Developing a virtual reality environment for petrous bone surgery: a "state-of-the-art" review

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Introduction

The increasing power of computers has led to the development of sophisticated systems that aim to immerse the user in a virtual environment. The benefits of this type of approach to the training of physicians and surgeons are immediately apparent. Unfortunately the implementation of “virtual reality” (VR) surgical simulators has been restricted by both cost and technical limitations. The few successful systems use standardized scenarios, often derived from typical clinical data, to allow the rehearsal of procedures. In reality we would choose a system that allows us not only to practice typical cases but also to enter our own patient data and use it to define the virtual environment. In effect we want to re-write the scenario every time we use the environment and to ensure that its behavior exactly duplicates the behavior of the real tissue. If this can be achieved then VR systems can be used not only to train surgeons but also to rehearse individual procedures where variations in anatomy or pathology present specific surgical problems.

The European Union has recently funded a multinational 3-year project (IERAPSI, Integrated Environment for Rehearsal and Planning of Surgical Interventions) to produce a virtual reality system for surgical training and for rehearsing individual procedures [1]. Building the IERAPSI system will bring together a wide range of experts and combine the latest technologies to produce a true, patient specific virtual reality surgical simulator for petrous/temporal bone procedures. This article presents a review of the “state of the art” technologies currently available to construct a system of this type and an overview of the functionality and specifications such a system requires.

The basis of Virtual Reality environments

Virtual reality (VR) represents computer interface technology that is designed to leverage our natural human capabilities. Today's familiar interfaces - the keyboard, mouse, monitor, and Graphical User Interface (GUI) - force us to adapt to working within tight, unnatural, two-dimensional constraints. VR technologies, however, let users interact with real-time 3D graphics, supplemented with other sensory interfaces (sound, touch, even smell) in a more intuitive, natural manner. VR encourages viewers to be participants immersed in the data rather than passive observers watching from a distance by using a combination of specialist computer peripherals to allow adequate user interaction. The familiar view of virtual reality is of users equipped with head-mounted displays (HMDs) and instrumented clothing, such as gloves and whole-body suits. However, the cost, reliability and health and safety issues associated with this form of *immersive* VR has led to diminished interest, with more basic head- and spectacle-mounted "personal information displays" dominating the market. Desktop implementations (using standard computer screens), together with conventional or stereoscopic image projection systems have become popular of recent years. "Higher-end" visualization techniques, such as the CAVE (small rooms defined by large video projection walls) and dome-based or "wrap-around" imaging systems are very impressive. However, in the medical world, they tend to be restricted to wealthy foundation or governmental research laboratories.

The most important change has been the arrival of low-cost, industry-standard multimedia computers and high-performance graphics hardware. Coupled with this, the spread of accessible VR modeling and run-time software, together with low-cost and free resources from the Web, is beginning to make VR much more accessible to the non-specialist user or developer than was the case just two years ago. Consequently, it is believed that practical VR based applications will soon become common-place in the hospital [2] .

The development of a virtual reality system to simulate petrous bone surgical procedures must involve the user in the loop of a real-time simulation mimicking a realistic synthetic operating environment. Ideally, the system should take as input anatomical reconstructions produced from standard medical imaging modalities and construct patient-specific virtual anatomic models that can be both autonomous and responsive to user actions. Data are beginning to emerge that demonstrate positive impact of this type of training experience when measured in the surgical environment [3,4].

Mastoidectomy, cochlear implantation and cerebellopontine angle tumor surgery are prototypical examples of ENT surgical procedures that require a high level of dexterity, experience and knowledge. They also represent a range of surgical complexity and are thus good targets for the development of specialized surgical simulators of direct interest to ENT. Such simulators must provide high fidelity visual simulation together with accurate haptic feedback simulating interactions between surgical instruments and tissues. These tissues must be modeled in order to provide a realistic sensory response that reflects individual tissue properties and reactions [5].

Ergonomic considerations in the design of virtual reality systems

The value of VR systems in training depends on their ability to transfer specific decision-making and physical skills to the operator. In practice this may be optimally achieved by simplified systems that model the ergonomic features of surgical tasks rather than providing an exact virtual reality replica of the surgical environment. The VR community is increasingly adopting this approach and measuring transfer of training and improved performance in the real world using objective techniques to measure the success of training.

The identification of the essential ergonomic components involved in a complex task such as petrous bone surgery requires a detailed task analysis. Without this step there is a risk that any VR system will fail to record or measure those elements of human skill that it was initially intended to target. The task analysis should form an early and central component of any project that involves a major human-centered component. This has recently been recognized by publication of the International Standard ISO 13407, *Human-Centred Design Processes for Interactive Systems* [6].

The task analysis can be complex. For the IERAPSI project for instance the initial task analysis involved a review of existing documentation describing operative procedures, detailed interview with experienced operators and review of existing training aides including cadaveric temporal bone drilling, synthetic bone dissection exercises and CD ROM training systems. Following this a detailed review was performed of video recordings from mastoidectomy, cochlear implantation and acoustic neuroma resections (both translabyrinthine and middle fossa approaches). In the final stage the ergonomist observed procedures being performed in theatre in order to test and refine the task analysis.

Constructing the VR Environment

In order to design a system that is capable of producing VR environments for surgical simulation in individual patients we must develop methods to rapidly model the anatomy and pathology based on the patients imaging investigations. We can use data from any imaging system that produces 3-D data sets and can combine information from several modalities that contain complementary data. To investigate the petrous bone, computed tomography (CT) and magnetic resonance imaging (MRI) are most commonly employed. CT provides high spatial resolution bone images whilst MRI provides images of soft tissues. In practice there is often a need to combine these images and there may, in the future, be a requirement to include images from other 3D imaging modalities such as single photon emission computed tomography (SPECT) and positron emission tomography (PET).

The process of defining a VR environment from this imaging data is a major challenge. For clinical use the procedure must be rapid and automatic. The process can be subdivided into three separate stages: 1) Spatial co registration of data from multiple modalities, 2) Identification of tissue types (segmentation) and 3) definition of tissue boundaries for the VR environment.

Spatial Co-registration

In order to use data from multiple modalities we must first co-register the data into a common Cartesian reference framework so that the same point in images of each modality represents a single point in the patient. Spatial co-registration of 3D medical image volumes is now a well-established and widely used technique in both research and clinical practice [7,8,9,10,11]. The process requires two steps. Firstly the translations and rotations required to match the images are derived. This may be achieved by manual image matching; co-registration of fiducial points or, more recently, by automated co-registration algorithms [10,12]. These algorithms commonly work by minimizing a defined statistic produced by registration of the image volumes. Secondly, the data must be “re-sliced” (transformed) so that each data set is in the standard Cartesian space and resampled into equal voxel spacings. The combination of automated co-registration and data re-slicing allows the production of matched volumes of imaging data in which true spatial registration exists at a voxel by voxel level.

Tissue Segmentation

In order to construct a realistic virtual reality environment it is necessary to identify which type of tissue is present at each coordinate in the data space and to identify the precise location of the edges between tissue types. ***The process of identifying the distribution of different tissue types within the data set is known as tissue segmentation***. In practice the development of automated algorithms for tissue segmentation is complex and many approaches have been described. Most of these use image intensity information from single or multiple images in order to identify which tissue type each voxel represents. At the most simple level a tissue might be identified in an image if it had a distinctive range of image intensities. If this range of image intensities were constant and showed no overlap with the intensities of other tissues then any voxel within this range could be confidently classified as belonging to this tissue (a process known as windowing or thresholding). In practice this idealized situation does not occur, several tissues will commonly display similar ranges of image intensity, these ranges may vary within the data set due to heterogeneity of imaging process (noise) and a single voxel will commonly contain multiple tissue types (partial volume averaging). Many simple segmentation techniques are designed to label each voxel as belonging to a single tissue type and ignore the fact that most voxels around a tissue boundary will contain mixtures of tissues. This problem of boundary pixels containing multiple tissues, all of which contribute to the image intensity is known as partial volume averaging. A more logical approach is to calculate the probability that each voxel conforms to each particular tissue type, which allows an estimation of the partial volume effect. The use of techniques to produce probability maps effectively transforms any set of imaging data into a series of probability images each representing a separate tissue. In a series of probability maps each voxel would have a separate value for each map, corresponding to the proportion of the voxel filled by that specific tissue type. The use of probability maps in tissue segmentation allows us to develop algorithms using strict statistical approaches to the segmentation task and to identify edges between tissues, which will lie at the point where each tissue probability is equal to 50 percent, this is used as the basis for many visualization techniques that require the identification of surfaces (vide infra [13]). .

A number of algorithmic approaches can be used to derive probability maps from original imaging data. In MR and CT data the grey levels in an image can be assumed to be formed by a linear process. This means that the contribution to the intensity in any pixel is simply proportional to the relative fractions of each tissue within the voxel [14]. On this basis the probability that any voxel contains a particular tissue type can be calculated using simple linear algebra using data from N-1 images (where N is the number of tissues to be identified). This approach will deliver unbiased estimates of tissue proportion [15,16]. However, it can only deliver correct estimates for the tissues within the model, meaning it cannot deal with unexpected (or pathological) behavior. From a medical standpoint this is equivalent to saying that it can only deal with normal tissues.

A more generic and useful approach is to develop a probability model for each tissue component present in the data, which also accounts for partial volume effects. The various parameters in the density model must be determined using an optimization algorithm to minimize the difference between the model and the data (The simplex algorithm [17]and expectation maximization[18] are appropriate). Estimation of relative tissue probabilities can then be made by the direct use of Bayes theory. This probability labeling technique will work with multiple tissues on a single image provided that the grey level distributions do not overlap significantly (Figure 1) . Overlapping tissues can be eliminated by the use of multiple images, as ambiguous regions in the data can be separated with additional information. However, this does involve a slightly more complicated analysis in order to determine all of the parameters in the multi-dimensional model. (Figure 2) This technique can be extended to deal with pathological (unmodelled) tissues by allowing an additional category for infrequently occurring data [19] . Variations in the probability distribution of individual tissues, which might result from heterogeneity of the image acquisition process, must also be considered. In magnetic resonance imaging in particular where marked heterogeneity in signal intensity occurs across the acquisition field, these can be corrected, with consequent improvement in the accuracy of tissue segmentation, by automated correction of the imaging data for heterogeneity prior to analysis [20]. Figure 3 illustrates the strategic considerations required to select the appropriate tissue segmentation strategy that will be most effective on any particular set of image data.

In practice the implications of these theoretical considerations are straightforward. The use of simple segmentation techniques such as thresholding, which classify each voxel as belonging to a particular tissue, will work only if the signal intensities of the tissue to be segmented are unique. This explains the common use of thresholding methods to identify bone from CT images where the massive X-ray attenuation of bone results in relatively clear distinction between bone and other tissues. Where tissue intensities are similar or overlap, which is common in MRI data then thresholding techniques will not work. In these data sets segmentation is best performed using statistical models of normal tissue which will attribute the probability of a voxel containing a particular tissue. This statistical approach has two other advantages in that it allows the use of information from multiple images (eg CT and MRI) which improves the confidence with which the segmentation can be made and it allow the estimation of the fraction of each voxel which is filled by a particular tissue (ie it deals with the problem of partial volume averaging).

Using these statistical approaches the accuracy of tissue segmentation is very high and manual intervention is rarely required. The main problems lie in the classification of pathological tissues such as a partially cystic and partially necrotic tumour where the statistical characteristics of the tissue vary considerably. In these cases a simple segmentation based on signal intensity will not work perfectly. However the use of anatomical information about how close similarly classified pixels lie to each other, combined with the statistical information provides a powerful solution to this problem since it uses the assumption that voxels of particular tissue types are likely to be connected together. The combination of statistical segmentation and these connectivity algorithms means that the accuracy of automated tissue identification is high and manual intervention will seldom if ever be needed..

Physical Modeling

If a virtual reality system is to accurately mimic the tactile (haptic) and auditory responses to specific actions then the VR environment must be equipped with a spatial physical model of the relevant characteristics of each of the tissues within it. Physical modeling is a computationally expensive approach to virtual reality but, in this specific field of application, it is essential since it is the only practical way to accommodate for the arbitrary positioning in the area effected by the operation of the surgical tools and the use of realistic anatomical models derived from patient images. The computational costs due to physical modeling are partially mitigated by the fact that the surgical procedures mentioned are constrained by a restrictive field of view and limited haptic interaction between the surgeon and the patient, The most relevant physical processes that should be addressed are: a) collision detection, b) bone dissection, and c) interaction with soft tissues.

Fast and accurate collision detection between models is a fundamental problem in computer-simulated surgical environments. In the context of physically based simulation, the output of a collision detection algorithm is used to impose non-penetration constraints and to compute reaction forces between surgical instruments and tissues and between tissues themselves (e.g., between tumor, bone, and drill during excision of a a cerebellopontine angle tumor).

Bone is hard and has a stress-strain relationship similar to many engineering materials. Hence, as discussed in Fung[21], stress analysis in bone can be made in a way similar to the usual engineering structural analysis. The simulation of the drilling of the temporal bone involves first the detection of collisions of the drill burr with the bone surface, then, depending on the type and location of the contact, a prediction on the amount of bone to be removed and of the forces that should be returned to the hand of the user via the haptic feed-back device. Given the particular nature of the process simulated, the natural way to model the temporal bone anatomy is by using a finite element volumetric approach. This means that a mathematical model is calculated using known data about the tissue (bone) including its hardness, rigidity and resistance to drilling and is used to calculate the responses to an intervention such as drilling by applying the model to each small component (finite element) of the bone involved in the interaction and its immediate neighbours. The geometric model can be directly derived from patient CT data [22]. The general problem of accurately modeling the dynamics of a deformable object , such as soft tissue, undergoing large deformations is complex and the standard technique used in computational science, (finite elements modelling) is computationally very demanding[23]. The complexity increases even further when it is required to model actions, such as cuts, that can change the topology and physical properties of the body itself.

Working in the Virtual Environment

Our sense of physical reality is a construction derived from the symbolic, geometric, and dynamic information directly presented to our senses and from prior knowledge[24]. The techniques and devices used to return sensory information are thus as important as the simulation methods employed. In the case of petrous bone procedures, the most critical aspect is the quality of haptic feedback from surgical instruments and visual feedback from the operating microscope.

Visual Feedback

Since the human body is a three-dimensional volume the issue of computer-generated three-dimensional volumes representing the human body is integral to the application of visualization (and VR) in medicine. Without the use of stereo displays, the main problem in volume visualization is how to render sampled volumetric information onto a 2D screen. Early algorithms of volume visualization utilize the "additive projection", which computes an image by averaging the voxel intensities along parallel rays from the rotated volume to the image plane (Figure 4). This simulates an X-ray image and does not provide information about depth relationships[25]. Another method is the "source-attenuation reprojection", also referred to as "opacity", allowing object obscuration [26]. The improvements in available computing power have allowed the implementation of more complex and more appropriate methods for the visualization of 3D objects, these include surface rendering and volume visualization.

The segmentation of anatomical structures described above produces 3D maps of probability. Each voxel in these maps describes the probability that the voxel represents the tissue in question. The boundaries of this object can be easily extracted and represented by a series of geometric primitives (ie triangles) derived from the volumetric data. The shape, position and size of these primitives can be calculated by a variety of techniques. These techniques use a variety of approaches to connect points in the 3D space with the same value (contour tracing), which generates a series of primitive shapes, which form a surface (surface extraction). The derived surfaces represent a plane in the 3D model on which all points have the same probability value and are called isosurfaces. In medical image data usually these are selected to correspond to the surfaces of anatomical structures or to surfaces of equal functional activity. The surface abstraction may go only as far as deriving a family of polygons to represent an isosurface for example by the application of the 'Marching Cubes' algorithm which calculates a series of primitives for each voxel based on the values within the voxel and its immediate neighbors (Figure 5)[12]. The advantage of the method is that an extracted polygonal surface may be displayed at interactive rates on a modern Personal Computer. An alternative that requires less pre-processing is to use a solution that does not explicitly derive geometric surface primitives, such as that used by Tan et al in their transputer based medical workstation.[27]. Another method commonly used for the visualization of 3D medical image data is known as 'volume rendering' [28,29]. This visualization technique works by projecting imaginary rays through the data volume which project onto a viewing plane with a value related to the physical property represented in the voxel array. For example, a volume of CT data containing bone with a high X-ray absorption coefficient might be projected with a high value. Generally, a volume rendered image appears different from that of a surface rendered image in that anatomical structures are presented as having some degree of transparency (Figure 6). For some clinical procedures such as image-guided biopsy or trans-cutaneous thermal ablation, transparency may greatly enhance depth perception and thus increase the accuracy of the procedure. The transparency which volume rendering offers also enables the placement of surgical instruments within 3D structures with great accuracy [30]. Until recently, volume rendering was considered to be inherently slow due to the large voxel data sets that had to be processed for each new view of the anatomy. However, the development of new ideas and algorithms for volume rendering using texture mapping hardware architectures has removed this obstacle[31].

Presentation of the rendered images

Visual simulations achieve the illusion of animation by rapid successive presentation of a sequence of static images. The critical fusion frequency is the rate above which humans are unable to distinguish between successive visual stimuli. This frequency is proportional to the luminance and the size of the area covered on the retina [32,33]. Typical values for average scenes are between 5 and 60 Hz. The method chosen for the presentation of the rendered images depends on the application.

Stereoscopic presentations require the rendering of two images with a disparity corresponding to the binocular disparity that would be expected for viewing the object at a chosen distance in real life. The single perceptually fused image has the appearance of a real three-dimensional object. This kind of presentation is suitable for the use of virtual and augmented reality in clinical applications. Stereoscopic images may also be similarly delivered to each eye by display on small Liquid Crystal Display (LCD) arrays placed close to the eyes in a head mounted display (Figure 7).

Some methods of image presentation project separate images via LCD arrays or video systems into each eye of the observer to simulate binocular parallax so that the visualized data appears to be floating in the viewing space. In this form it is amenable to direct 3D physical measurement [34,35,36,37,38,] An important aspect of this kind of display is that the viewer is unencumbered as is the case with using a head mounted display, and does not have to adopt a tiring posture. Many attempts have been made to get stereoscopic projection without needing additional glasses using devices called Autostereoscopic Displays (ASDs). These systems also display separate images to each eye in order to simulate binocular parallax but present these images using technical approaches that allow the user complete freedom [39]. In order to achieve this, these systems commonly require mechanisms to monitor head position and eye movement. The Dresden 3D Display (D4D), used in the IERAPSI project (Figure 8), features several properties not found in other ASDs. All components – head tracking, eye position finder, and appropriate adjustment of the visualization display – are integrated into the D4D. The combination of binocular stereoscopy and head tracking effectively constitutes a 3D television display.

Haptic Feedback

Haptic feedback systems are designed to provide touch and proprioceptive information. Haptic devices not only provide this information to the user but most also sense physical input from the user to guide actions within the virtual reality environment. The primary input/output variables for the haptic sense are displacements and forces. To manipulate an object, move it, rotate it, or pinch it, the haptic system must issue motor action commands that exert forces on the object. These forces are highly dependent on the type of grasping that is used. The physical interaction between the user and haptic devices must accurately simulate the ergonomic requirements of the task that is being simulated. The IERAPSI user requirement analysis identified the PHANToM force feedback arm as the most appropriate commercial device (Figure 9). The PHANToM system is capable of 6 degrees of freedom position input and 3 degrees of freedom force output allowing simulation of a full range of instrument movement and the provision of force feedback to simulate resistance and vibration.

Spatio-temporal Realism

The preceding discussion has emphasized that a virtual reality simulator for petrous bone surgery is required to offer multiple synchronized input/output modalities and that for each of these modalities timing constraints have to be met in order for applications to be usable. Moreover, varying delays in the various output devices makes proper synchronization even harder [40]. Human beings are very sensitive to these problems. Since the various components of a petrous bone simulator have to receive input and produce output at considerably variable rates, it is expected that accurate simulators will require improvements in computing performance which can only be achieved by the use of parallel processing techniques in order to meet the timing constraints imposed by the task. The recent improvement and proliferation of high performance multiprocessor PCs and high speed network interfaces make this solution practically viable for a large community of users.

Currently Available Systems

There are training aids available for otolaryngology, and some use of virtual reality for this purpose has already been reported (see below). Widely used are the Pettigrew Plastic Temporal Bone series (Figure 10). Using Pettigrew's models the complete temporal bone, for example, can be fully dissected using standard theatre equipment, with a similar effect to that achieved during cadaveric exercises. The Pettigrew bones incorporate clever canal modelling techniques and innovative use of material. The trainee is required to perform a mastoidectomy and then continue to expose and identify such features as the horizontal and vertical portions of the facial nerve, the ossicles, the round window niches, the lateral semicircular canal, and so on. Food dye has been added to create bleeding effects during irrigation.

A multimedia solution is the Temporal Bone Dissector CD, published by Mosby (Figure 11) [41] . The CD has been developed using a combination of Macromedia animation and QuickTime movies and provides good introductory material. However, it does not provide a virtual training environment.

Among the earliest reports on the clinical use of 3D data visualization were applications in craniofacial surgery. CT data was ideal for imaging bone and had an acceptable spatial resolution. Craniofacial surgery also requires careful preoperative planning since the effect of surgery will be both functional and aesthetic. It was possible to use the relatively slow computers available at the time since most procedures are non-urgent in nature [42] . Later, surgical simulation systems [43]and interactive workstations[44,45,] were developed with functions that specifically addressed the problems of simulating, rehearsing and planning craniofacial surgery interactively [46]. The latest systems use physical models of tissue behavior to provide accurate predictions of post-surgical facial appearance [47]. Clinical assessments have demonstrated the superiority of computer based visualization over other methods in craniofacial and orthopedic diagnosis and the application of these methods to craniofacial surgery has now been thoroughly validated [48].

In the area of ENT surgery an endoscopic sinus-surgery (ESS) simulator has been developed by the Ohio Supercomputer Center and Ohio State University Hospital [49]. This simulator provides intuitive interaction with complex volume data and haptic (force) feedback sensation (Figure 12). A laboratory at the Univ. of Washington subsequently carried out a joint project to construct and evaluate a VR-based simulator for training physicians in endoscopic sinus surgery. This project used the ESS simulator as its starting point. The results of the validation concluded that the simulator did provide a valid and useful implementation of many endoscopic sinus surgery tasks, but needs to be carefully integrated into the training curriculum for optimum benefits[50].

The Ohio Supercomputer Centre has also been involved in more recent work with the Children's Hospital in Columbus, Ohio, to develop a virtual temporal bone dissection simulator. A Virtual Workbench has also been used to develop a system for planning base of the skull surgery, and a commercial product – Virtual Intracranial Visualization And Navigation (VIVIAN) is available. This work has been carried out at the Kent Ridge Digital Laboratories in Singapore.
Harada produced volume visualisations of the temporal bone from histological slices and have also proposed their use for surgical training [51].

A group at Guy's Hospital, London is developing an AR microscope system for neuro and ENT procedures. Features from preoperative radiological images are accurately overlaid in stereo in the optical path of a surgical microscope. Their system is already adequate for several procedures and has been used in the operating theatre. They are also working on extending their system to deal with soft tissue deformation [52]. The University of Illinois Chicago (UIC) VRMedLab networked facility (Figure 13) [53] is designed to provide an educational resource to surgeons of otolaryngology, enabling them to visualize bone-encased structures within the temporal bone using interactive 3D visualization technologies. Digital sections of the human ear and temporal bone (prepared from actual glass slide specimens) make up the VR model, supplemented with special sculptures and converted CT records of objects too small to reconstruct from the physical samples (eg. ossicles). This VR system has not been designed to replace the cadaveric drilling experience but does appear to provide users with an improved mental model of regional anatomy. A group at the Institute of Otolaryngology in London have reported a number of trials to determine the accuracy and precision which may be achieved in an operating microscope augmented reality environment. Various procedures which are used in surgery were carried out in this environment. An autostereoscopic system was used for 3D image presentation. The accuracy and precision achieved demonstrated that the use of augmented reality is entirely feasible for skull base surgery [54]

Summary

The technical limitations restricting the production of VR surgical simulators have largely been surmounted. Improved imaging devices can produce data of adequately high spatial resolution and signal to noise ratio to provide a basis for modeling of the virtual environment. Co-registration of data sets and the automated segmentation of anatomical structures is made possible by improvements in algorithmic approaches and computing power. Physical modeling, at least of rigid structures is becoming increasingly sophisticated and the improvements in visual and haptic feedback systems allow true subject interaction in a stereoscopically rendered 3D environment. Most importantly the use of dedicated graphics hardware and multiprocessor computers has reduced the time taken for volume rendering techniques to the point where it is feasible to perform these tasks at a rate sufficient to appear as continuous motion to a human. The combination of these technologies will be challenging but offers every promise of a routine clinically useable surgical simulator for use in hospital settings.

Figure Legends

Figure 1:

Figure 1A shows a T1 weighted MRI demonstrating a 1cm acoustic neuromas in the right cerebellopontine angle (1A). Figure 1B shows a probability map showing the results of a tissue segmentation on this single image, white represents a probability of 1 that the image is acoustic neuromas, black represents a probability of zero. Figure 1C shows the intensity distribution of the pixels within the red sample area shown in figure 1A. The pixel intensities are shown in red and the fitted probability function in blue. The central peak (blue arrow) corresponds to pixels of brain tissue and the upper peak (red arrow) to pixels of enhancing tissue. Calculating the probability that they belong to the distribution of enhancing tissues derives the probability map in fig 1B. Figure 1B shows that the probability map clearly identified the tumour but also identifies a small blood vessel to the left of the mass which is also enhancing.

Figure 2:

Figures 2 A-D show a large left sided acoustic neuromas on 4 different MR sequences (2A: T1 weighted with contrast; 2B: T2 weighted; 2C: T1 weighted inversion recovery; 2D: time of flight MR angiogram showing areas of blood flow). Figure 2E show a plot of the signal intensity of the pixels in these images in a multi-spectral space that optimizes the separation between the individual tissues.. In the multispectral scattergram colours represent tumour (purple), CSF (brown), Bone (yellow), Fat (orange), Grey matter (blue), white matter (red) and peripheral soft tissues (green).

Figure 3:

Figure 3 illustrates the strategic considerations required to select the appropriate tissue segmentation strategy that will be most effective on any particular set of image data.

Figure 4:

Maximum intensity projection (additive projection technique) of a T2 weighted image of the inner ear.

Figure 5:

Figures 5A and 5B show 3D isosurface renderings of the same data set illustrated in figure 4. Figure 5B is rendered at a higher isosurface level than 5A showing the effect of changing the isosurface in single dataset. Figure 5C shows a rendering of the cochlea from a patient being assessed for cochlear implantation. Note the proximal obstruction of the scala tympani.

Figure 6:

Volume rendering of the same data set as figure 4. Figure 6A shows a rendering without transparency. Figure 6B shows the effect of increasing transparency on the rendering.

Figure 7:

Head mounted stereo video display unit

Figure 8:

The Dresden 3D Display (D4D), used in the IERAPSI project (Figure 8). The unit features head tracking and eye position tracking using the sensors on the top of the unit. These are used to present a realistic 3D representation of the visualization which appears between the flat panel and the user.

Figure 9:

The PHANToM force feedback arm used in the IERAPSI system. The PHANToM system is capable of 6 degrees of freedom position input and 3 degrees of freedom force output allowing simulation of a full range of instrument movement and the provision of force feedback to simulate resistance and vibration.

Figure 10:

The Pettigrew Plastic Temporal Bone series for rehearsing petrous bone drilling.

Figure 11:

Excerpts from the Mosby multimedia Temporal Bone Dissector CD.

Figure 12:

Demonstrates the endoscopic sinus-surgery (ESS) simulator developed by the Ohio Supercomputer Center and Ohio State University Hospital [55]. This simulator provides intuitive interaction with complex volume data and haptic (force) feedback sensation.

Figure 13:

A 3D volume rendering from the University of Illinois Chicago (UIC) VRMedLab networked facility[56]. The system is designed to provide an educational resource to surgeons of otolaryngology, enabling them to visualize bone-encased structures within the temporal bone using interactive 3D visualization technologies.

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