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A three-dimensional visualization and navigation tool for diagnostic and surgical plannig applications

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ABSTRACT

This study aims at providing the radiologist and the surgeon with a diagnostic and planning tool. To this end multimodal (T_1, T_2 and PD-weighted) sets of MR images representing a human head and a human knee with and without neoplastic formations were acquired. All the software was developed in C++ language using Open Graphics Library (OpenGL) and OpenGL Volumizer. It was tested on a Silicon Graphics O_2 workstation. The medical user can rotate along the x-y-z axes the volume under investigation and zoom in and out the data, can make cuts of the set of images in all directions and display volume intersections with the three conventional anatomical planes. By enfolding the volume in a cube and by moving its apexes, the user can dig the volume. The surfaces of the anatomical districts can be visualized. The tool renders a composite volumetric image by using the false-colouring technique and it can combine morphological information of the surface and data about the nature of the volume by using the different distribution of the intensity levels of the pixels. It is also possible to set transparency to obtain an image representing simultaneously the 3D volume and its internal structure. The tool can display surface information and volume information at the same time and provides endo-navigation facility that helps the user to move into an anatomical district in order to find the correct position of potential lesions and the way to remove them.

Keywords: bioengineering, image analysis, image segmentation, false-colouring technique, 3D visualization, endo-navigation

1. INTRODUCTION

Although scientific research in oncology led to remarkable results in prevention and treatment of such diseases, some further steps are relevant with reference to the surgical domain. Early detection of neoplastic lesions and an effective capability to plan the surgical intervention play a basic role for improving patients survival. From techniques such as X-ray computed tomography (CT), magnetic resonance imaging (MRI), single photon emission tomography (SPECT) and positron emission tomography (PET), it is possible to obtain detailed structural and functional information on the anatomical structures analyzed without performing surgical operations. The 2D images generated by such medical imaging system are often sufficient to get the desired medical information. Nevertheless nowadays, by using recent advances in rendering algorithms and increased computer workstation power, a 3D visualization of the anatomical structure under analysis can be carried out in a short additional time. However, medical tasks like diagnosis, monitoring therapy, surgical planning are actually performed by the operator mentally integrating the complementary information as provided from the different diagnostic modalities. This visual analysis asks for a remarkable mental effort from the surgeon, and meaningful examples exist for which this is not even enough to obtain a correct classification of the tissue under examination. It is difficult to mentally reconstruct the three dimensional shape of the anatomical structure under analysis. The doctors, above all the radiologist, are going to understand that by using a computer system there is a potential efficiency in working with 3D models, if they can be easily constructed, in order to define the extent of the lesions and the connections with the surrounding tissues and anatomical structures. The 3D commercial visualization tools usually include some facilities to enhance the contrast of the pixels, to perform a fully automatic segmentation by thresholding and 3D region growing, to calculate a 3D model of the selected area, to offer real time rotation along with the ability to apply transparency and/or depth shading to the model. As a function of such consideration, the medical user must be provided, as a support to his actions, with a guided software environment made not only of different methods of high performance image processing, but also of an highly integrated graphic capabilities.

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2. METHODOLOGY

The first step to develop tools devoted to processing and visualization of medical images is the acquisition of medical data by using dedicated devices. Some of them yield output data in a standard format that can be easily read by subsequent software tools. However, often it is necessary to make a conversion from the proprietary format of the medical device to a standard one (e.g. DICOM-3). Moreover, the output is a set of 2D images, acquired as parallel planes with a gap among them, to be further collected into a stack to create a 3D data set. The interpolation facilities allow to insert correct data in the gap between two adjacent slices. There are a lot of different interpolation techniques. The simplest one is to duplicate existing slices, a better algorithm is the nearest neighbour interpolation where the gap is filled of voxels by using the value of the pixel nearest to the edge. We can not also forget the linear interpolation where the value of the pixel changes in a linear manner between two adjacent slices. Usually the set of images obtained by medical devices, after the conversion, include a lot of noise. This noise is created by the medical device itself and the behavior of the patient during the acquisition phase. Some software techniques are able to decrease such noise. The diagnostic characterization of MR slice images is based on the differences in relaxation times (T_1, T_2) and proton density (PD) between normal and pathological structures. Such differences show up as different intensity of the T_1, T_2 or PD-weighted signals. All of this information is usually not acquired in clinical situations; however, it can be used to create "composite" [1,2] colored images. A single such composite image contains all of the information contained in the separate T1, T2 and PD-weighted monochrome images. Although an HSV (hue, saturation, value) or similar model can be used to generate MR composite images, we choose to make use of the hardware-oriented red, green, blue (RGB) model. We took T_1, T_2 and PD-weighted images sets and directly assigned them, respectively, to the R, G, B color channels of our display. We choose these assignment because (a) PD is a kind of signal carrier since the T_1 and T_2 signal strengths are proportional to proton density (PD). We made the analogy between this and the green light component (G) of the RGB space. The G component is the mayor part of the human visual system's intensity (luminance) signal and carries most of the spatial information. Furthermore, (b) we subsequently tried assigning T_1, T_2 to R, B, respectively, and also to B, R and found that the latter gave better color discrimination for the datasets we were using. This approach offers an immediate visualization of the complete MR data. It allows for a better discrimination between normal and pathological tissues than by viewing the T_1, T_2, PD-weighted images as three monocrome images sets. The user, in this case a radiologist or surgeon, can easily discriminate tissues by their color characteristics. The color statistics of the image can be used to interactively segment out different tissues and to facilitate simulation of surgical interventions. Starting from 3D MR slice-stacks in PD, T_1 and T_2 we constructed a three-dimensional color composite volumetric image. If the user decide to load just one set of images he will of course obtain a gray level 3D image based on relaxation times T_1 or T_2 or proton density. By loading two or three sets of images the user will obtain a colored 3D image where each color represents a different acquisition modality according to the false-colouring technique. In order to be able to set the transparency of the 3D image, a threshold value (alpha) is used together with RGB values for every pixel. A typical problem during the visualization of 3D images is the time performance. In order to be able to work in real time we decided to use a preview window, at the bottom left of the display (see Figure 3), where an under-sampled (about 64 times the original one) 3D image is represented. Firstly the user can apply current interactive utilities (cut, rotation, zoom, surface detection, luminance set) and view the result by looking at the preview window and then he can transfer the final result into the large volume where full 3D visualization is performed in real time.

Segmentation is the partitioning of an image into regions. These regions should correspond to different tissues, organs or functional areas. We devoted particular attention to this step. We tested some new segmentation tools in order to increase the detection of a particular region in the image under examination, resulting in an improvement of the accuracy of the image set. While a lot of segmentation tools exist, it is difficult to select one with a good precision for all the anatomical districts under examination. For example, a good segmentation tool for the brain could not be as good in segmenting the bone. Often human intervention is needed. After segmentation the regions belonging to the same tissue may result divided in subgroups. For example, we can distinguish a vein from an artery (both being vessels).

The application of segmentation techniques to multimodal biomedical volumes is, moreover, difficult to implement since the voxel values, especially at the border between volumes of interest, correspond to complex mixtures of different anatomical tissues. Scanning all the co-ordinates which compose the multimodal volume, we can obtain a new space, named "the feature space", in which it is represented the statistical distribution of each feature of the acquired data. In other words, the different "n" values of the intensities related to each voxel in the multimodal volume can be viewed as co-ordinates of that voxel in the n-dimensional feature space, in which further analysis can be carried out. Our approach to the process of segmentation consists in the definition of clusters into the n-dimensional feature space and in the classification of all voxels of the multimodal volume in the resulting classes. In this sense, it is clear that techniques such as edge detection and thresholding cannot produce results comparable with those obtained by the supervised and unsupervised techniques of classification by using the feature space. Supervised techniques have been largely used in medical imaging segmentation studies, but they provide results that hardly comply with the clinical environment, being the definition of the proper label for each tissue a time consuming and expensive process, since relevant errors can be introduced, especially from an unexperienced user. On the other hand, the unsupervised approach is interesting, since it makes transparent to the user the entire process. Furthermore, unsupervised techniques can be used interactively, leaving to the user the final control over the achieved segmentation. In this context, the core of our system is composed of unsupervised algorithms that will be explained in the following. Our system includes fuzzy methods able to compensate the effect of the sensor near the border of the tissues. The whole system has been structured with strong attention to modularity, in order to simplify and speed-up the use of its functionality. Our segmentation system has been composed of different algorithms, some able to partition the feature space into clusters of similar extension, and some having the capability to perform adaptive clustering. This means that some of the algorithms are able to identify yet cluster of small extension (high probability density clusters) or clusters of large extension (small probability density clusters). The problem of clustering of biomedical images requires further consideration due to the complexity of the data and to the distribution of the anatomical tissue in the feature space. Let us describe in details the structure of the data related to multimodal volumes. As already mentioned, it is necessary to describe structural objects which are mapped into the feature space as clusters of small extension (high probability density clusters) or as clusters of large extension (small probability density clusters). This is equivalent, in the image space, to the existence of both large extended regions with good defined boundaries and regions of small extension with fuzzy borders. Both type of clusters are visible in the feature space at different resolution level. An adequate cluster analysis must be able to identify with accuracy both categories, by performing the analysis at different resolution of the probability density distributions. Our system is composed of different algorithms for unsupervised clustering, able to perform clustering of features that can exhibit different resolutions, such as the classical hard C-Means [3], and the Capture Effect Neural Network [4] and of fuzzy clustering algorithms [5], [6], [7]. Fuzzy methods, in particular, are not affected from the dimensionality of the feature space and are able to perform an optimal unsupervised clustering at different resolutions. Our approach to the segmentation consists, after the clusterization of the feature space with the aforementioned algorithms, in the identification of the resulting classes. This can be achieved by using the labelling or coding algorithm that is introduced in the following. Once the clusters in the feature space have been identified, another step is required to generalise the obtained results to the whole volume. Two information are explicitly provided from the previously mentioned algorithms: the first one is related to the centres of the clusters, the second one is related to the membership between a pattern of the feature space and every cluster. These information can be combined by using an unsupervised labelling algorithm to obtain a fully unsupervised segmentation map. To this end, voxels with different distances in the feature space are not directly assigned to clusters, i.e. voxels are non assigned to a cluster as a consequence of their distance only. For each voxel, instead, its membership to a cluster is computed and the voxel is assigned to that cluster where membership is maximum. The fully unsupervised segmentation map can be often considered already a satisfactory result. Otherwise, if the knowledge domain of the human expert is needed, various modules have been developed to fine tune the segmentation map in order to improve the overall performance of the system. It is therefore possible to perform interactive correction of each voxel in the image space. The classes representative of the anatomical structures are rendered in a three-dimensional environment and they can be reduced or expanded if their aspect does not match the real one. These operations can be performed by using morphological operators [8], [9], which are composed of structural elements constituted of geometrical shapes related to the anatomical structures under analysis. The user can iterate the procedure to obtain the final segmented volumes. Furthermore, the system allows the definition of a region of interest directly on the segmented volume. This procedure makes possible for the user the fine tuning of the entire process, for example by manually leaving out non relevant details while pointing out to the anatomical structures of interest. The combination of unsupervised segmentation techniques and of interactive post-processing algorithms assigns to the user the control of the final result, minimising the need of time consuming and error prone manual procedures.

Once the segmentation phase has been accomplished the next step is visualisation. A description of the available rendering utilities of our tool is given in the following. A utility of our tool is devoted to make a planar cut of the 3D image. If user chooses this utility a semitransparent rectangular plane will show up on the 3D image. The user can change the position of the plane with a rotational and translational tool. Before making the cut, the user has to decide the cut direction (from the selected plane to the bottom of the 3D image or from the selected plane to the top of the 3D image). This tool is considered very important according to user experience since it is able to show, as a 2D image, after cutting the top of a portion of 3D image. In order to improve quality of surgical planning our tool allows to make also generalized cuts. The 3D image is 'encapsulated' inside a cubic volume. The user can deform it by moving its apexes. All the deformation made on the cube are transferred to the 3D image. So, the result is a 3D image with a lot of cuts that is equivalent to dig into the volume and the final effect is a better visualization. By using a look up table (LUT) the tool is able to change the color value and the opacity of the volume. The user can, for example, separate the gray and white matter, skin and muscles, tissues with different densitometric characteristic. It is possible to show the intersections between the three principal planes (transversal, sagittal, coronal) giving the user the possibility to navigate through these planes in order to investigate the anatomical structures. Surface reconstruction builds a surface between the edges of adjacent slices. To achieve this goal a lot of segments that connect the vertexes of adjacent planes are displayed. The surface is defined as a texture of triangles. This procedure is not so good as we can think since if there is more than one edge for each slice the resolution decreases and the time process increases. By working on the edge detection and by using a surface reconstruction it is possible to improve 3D visualization. In order to obtain the surface reconstruction we followed these steps (see Figure 1). Firstly, the decision about which regions have to be connected among the images in the set has to be taken. Every 2D segmented region can be connected to one, to more than one or to no one near regions. Secondly, for each region that needs to be connected we have to decide how to connect the points of the edge. It is important to find out the triplets of vertexes that will build the triangles creating the network elements of the surface of the anatomical structure. Thirdly, to make a final visualization of a 3D surface image, our algorithm connects all the triangles by using the idea of a polygons mesh: every side of the triangle is shared at maximum between two polygons, a side connects two vertexes, and a side has to participate in some polygon. In order to decrease memory space and to increase the speed of the tool each side is defined by three pointers to the elements of a list of vertexes.

The use of a luminance model is necessary in order to make the 3D visualization a more realistic one. In fact, without a luminance model, all the surfaces will have the same luminance. By using different shades of colors a 2D image on the screen can become a realistic 3D image. It is needed to have an interaction between the light and the surface, balancing the light and the surface effects. Of course, the increase of the realism by using the luminance technique is proportional to the increase of the computational cost. To make a correct analysis of the luminance, we have to consider: power of calculus, the average complexity of the scene measured as number of polygons, the need to have real time images, the total number of images. We used the Phong model [10], where the reflection of the light is divided in three different kinds: (a) diffused: it reflects in the direction proportional to the incident angle with the reflecting surface. It is the same for all the observers; (b) specular: there is a favored direction as function of the incident direction and of the normal to the surface direction; (c) ambient: it reflects in all directions with the same intensity and it is used to clear up the shaded zone. Each kind of light has a different component for each RGB channel, so every pixel of the image has a 3x3 matrix representing the three kinds of light and the three RGB components. In order to find the best balance between the light and the surface, three different kinds of surfaces need to be considered: (a) specular: it seem shining because the reflected light is diffused in a short range of directions near the reflecting point; (b) diffusive: the reflected light is scattered in all the directions; (c) translucid: the light can enter into the volume but it comes out from a different point. We decided to use in our tool a hybrid visualization in order to be able to merge surface information with volume information. In this way, the user can have access to more information (morphological one resulting from the different surfaces and densitometric one resulting from the volume). The only problem is to not be able to work in real time since a lot of information must be processed. We adopted the preview window tool.

Our endo-navigation tool can be used with all the images of all anatomical districts. It needs as input a set of 2D images. At the beginning, we thought to use a sphere (simple to be developed for its symmetry) as shape to dig the volume, but later we decided to use a paraboloid that gives the user a better perception of the depth and that is able to show images very similar to the conventional endoscopic ones. In the working window, the user has always access to the 2D set of the medical images of the anatomical district under investigation and their 3D reconstruction. To start the endo-navigation the user must choose the point of view (P_2) and the point of center of the scene (P_1). Our technique puts the user at the point P_2 from where he can start his navigation. At the bottom end of the image he see P_1. During his navigation he can change the amplitude of the paraboloid and the distance P_1-P_2. During the endonavigation the user can look at his position (P_2 of course) and at the center of the scene (P_1) in the 2D or 3D image in order not to loose his position (see Figure 2).

3. DATA

We have chosen MR as our test vehicle because co-registered sets of three parametric images are easily obtainable directly from a single scanner in a single session. The images of a human head with and without neoplastic formations have been acquired at Advanced Biotechnology Center (ABC) in Genoa by using MR device and the images of a human knee at ESAOTE S.p.A in Genoa by using an ARTOSCAN system. The original data were three sets (T_1, T_2 and PD weighted) of 27 slices (256 x 256 x 12 bits) with a gap of 5mm between slices.

All the software was developed in C++ language using OpenGL and OpenGL Volumizer, graphical libraries which combine low level functions (characteristic of libraries like OpenGL) with other devoted to the treatment of volumetric data like the biomedical ones. The whole system has been implemented on a Silicon Graphics O_2 workstation with UNIX IRIX 6.5 operating system. By using QT, a user-friendly interface for visualization and navigation was developed. Qt is a cross-platform C++ GUI application framework. It provides application developers with all the functionality needed to build state-of-the-art graphical user interfaces. Qt is fully object-oriented, easily extensible, and allows true component programming. We decided to develop all the software in a Silicon Graphics O_2 workstation because the better performances of processing and visualization with respect to a Personal Computer environment. The high 3D quality and good performances of the Silicon Graphics O_2 are based on: (a) 32 bit graphic capability, (b) hardware support for texture mapping that gives the possibility to make zoom, rotation of large size and high resolution images in real time, (c) dedicated processor for imaging facilities. OpenGL is the computer industry's standard application program interface for defining 2D and 3D graphic images. Prior to OpenGL, any company developing a graphical application typically had to rewrite the graphics part of it for each operating system platform and had to be aware of the graphics hardware as well. With OpenGL, an application can create the same effects in any operating system using any OpenGL-adhering graphics adapter. OpenGL specifies a set of commands or immediately executable functions. Each command directs a drawing action or causes special effects. OpenGL comes with a large number of built-in capabilities. These include hidden surface removal, alpha blending (transparency), antialiasing, texture mapping, pixel operations, viewing and modeling transformations, and atmospheric effects (fog, smoke, and haze). OpenGL Volumizer is a graphical API (Application Program Interface) that allows standard graphics applications to treat volumetric and surface data in a similar fashion. With this API, more sophisticated applications can be created, providing researchers with unparalleled visualization and exploration capabilities, significantly reducing time to insight and discovery. OpenGL Volumizer enables: (a) mixing of volumetric objects and geometric objects within the same 3D scene, (b) deformation of the shape of a volume (c) specification of arbitrarily shaped regions of interest. Additionally, as volumetric data can coexist in the same scene as surface geometry, most any operation that can be performed on a 3D surface such as shading and picking can be applied to 3D volumes as well.

4. RESULTS

The system has been tested by using acquisition of human brain districts from multimodal MR images. The medical user does not need to be computer trained for using the tool for his work and he can view all the images in a format very similar to the one he is used to. The medical user can rotate along the x-y-z axes the volume under investigation and zoom in and out the data. He can make cuts of the set of images in all directions and display its intersections with the three conventional anatomical planes (transversal, sagittal and coronal) in order to analyze the anatomy and the pathology of the body district under visualization. By enfolding the volume in a cube and by moving its apexes, he can dig the volume and, therefore, obtain information that it was impossible to obtain by using only the 2D images along the three conventional planes. The medical user is able to visualize in 3D with great detail the lesions and to characterize with high precision their position. The surfaces of the anatomical districts can be visualized to study the features of the shape that can guide the medical user to identify lesions of different nature. The tool increases diagnostic and planning capabilities since it renders a composite volumetric image in color from multimodal MR data sets (T_1, T_2 and PD-weighted), by using the false-colouring technique. The tool can also combine morphological information of the surface and data about nature of the volume by using the different distribution of intensity levels of the pixels. It is also possible to set the transparency to obtain an image that represents the 3D volume and its internal structure. The tool does not delete the volume that hides other volumes placed on the back side, while it makes the volume itself transparent: in fact, it is important, according to medical doctor experience, to be able to look at those volumes near to the one under close attention. The tool can display surface information and volume information simultaneously. The results of the application of the 3D visualization and navigation tool are shown in the attached figures (see Figures 3 to 9). To increase the capability of our tool for a surgical planning we developed endo-navigation facilities. Our endonavigation tool is able to help the medical user to move into a anatomical district in order to find the correct position of lesions and the way to remove them. In particular this is effective when the position of the lesion is not near to an edge and when it is not possible to use the traditional endoscopy methods as during the investigation of the brain. Our tool is provided also with a small database in order to be able to save and load all the data of a patient: all the examinations of a patient can be easily saved and retrieved (see Figures 10 to 12).

5. CONCLUSION

Multimodal digital images, integration of morphological and functional data, multidimensional rendering techniques and endo-navigation facilities have become powerful "navigation aids" for tissue sampling, functional therapeutic procedures and computer-aided surgery. We think that our tool could be considered as the first step to obtain solutions totally integrated with the surgical instrumentation and of ease use in several fields of surgery.

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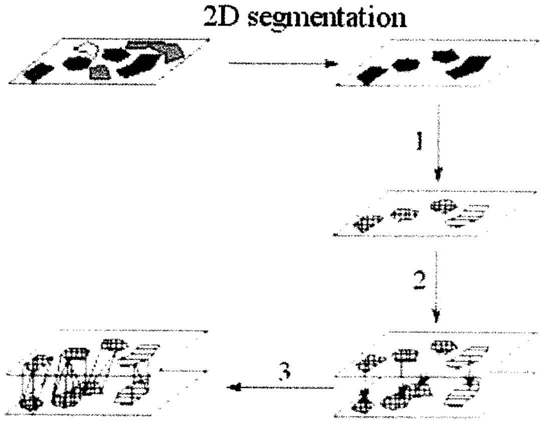

Figure 1: 3D region surface segmentation

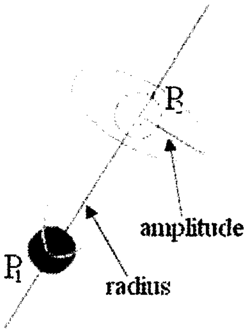

Figure 2: endonavigation tool scheme

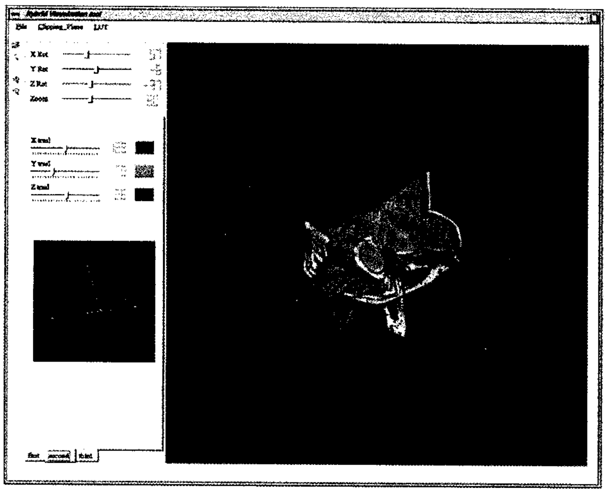

Figure 3: rotation and visualization along the three conventional planes of a composite volumetric image from multimodal MR data

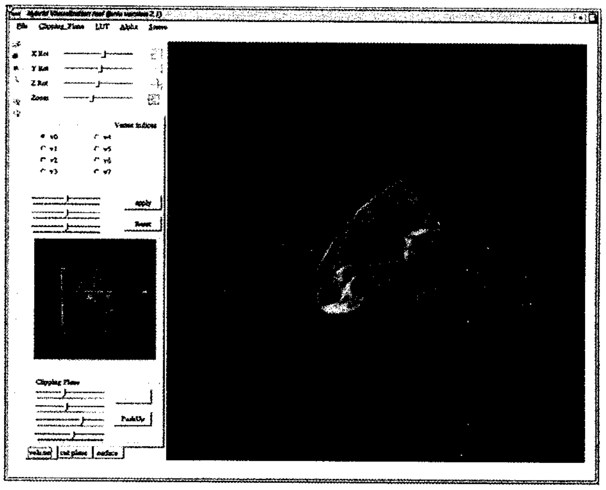

Figure 4: cut along a generic plane of a composite volumetric image from multimodal MR data

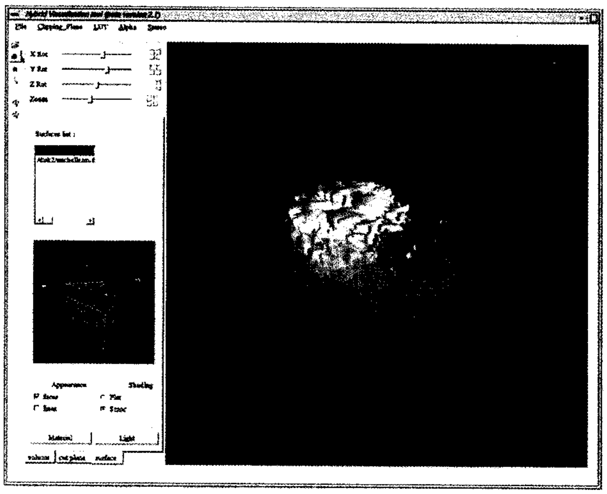

Figure 5: composite volume transparencies

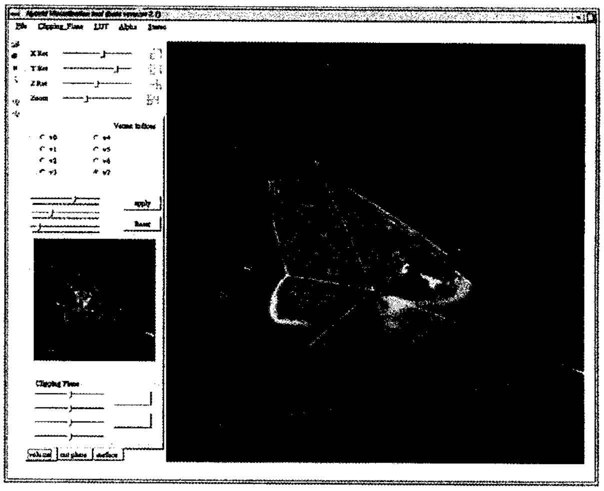

Figure 6: composite volume manipulation

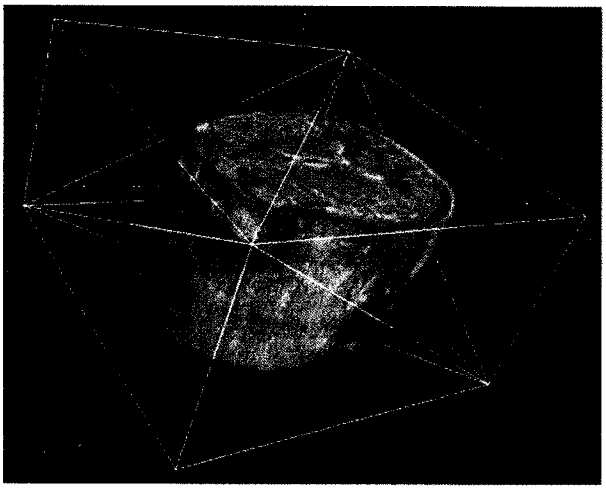

Figure 7: composite volume manipulation

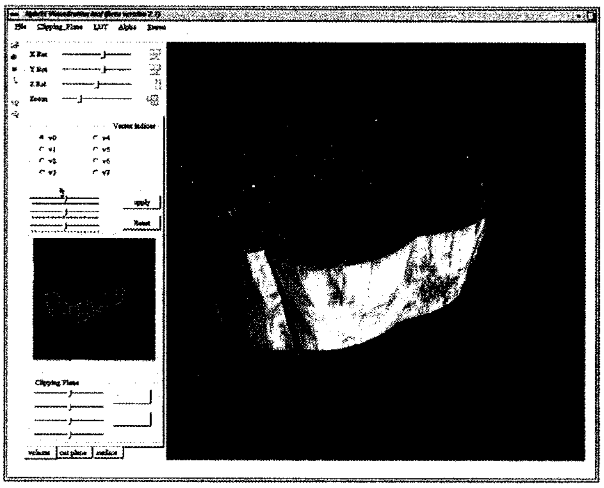

Figure 8: composite volume and surface visualization

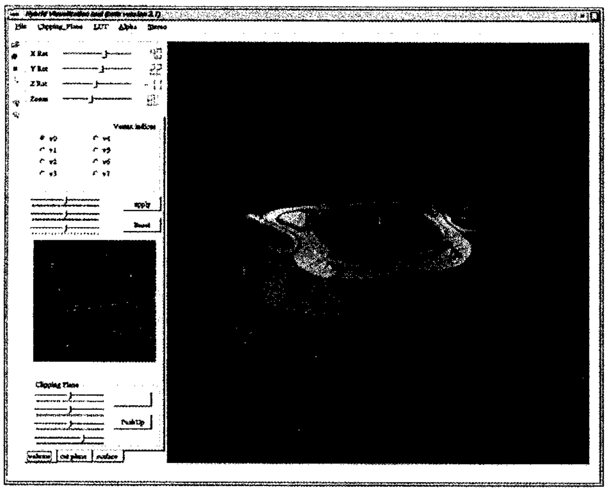

Figure 9: composite volume selection

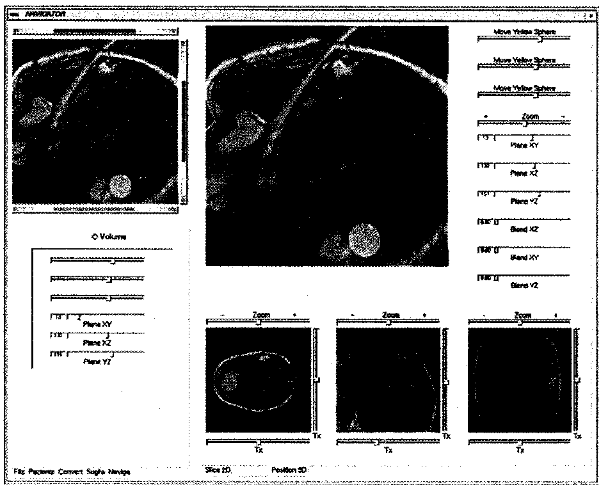

Figure 10: endonavigation tool: 2D and 3D external display and GUI

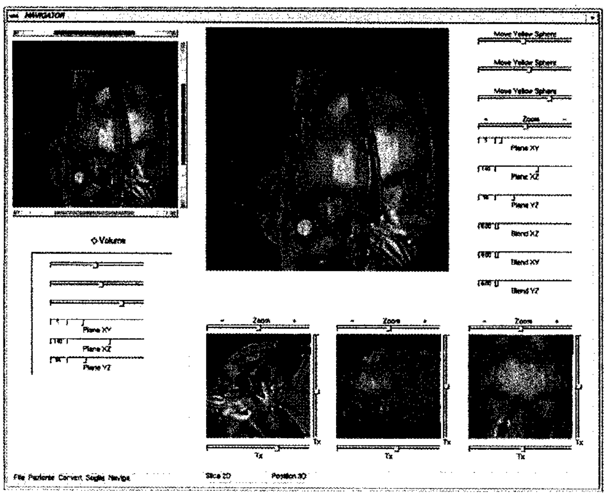

Figure 11: endonavigation tool: 2D and 3D external display and GUI

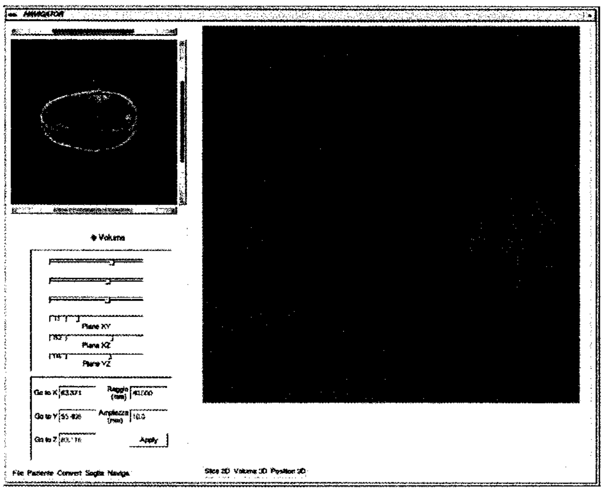

Figure 12: endonavigation tool: 3D display from an interval point of view