Three-dimensional Display of Cortical Anatomy and Vasculature: Magnetic Resonance Angiography Versus Multimodality Integration

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We present two methods for acquiring and viewing integrated three-dimensional (3D) images of cerebral vasculature and cortical anatomy. The aim of each technique is to provide the neurosurgeon or radiologist with a 3D image containing information which cannot ordinarily be obtained from a single imaging modality. The first approach employs recent developments in MR which is now capable of imaging flowing blood as well as static tissue. Here, true 3D data are acquired and displayed using volume or surface rendering techniques. The second approach is based on the integration of x-ray projection angiograms and tomographic image data, allowing a composite image of anatomy and vasculature to be viewed in 3D. This is accomplished by superimposing an angiographic stereo-pair onto volume rendered images of either CT or MR data created with matched viewing geometries. The two approaches are outlined and compared. Results are presented for each technique and potential clinical applications discussed.

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NEUROSURGICAL planning often employs information derived from multiple imaging modalities, the most common being computed tomography (CT), magnetic resonance (MR) and digital subtraction angiography (DSA). The complementary information provided by each is often necessary to fully appreciate the position of a lesion with respect to surrounding structures. However, difficulties arise in correlating observations among each type of image data. Such is the case when a lesion appears in a DSA projection but not in the tomographic data, or vice versa. We wish, therefore, to integrate the data obtained from each modality to create a three-dimensional (3D) image in which the spatial relationships between anatomical and vascular structures may be viewed directly. In so doing, the composite image may provide information that may not otherwise be obvious from a separate analysis of the component images.

The ability to acquire 3D images of blood-flow as well as soft tissue using MRI is an attractive feature of this modality, and technol-

ogy in this field is advancing rapidly. Currently, 3D magnetic resonance angiography (MRA) with near isotropic spatial resolution may be acquired either as a series of two-dimensional (2D) slices, or a 3D volume with the mechanism of flow detection being grouped into two broad classes. The first class is based on the concept of tagging the blood at one location and detecting it at another. Tagging may be achieved actively using saturation,¹ inversion^{2,3} or velocity selective⁴ pulses, or passively by using fully relaxed spins.^{5,6} These methods are referred to as tag detect, time of flight, or inflow, depending upon the specifics of the implementation. The main limiting factors with these methods are the velocity of the blood and the lifetime of the tagging.

Phase-sensitive methods comprise the second class of techniques and are based on a phase modulation of the transverse magnetization due to motion during the application of imaging gradients. Angiograms are produced by performing a complex image subtraction on two images acquired with different velocity-induced phase shifts. Phase differences between images may be obtained by modifying the first moment of the gradient waveforms between acquisitions,⁷⁻⁹ or by acquiring the images at points in the cardiac cycle exhibiting different blood velocities.¹⁰ Phase-sensitive approaches are limited by incomplete stationary tissue subtraction, intravoxel velocity dispersion artifacts, and long acquisition and processing times for 3D flow studies.

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Presented here are the results from a 3D MRA technique based on a velocity compensated field echo (FE) sequence. Passive tagging is employed for the detection of flowing blood, and active tagging, in the form of saturation pulses, is used to suppress undesired blood signals (eg, arterial or venous flow). Here the method is referred to as an inflow technique.

3D MR angiograms may be displayed by performing projections through the volume at a variety of angles and viewing the projections as a cine loop. Among the most common projection methods are: maximum value,⁹ integral, volumetric rendering^{11,12} and surface rendering.¹³ Currently, we use standard volume rendering techniques that have been modified for this application and implemented on a PIXAR (San Raphael, CA) image computer.

Prior to the advances in MRA, the lack of a single modality capable of satisfactorily imaging anatomy and vasculature prompted efforts to combine data acquired from separate modalities. Past techniques for combining angiograms with tomographic data have required that the vasculature be reconstructed in 3D.¹⁴⁻¹⁷ These approaches are complex and seldom fully automated.

Instead, we generate a volume rendered image of the tomographic data (either CT or MR) that precisely matches the viewing geometry used to acquire the projection angiogram, allowing the two to be superimposed. This permits us to take advantage of the high resolution and contrast available with DSA, while relying on the volume rendering algorithm to create a realistic 3D image of the tomographic data (using the PIXAR image computer). Since we routinely acquire DSA projections stereoscopically, the process is performed for each image of the stereo-pair. However, the implementation of this scheme requires that the image data be acquired under stereotactic conditions.

In this paper, a brief description of the two methods used to obtain composite 3D images of cerebral vasculature and cortical anatomy is presented, including image acquisition and display requirements. Results are presented in each case, and the two approaches are compared in terms of their limitations and potential applications in the clinical environment.

MATERIALS AND METHODS

Multi-modality Image Integration

This method of combining images requires that the data be acquired under stereotactic conditions. However, beyond providing the information ordinarily required to analyze the images stereotactically, the image matching procedure is entirely automated. The mathematics underlying this approach are straightforward, and image quality has no effect on the speed of the algorithm. The stereoscopic aspect allows the image pair to be interpreted in 3D without losing the ability to derive stereotactic coordinates from the data. In this respect, the technique may serve as an alternative to the traditional methods of stereotactic surgery planning, all of which are based on 2D image analysis. Details concerning the methodology and apparatus used in stereotactic surgical planning have been described previously^{18,19} and are omitted here.

Image Acquisition and Preparation

Either of two techniques are used to acquire stereoscopic DSA projections. Currently, the preferred method uses an isocentric c-arm gantry rotation with a nominal seven degree angulation. This approach yields stereo-pairs with greater disparity than the alternative approach that uses a dual-focus x-ray tube whose effective lateral shift is only 25 mm. For stereotactic purposes, a 270 mm field of view is used for lateral projections, while 170 mm is used for anterior-posterior views. Projections are acquired in a $512 \times 512 \times 10$ -bit image matrix.

Tomographic images (CT or MR) are acquired over the area of interest within a 325 mm field of view. CT data are stored as $512 \times 512 \times 12$ -bit images, while MR slices are $256 \times 256 \times 12$ bits. Once loaded into the PIXAR frame buffer, the data comprise a "volume" in computer memory. A skew correction is performed if necessary, and additional slices are interpolated to create a volume that is scaled isotropically (ie, with cubic voxels).

The fiducial markers appearing in each DSA image are used to compute a homogeneous transformation matrix that relates the 3D stereotactic frame coordinates of a target to its projected position in the image plane (Fig 1) (ie, to convert frame coordinates x, y, z to image coordinates u, v).²⁰ Similar transformations relate positions between the stereotactic frame and the voxel-based volume contained in PIXAR memory. These allow frame coordinates to be converted to volume coordinates, and vice versa.

Image Matching Procedure

The volume rendering technique we employ on the PIXAR is a modification of the scheme described by Drebin²¹ and is similar to the direct volume visualization technique of Levoy.²² Like the PIXAR method, our rendering algorithm uses an orthographic projection technique. However, the image is created by light reflected off surfaces within the volume, as opposed to light transmitted through semitransparent data, as in the Drebin²¹ model. By employing orthographic projections, the PIXAR avoids the overhead associated with random data access. A high degree of efficiency is achieved in this manner, allowing a rendered

Fig 1. Projection imaging geometry illustrating the coordinate systems of the stereotactic frame and image plane. A mathematical transformation determines where a given 3D point within the frame is projected in the image plane.

image to be generated within minutes. Nevertheless, the technique does not lend itself easily to creating the perspective view needed to properly match the divergent ray projection of the DSA image.

For a volume rendered image to properly match the DSA projection, each voxel in the tomographic data must be projected to its corresponding position in the DSA image. This is accomplished by re-orienting and re-shaping the volumetric data, prior to rendering, in a fashion that allows the orthographic projection to yield the desired result.²³ The necessary operations are performed using routines on the PIXAR included in its Chap Volumes software library.

The matching process is initiated by decomposing the DSA-associated transformation matrix into its separate components (translation, rotation, perspective, scaling, and cropping).²⁴ These specify the orientation and position of the stereotactic frame with respect to the image plane (image intensifier) during the projection. Similarly, we determine the orientation of the frame within the tomographic volume (relative to the intended axis of projection to be used when creating the rendered image) then rotate it to match its orientation in the DSA projection.

In order to simulate the diverging ray geometry of the DSA projection, successive planes within the volume are progressively resized from front to back, perpendicular to the projection axis. The resizing operation must be centered along the principal ray (which intersects the piercing point). The entire volume is then translated by the amounts required to align the principal ray with the piercing point in the DSA image. This assures that the data are properly centered within the intended field of view. Each plane of voxels parallel to the image plane is then scaled to match its projected dimensions in the DSA image.

Prior to rendering the volume, it is sometimes useful to remove a portion of data overlying an area of interest to prevent it from being obscured. The angiogram is superimposed on the rendered image using an image compositing utility on the PIXAR. Transparency may be varied between the two images to prevent one from entirely occluding the other. The procedure is repeated for the second projection of the stereo-pair, and the images are transferred to a

stereoscopic workstation where they may be analyzed conveniently using specialized software.²⁵

MR Angiography

Data acquisition. The radio frequency (RF) and gradient pulse sequence shown in Figure 2 was used to acquire the MRA data presented here. The sequence is a velocity compensated spin-warp field echo (FE) experiment that is performed in either a 2-D or 3-D acquisition mode. The phase preparation gradient is rephased after echo collection to avoid artifacts from a spatial variation in transverse magnetization coherences,^{26,27} and constant amplitude spoiler gradients are applied in the slice-select and read directions. The phase of the RF excitation pulse is cycled over a sequence (0, 0, π , π) throughout the measurement to increase T1 contrast. Signal from blood flowing into the imaged region may be suppressed using an optional $\pi/2$ saturation pulse applied in the presence of a weak slice-select gradient and followed by strong spoiler gradients. This has the effect of destroying the magnetization from a slab adjacent to the imaged region. In order to minimize susceptibility artifacts and higher order gradient moments, the echo time (TE) is reduced to a minimum within the time constraints imposed by the velocity compensation waveform. This is achieved through the use of a partial (5/8) echo acquisition and an asymmetric excitation pulse.

A 1.5 T whole body imaging/spectroscopy MR scanner (Gyrosan S15HP, Philips Medical Systems, Shelton, CT) with 10 mT/m nonshielded gradient coils and a linearly polarized head coil was used in all imaging experiments. Each study consisted of 64 to 128 slices with a thickness of 2 to 4 mm and 0 to 2 mm overlap. Repetition times (TR) of 50 to 200 milliseconds, TEs of 12 to 18 milliseconds and excitation angles (θ) of 45° to 90° were used depending upon slice thickness and desired contrast characteristics. Two studies were performed. In the first, a single acquisition was used to image both the stationary tissue and flowing blood. The second used two acquisitions in which the above sequence was used to first image flowing blood, followed by a separate FE or spin-echo acquisition to image the stationary tissue. In the first case, TRs of 150 milliseconds to 200 milliseconds were used with a θ of 90°, while a TR of 50

Fig 2. Radio frequency and gradient pulse sequence employed to acquire MRA data.

milliseconds and θ s of 45° to 90° were used for the flow images in the second case.

Post Processing and Display. The acquired data are transferred from the MR scanner to the PIXAR image computer via a high-speed data link. The intensity values in each slice are normalized for the entire data set. The volume(s) are then interpolated to produce isotropic voxels. For a single acquisition data set, the rendering method begins with two linear continuous classifications of the original data to produce an optimal contrast static tissue volume and a segmented vasculature volume. Each voxel in both volumes is assigned a three component color (Red, Green, Blue) and an opacity value ($\alpha, 0 < \alpha < 1$). An α value of 1 represents a completely opaque voxel, while 0 represents complete transparency. For the stationary tissue volume, a heated metal color mapping is used, while a grey scale is used for the vessels. A 3-D gradient calculation is also performed on the original data to produce an estimate of the surface strength and direction at each voxel. This information is used by a simple lighting model to modulate the intensity of the static tissue data as a function of the angle between the light source and the surface normal. The segmented vasculature is then composited²⁸ over the static tissue and rendered using a depth-cued opacity integral.¹¹ Prior to the gradient calculation, a portion of the static tissue volume may be suppressed to expose underlying vessels in the final rendered image(s). In the case of a dual acquisition, the processing is identical with the exception of the initial mapping to segment vessels from soft tissue.

RESULTS AND DISCUSSION

An example where CT and DSA have been matched is presented in Fig 3. A slice thickness of 5 mm was used for the CT data that were acquired with no overlap. The DSA stereo-pair was acquired using a 270-mm field of view and a gantry rotation of approximately 7° . A section of the tomographic volume was removed prior to rendering in order to better view the hemi-

sphere of the brain containing the imaged vasculature. A large amount of transparency was applied to the CT data, and a dual-phase angiogram was selected in order to observe both arterial and venous phases. The CT stereotactic plates are seen at the sides.

A matched MR-DSA stereo-pair is presented in Fig 4. The MR data was obtained using a T1 weighted 2D multislice spin-echo acquisition, giving 64 2 mm transverse slices. An arterial phase DSA stereo-pair was selected and overlaid on the rendered volume that was cut along a midsagittal plane. Vessels in the same plane are seen to follow the appropriate convolutions in the rendered data.

An example of combined MRA and cortical data is shown in Fig 5. The rendered image was produced using the method described above and was based on a single acquisition. The data were obtained in a multislice acquisition mode (ie, more than one slice interrogated per TR) with a 230 millisecond TR, 17 millisecond TE, 90° excitation pulse angle, two signal averages, and no presaturation. A total of 128 3 mm transverse slices, with a 2 mm overlap were acquired (1 mm center-to-center slice spacing). The cortical surface in each slice was traced using a semiautomated intensity contouring routine. The contours were filled and convolved with a gaussian kernel to produce a blurred mask that was then multiplied against each slice. The blurring operation ensures a continuous change at the edge of the mask to avoid artifacts in the final rendered image that would

Fig 3. Stereo pair of matched CT and DSA. A dual-phase angiogram was combined with volumetrically rendered CT data comprised of nonoverlapping 5 mm thick slices. The CT fiducial marker plates are seen at the sides.

Fig 4. MR-DSA matched stereo-pair. Prior to processing, the tomographic volume was cut through the midsagittal plane. Vessels in the same plane in the arterial phase angiogram follow the appropriate convolutions.

Fig 5. Volume rendered image of MRA data showing cortical surface and vasculature. The data were obtained using a multislice acquisition mode, giving 128 3 mm transverse slices with a 2 mm overlap.

arise due to artificially high gradient values at a binary mask boundary.

An important consideration in evaluating the multimodality image integration technique is its ability to accurately match the viewing geometry of the DSA projection. In a perfectly matched volume rendered image, every voxel in the tomographic data is projected to its corresponding position in the DSA image. With our approach, the tomographic volume is subjected to several interpolation steps involved in rotating, translating and reshaping its contents. As a result, aliasing may affect target localization accuracy when analyzing the final image(s). Preliminary studies of the matching accuracy indicate that, within the first standard deviation, voxels within the tomographic data are rendered to within 1 pixel of their true positions in the DSA image.

Although we are limited to generating views defined by the acquisition geometry of the DSA projections, the stereoscopic technique provides depth cues that enhance the 3D perception of the rendered volume. The composite stereo-pair allows the entire 3D structure to be viewed directly and removes the necessity for the neurosurgeon to mentally combine anatomical and vascular images. Surgical planning may be simplified and the image database size reduced dramatically.

MRA acquisition techniques, on the other hand, allow vascular and anatomical data to be acquired in a single study. Such data are inherently 3D. Thus, while the display modality may be similar to that used with the former technique, the observer is not constrained to only one view. He may, in fact, select the optimum view for the procedure, or generate a cine loop to rotate the entire volume. In addition, the 3D relationships between structures and marked targets or probes may be determined directly, rather than by inference from stereoscopic disparity.

Because the MR data are truly 3D, it is possible to enhance depth cues in the rendered volume through the use of shading, transparency, etc. However, the image matching procedure discussed earlier lacks the ability to prop-

erly apply such cues to the imaged vasculature. Since the projection angiogram is overlaid on the rendered image, it may occlude portions of the data actually overlying the vessels. We attempt to minimize these conflicting depth cues by introducing some transparency when superimposing the vasculature. Stereopsis may be relied upon to resolve further ambiguities that may arise.

DSA, nevertheless, provides sufficiently high resolution to image the finer intracranial vessels, whereas MRA techniques encounter some difficulties. Furthermore, in instances where it is necessary to visualize bone, CT may be the preferred modality and so, MRA may not be suitable. The most appropriate technique, therefore, depends largely on the nature of the information that is sought from the image data.

CONCLUSION

We have demonstrated the methodology for two techniques of displaying composite 3D images of anatomy and vasculature in the brain. The first approach, using conventional angiography combined with a tomographic modality, requires that both studies be performed under stereotactic conditions. Although this method yields only a limited number of views, it provides a higher resolution rendition of the vessels than MRA and may be selective with respect to the vessel system imaged. The MRA technique, on the other hand, while yielding a 3D volume of vasculature and anatomy in a single examination, cannot match the resolution provided by DSA, neither can it provide selective arterial, venous, or vessel system imaging. However, based on the results presented here, it could well provide an alternative to DSA methods for the planning of stereotactic neurosurgery since it provides a 3D data set and avoids errors that might otherwise be encountered in accurately combining data obtained from separate modalities.

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