsystems are registration, fusion and warping. All images are denoted by “I”. The output of registration is a displacement field, denoted by “D”. The warping subsystem actually performs the required transformation “D” on each of its input images. We distinguish images taken before and after deformation via the qualifiers “before” and “after”. Since we typically deform “before” entities into “after” entities, we also use the synonymous terms “moving” and “fixed,” as well as the corresponding subscripts M and F. The superscripts A, B, C, etc, refer to distinct modalities, such as pre-op MRI-T1. Figure 3 clearly shows that each of the 3 subsystems can be discussed independently.

3 Registration

Figure 4 depicts our approach to nonrigid registration. It is inspired from [8, 9]. Besides the usual notation “I” for an image, we use “R” for the region corresponding to an object (here the cortex), “C” for its contour (i.e., its boundary), “S” for a surface mesh of triangles, and “V” for a volume mesh of tetrahedra. The superscript R denotes the “reference” image that is used to get the displacement field. The reference image in Fig. 3 is denoted by A; it is generally the MRI modality. The inputs to the registration are the reference images $I^R_M$ and $I^R_F$. The output is the displacement field $D^R_F$. All processing described below is done in 3D.

Segmentation: We segment the cortex out of the moving and fixed reference images by using either the
segmentation tool in SPM\(^1\) or the level-set (LS) tools in ITK\(^2\). The output is a binary image corresponding to the cortex region (volume) in the 1st case and to the cortex contour (surface) in the 2nd. Going from either characterization to the other is straightforward.

**Boundary-voxel matching:** The inputs are the contour images \(C_M\) and \(C_F\). The output is a list of correspondences associating to each contour voxel in \(C_M\) a contour voxel in \(C_F\). Some contour voxels in \(C_F\) may remain unmatched. One approach for establishing the correspondences is the `itkDanielssonDistanceMap-ImageFilter` tool in ITK, which is based on the calculation of an euclidean distance map \([6]\). Another approach we are implementing is similar to a method proposed in \([4]\): it is based on an ICP (Iterated Closest Point) technique \([5, 11]\), accelerated by the use of a distance map together with a closest-point map. (The last map avoids the repeated search for closest points.)

**Surface meshing:** We start with the cortex region \(R_M\) and produce a triangle surface mesh \(S_M\). If \(R_M\) contains holes, \(S_M\) consists of several disjoint surfaces. Since our ultimate goal is finite-element (FE) calculations, we have to pay particular attention to the quality of the surface meshes that are generated. Given that the cortex is a “natural” object, i.e., with irregular shape, there are very few tools that can generate FE-quality surface meshes for such an object.

The most promising tool found so far is ISO SURF v1.5d\(^3\) \([15]\). If confronted with a grayscale image, ISO SURF begins by binarizing this image. This is not necessary in our case, since the input \(R_M\) is binary. The tool considers the binary image as a vertical pile of horizontal 2D slices. If the vertical separation between slices is larger than the horizontal resolution, ISO SURF interpolates between slices. A significant feature of the tool is that it is not constrained to interpolate vertically. For example, if ISO SURF finds 2 horizontally-offset ellipses in 2 successive slices and judges that these ellipses ought to be linked, then the tool has the ability to interpolate from one ellipse to the other despite the offset. The technique used is called disc-guided interpolation \([16]\). The technique of regularized marching tetrahedra \([15]\) is then applied either to the original 3D binary image or to the new 3D binary image that also contains the interpolated slices. It is the combination of these 2 techniques that produces the high-quality surface meshes that are well suited for subsequent FE calculations. Even though ISO SURF works band-by-band, the surface meshes it creates do not necessarily consist in a series of horizontal bands linked in the plane of the original slices. Figure 5 contains typical examples of the surface meshes obtained.

**Volume meshing:** We need to create a tetrahedron volume mesh \(V_M\) for \(R_M\) starting from the triangle surface mesh \(S_M\) just created. Here too, the volume mesh must be of FE-calculation quality. The suitable tools we have identified are GRUMMP\(^4\) \([12]\) and TETMESH-GHS3D V3.1\(^5\). For additional details on these tools, the reader is referred to \([17]\). At the present time, we favor TETMESH-GHS3D (GHS3D for short). Indeed, the tetrahedral mesh is computed very rapidly and its quality is high. The problems reported in \([17]\) regarding our inability to generate volume meshes of more than a few hundreds tetrahedra have been resolved.

**Displacement calculations:** We now go back to

\(^{1}\)http://www.fil.ion.ucl.ac.uk/spm.

\(^{2}\)http://www.itk.org.


\(^{4}\)http://tetra.mech.ucb.ca/GRUMMP.

\(^{5}\)http://www.simulog.fr/tetmesh.
Figure 4: Nonrigid registration system.

Figure 5: Views from below of triangle surface meshes produced by IsoSURF for different image data and for different segmentation methods. (a) SPM segmentation of an SPM image. (b) LS segmentation of a Brain-Web image.

the output of the “boundary-voxel matching” processing step in Fig. 4. The pairings obtained at that stage are the starting point of a series of displacement calculations that lead to the output displacement field \( D_{RF} \).

(1) Given the pairings, we can compute the displacement of each boundary voxel in \( C_M \).

(2) Since \( S_M \) was extracted from \( I_M^F \), we can find, with subvoxel resolution, the exact position of each vertex of \( S_M \). We can thus assign a displacement to each such vertex. (3) Using FE calculations, we can derive the displacements of each vertex in \( V_M \). So far, we have limited ourselves to linear elastic FE calculation. For this, we use the \( \text{ASEF} \) solver from the SAMCEF suite of tools [14] from Santech. (4) From the displacement of all the vertices of \( V_M \), we can easily estimate the displacements of all the 1-voxels in \( R_M \). This essentially tells us where each pixel “moves to.” (5) However, we have defined the displacement field \( D_{RF}^B \) as being aligned with \( I_F^R \). Therefore, \( D_{RF}^B \) must tell us where each pixel “comes from.” Deriving the desired field is more than simply changing a sign and is not a simple matter.

We could generate the warped image by scanning the moving image and using the “WHERE-TO” displacement field. However, this would lead to “holes” in the output image. This is a standard problem in performing image transformations. The solution is to scan the output image. There will be no holes, but the inverse “WHERE-FROM” displacement field is required.

At this point, we have the desired output displacement field \( D_{RF}^B \) and we can thus warp any input image as just described.
4 Fusion

The technique we favor for multimodality image fusion is based on an artificial neural network inspired from the human visual system (HVS). This technique was originally developed at MIT Lincoln Laboratory for the fusion of EO, IR, and SAR imagery. Details are in references [13, 7, 18], which in turn provide references to earlier publications. Initial experiments in HVS-based fusion of medical imagery were performed at Lincoln Laboratory. The first paper describing the application of the technique to medical imagery is [2]. Subsequent papers on this topic are [1, 3, 17]. In the fusion of EO, IR, and SAR, the processing was performed on registered 2D images. The resulting fused image, as well as the individual modalities, could then be draped over 3D terrain surfaces. In the initial experiments with medical imagery, fusion was also performed in 2D. However, today, medical imagery is often inherently 3D, especially in the case of brain imagery. The first report of 3D HVS-based fusion is [3]. Whereas our initial work in [17] was 2D, the results reported here are also 3D.

HVS-based fusion is commonly applied to 2 to 4 modalities. Figures 6 and 7 illustrate the technique via an architecture well suited for fusing MRI-PD (proton density) and SPECT images. The first two stages of processing are illustrated in Fig. 6. Both stages rely on a common operator known as the center-surround-shunt operator [10], denoted by 2 concentric circles. The 3D operator is a simple generalization of the more common 2D operator. In the 1st stage, the different modalities are individually contrast-enhanced and normalized. In the 2nd stage, the resulting images are decorrelated and fused. The 3rd stage of processing is illustrated in Fig. 7. In this particular case, the 3 outputs from the 2nd stage of processing are considered to be the RGB components of a color image. If the resulting colors are not satisfactory, they can be remapped. This is performed in HSV (hue, saturation, value) color space. For 3 or more modalities, the outputs of the 2nd stage (after decorrelation and fusion) are interpreted, not as RGB components, but as YIQ components, with the highest-resolution modality generally fed to Y, directly after enhancement. Color remapping is also performed in HSV space.

Finally, Fig. 8 shows an example of fusion of 3 and 4 modalities. Fig. 8(e) shows the fusion of the first 3 modalities shown on the first line of the figure and Fig. 8(f) shows the fusion of all 4 modalities shown on that same line. Sources of imagery are as in [17].

5 Conclusion

We have developed some elements of a system for nonrigid registration and HVS-based fusion of multimodality neuroimagery.

We are currently validating our approach to surface displacement estimation by using synthetic deformations and verifying that we can recover them. We are also working on several improvements to the system. In our work reported so far, we have considered that the cortex is made up of a single material. However, the cortex consists of several distinct regions such as gray matter, white matter and cerebro-spinal fluid. Several techniques exist for segmenting neuroimagery into these regions and we are currently building on them. If we view the cortex as consisting of several distinct regions with different mechanical properties, then we must generalize our approach to FE calculations. One idea that comes to mind is to volume-mesh each region independently. However, this will almost certainly lead to incompatibilities of the volume mesh boundaries near the junctions between distinct regions. In the approach we are pursuing, we first volume mesh the whole cortex as though it consisted of a single region and we then assign the proper properties to the various tetrahedra, based on the label image which corresponds to the output of the segmentation. Refinements may be needed near the boundaries.

Several challenges remain. One challenge is to model the retraction and resection of tissue, which lead to deformations of the cortex. Dealing with resection is particularly problematic since resection implies removal of material. Another challenge is the calculation of nonrigid registration in real-time to allow all modalities to remain aligned at every instant during the course of surgery. Performing the FE calculations in real-time represents perhaps the biggest challenge. Indeed, such calculations are generally done off-line, without any real-time constraint.

References


Figure 6: Stages 1 and 2 of an example system well suited for fusing MRI-PD and SPECT images.


Figure 8: Examples of fusion of 3 and 4 modalities.