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Principles and Applications of Computer Graphics in Medicine

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Abstract

The medical domain provides excellent opportunities for the application of computer graphics, visualization and virtual environments, with the potential to help improve healthcare and bring benefits to patients. This survey paper provides a comprehensive overview of the state-of-the-art in this exciting field. It has been written from the perspective of both computer scientists and practising clinicians and documents past and current successes together with the challenges that lie ahead. The article begins with a description of the software algorithms and techniques that allow visualization of and interaction with medical data. Example applications from research projects and commercially available products are listed, including educational tools; diagnostic aids; virtual endoscopy; planning aids; guidance aids; skills training; computer augmented reality and use of high performance computing. The final section of the paper summarizes the current issues and looks ahead to future developments.

Keywords: visualization, augmented and virtual realities, computer graphics, health, physically-based modeling, medical sciences, simulation

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1. Introduction

Over the past three decades computer graphics and visualization have played a growing role in adding value to a wide variety of medical applications. The earliest examples were reported in the mid 1970's when three-dimensional visualizations of Computerized Tomography (CT) data were first reported [1,2]. Today, a variety of imaging modalities are in common use by the medical profession for diagnostic purposes, and these modalities provide a rich source of data for further processing using computer graphics and visualization techniques. Applications include medical diagnosis, procedures training, pre-operative planning, telemedicine and many more [3].

This paper traces the development of the use of medical visualization and virtual environments (VEs), and highlights the major innovations that have occurred to date. Dawson [4] summarizes well the three most important criteria for surgical simulators: (i) they must be validated, (ii) they must be realistic, and (iii) they must be affordable. It is only by satisfying all of these criteria that a surgical simulator can become an accepted tool in the training curriculum, and provide an objective measure of procedural skill. These factors are generally equally relevant to all medical applications where computer

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graphics are used. The latest generation of computer hardware has reached a point where for some applications realistic simulations can indeed be provided on a cost-effective platform. Together with the major changes that are taking place in how medical professionals are trained, we believe that there will be a growing demand for the applications and techniques that are discussed in this survey. Validation of these tools must of course be carried out before they can be used in everyday clinical practice, and this area is sometimes overlooked. The most successful projects are therefore multidisciplinary collaborations involving clinicians, computer scientists, medical physicists and psychologists.

This paper includes a comprehensive reference list and summarizes the important work and latest achievements in the computer graphics field and their application in medicine. Section 2 begins with an overview of the current constraints on clinical training, which highlight the need for extending the use of medical visualization and virtual environments. Section 3 focuses on the main classes of computer graphics algorithms that are commonly used within medical applications. Section 4 highlights the problems of modeling the physical properties of tissues. Section 5 then shows how the technologies described have been used to provide innovative applications. Applications are categorized into the following areas: educational tools; diagnostic aids; virtual endoscopy; planning aids; guidance aids; skills training and assessment; computer augmented reality and use of high performance computing. Section 6 discusses the current and future challenges in this domain, and we end with conclusions.

2. Background

The increasing availability of high resolution, volumetric imaging data (including multi-detector CT, high field strength Magnetic Resonance (MR), rotational angiography and 3D ultrasound) within healthcare institutions has particularly impacted on the specialities of Surgery, Radiology and other clinical areas where such data are in widespread use for diagnosis and treatment planning. There has been a paradigm shift in our ability to visualize, evaluate and measure anatomical and pathological structures which has overtaken and all but eliminated previous invasive diagnostic modalities and diagnostic surgical exploration (laparotomy). The ability to diagnose smaller abnormalities is improving the early detection and management of sinister diseases such as cancer. The generation of patient specific virtual environments from such information could bring further benefits in therapeutics, teaching, procedure rehearsal, telesurgery and augmented reality. One area of rapid development within radiology, interventional radiology (IR), uses various combinations of imaging data to guide minimally invasive interventions [5]. This includes fluoroscopy for angioplasty and embolisation (vessel dilation and blockage), CT and ultrasound for biopsy and nephrostomy (kidney drainage), and at the development stage, open MR scanning for real time catheter guidance.

Amongt new paradigms for learning clinical skills, there is a precedent in surgery for training and objective assessment in validated models [6–11] where global scoring systems objectively assess surgical skills [12]. The metrics used in this process may be determined by decomposing the tasks and procedural steps of invasive procedures, using appropriate methodology. These performance objectives can then be used in simulators to automatically provide truly objective assessment of a trainee's performance in the simulated procedure. Virtual environments have been used to train skills in laparoscopy and a number of validation studies have confirmed the efficacy of such simulator models. Laparoscopy simulators available include the MIST VR (Mentice Corporation, Gothenberg) [13], VEST system *one* (VEST systems AG, Bremen, Germany) [14], LAP Mentor (Simbionix, Cleveland, Ohio) [15]. Application of virtual VE to train in interventional radiological (and other) procedures, with specific objective measures of technical skill, would allow radiologists to train remotely from patients, for example, within the proposed Royal College of Radiologists' Training Academies. These establishments are under development at three pilot sites in the UK and will provide a unique environment for radiology training, including clinical skills, outside the confines of the UK National Health Service. This would improve access to clinical skills training while increasing efficiency in the NHS due to removal of the time consuming early period of training from patients. This trend is not specific to the UK, and statutory and other organisations in many countries are operating similar policies to achieve the benefits of skills training. Prevention of deskilling and a wider availability of skills could address unmet needs for these procedures, for example, in cancer patients. One interventional radiology simulator, the VIST-VR, has recently been partly validated for training of carotid stenting and its use has been recommended by the United States Food and Drug Administration prior to commencing procedures in patients [16].

3. Algorithms and Techniques

This section reviews some of the basic algorithms and techniques for medical volume visualization. The section is divided into four parts. First, it describes the filtering and segmentation steps which act as a pre-processing of medical image data before application of a visualization technique. Secondly, it discusses the main approaches to the visualization of volumetric data in medicine. Brodlie & Wood [17] provides a more detailed survey. It then provides an overview of the main techniques used for combining two or more image modalities of the same patient—often called image fusion or 3D image registration. Finally, it summarizes the algorithms being used for soft tissue modeling—an important component of a medical virtual environment.

3.1. Filtering and segmentation

Image data acquired from a scanner will inevitably contain noise. Many filtering techniques have been proposed in order to remove the noise, typically smoothing by replacing the value at each voxel by some averaging over a local neighbourhood. However, in medical applications, this smoothing can blur the boundaries of anatomical features. A better approach for medical data is to use an anisotropic diffusion-based technique where the image intensity values iterate toward an equilibrium state, governed by an anisotropic diffusion partial differential equation. The diffusion function depends on the magnitude of the gradient of the intensity, and so diffusion occurs within regions, where the gradient is small, and not across region boundaries where the gradient magnitude is high. This approach was first proposed by Perona and Malik [18] and is now widely used. It was applied to Magnetic Resource Imaging (MRI) data in a seminal paper by Gerig *et al.* [19] and the algorithm is included in many software libraries, providing a reliable means of enhancing image quality. For example, it has recently proved successful in application to ultrasound data which typically contains speckle noise [20].

The next step is to apply a segmentation algorithm to identify different parts of the anatomy of particular interest. This will label voxels with an identifier indicating the type of material. The process typically remains semi-automatic, with user guidance needed in order to help correct identification. Indeed, segmentation is often the major bottleneck in clinical applications—it takes a long time and the results are often hard to reproduce because of the user involvement.

Segmentation is a major field of study, supported by a considerable body of literature, and given here is only a very brief overview. A useful survey paper is that by Pham *et al.* [21]. A typical strategy is to employ simple techniques initially, and seek a more sophisticated technique if these fail. Perhaps the simplest technique of all is thresholding, where the image is partitioned according to the pixel intensities. A single threshold will divide the image into two classes: pixels greater than, and less than, the threshold intensity—making it a useful technique, for example, in applications when two distinct classes of tissue are present (say cancerous and noncancerous).

Another fundamental approach is region growing. In its simplest form, a seed point in the image is manually selected, and an automatic algorithm 'grows' the relevant region up to its boundary, using some definition of connectedness of neighbouring pixels. An important and more sophisticated technique in the region-growing class is the watershed transform. This method comes from mathematical morphology. The intuitive idea is simple: imagine a landscape flooded by water, then watersheds are the dividing lines between domains of attraction of rain falling over the region, i.e., basins. In terms of image segmentation, the height of the landscape is interpreted as the magnitude of the gradient of the pixel intensities—the high dividing lines are thus sharp changes in image intensity. The paper by Roerdink and Meijster [22] gives a precise description of the transform, and the papers by Hahn and Peitgen [23] and Grau *et al.* [24] describe recent extensions and applications of the approach to medical imaging.

A further class of methods involves the user in guiding the segmentation process. In the 'Livewire' approach, a user selects an initial boundary point by cursor position; as the cursor is moved, the algorithm dynamically calculates the best boundary approximation as a curve from initial to current point. When the user is happy with the approximation, the current point becomes the new fixed point and the process continues, with the live wire snapping on to the boundary at each step. A good review of livewire, and the related livelane, is given by Falcao *et al.* [25].

Many segmentation techniques are based on the concept of deformable models where an initial shape is evolved to fit as closely as possible a feature in the image. Freedman *et al.* [26] provide a useful taxonomy of these techniques, based on the learned information they use. One important class of methods creates a shape model from a training set, and then modifies the model to match the image data. An example of this is the deformable M-rep approach [27]: this works by creating a set of points, or atoms, which form a medial representation (hence, M-rep) of an object. Each atom has not only position but also width and other geometric information sufficient to describe a solid region. Segmentation proceeds by a coarse-to-fine deformation of the M-rep, and successful segmentation of kidney from CT data sets and hippocampus from MRI data sets are described.

Another class of methods is based on a process of learning both shape and appearance. This approach models shape and appearance of an object from training data using Principal Component Analysis (PCA), and was pioneered by Cootes and Taylor (see [28] for an overview of their work). A recent variation is described by Freedman *et al.* [26] in which the calculation of pixelwise correspondence between model and image is replaced by a faster comparison of probability distributions, with examples of segmentation of prostate and bladder from CT images.

A further class of methods does not use any learned information. An important approach here is that of level sets, where an implicit representation of a surface is evolved under the control of a Partial Differential Equation (PDE) on a volume. Traditionally, this approach has been computationally expensive because of the need to solve the PDE, but recently research has shown how modern graphics hardware can be exploited to compute the level set models at interactive speeds [29]. The method is illustrated with examples of brain tumour segmentation.

Finally, there are some methods which are targetted at specific parts of anatomy that are difficult to segment. For example, Bartz *et al.* [30] describe approaches to the segmentation of the tracheo-bronchial tree of the lungs, using a combination of techniques. Persoon *et al.* [31] describe approaches to virtual colonoscopy (where there is a low signal-to-noise ratio) in which noise removal is integrated into the visualization process.

3.2. Volume visualization

A simple approach to visualizing a volume is to visualize a series of slices, either parallel to one of the faces of the volume, or in an oblique direction. This is often called multiplanar reformation (or MPR) and is arguably the most popular visualization technique in clinical practice [32]. Radiologists are trained to move through the slices and to recognize branching patterns through this process. Their experience in moving through 2D slices in this manner allows them to build a 3D mental model of the real anatomy. One difficulty with this approach is that branching structures of interest such as blood vessels are nonplanar and therefore can be difficult to follow. A recent idea is to use curved planar reformation (CPR), where a 'curved' slice following the trajectory of a vessel is presented [33]. Note, however, that this requires prior identification of the centreline of the vessel and so significant effort is required to create CPR visualizations.

While MPR is commonly used in practice, there are situations where a 3D view will give the radiologist valuable insight—for example, when patients have an unusual or complex anatomy or pathology. This has sparked a very active field of research among computer scientists, to develop fast and effective ways of presenting 3D medical visualizations. This is the subject of the remainder of this section, which assumes the data are in the form of a 3D volume, and more precisely, as a rectilinear grid of voxels.

Many medical applications can take advantage of surface extraction algorithms to extract explicitly an intermediate representation that approximates the surface of relevant objects from a volume data set [34], for example, to extract the surface of bones from a CT dataset. The surface extracted corresponds to a specified threshold value. The surface is extracted as a polygonal mesh, typically a triangular mesh, which can be rendered efficiently by graphics hardware (Figure 1).

The most popular algorithm in use is Marching Cubes [35], or one of its later variations which improve the accuracy of the surface representation (see, for example, [36]). The idea is to process the data cell-by-cell, identifying whether a piece of the surface lies within the cell, and if so, constructing an approximation to that surface. For large data sets, it is important to be able to quickly identify cells that contribute to the surface: current research on fast identification of isosurface location can be found in [37] (contour trees, where the isosurfaces are traced from cell-to-cell starting from a specially constructed seed set); [38] (span space, where cells are identified as points in a 2D space of maxima and minima, from which cells containing portions of an isosurface can be efficiently extracted); and [39] (interval trees, where the set of intervals—maximum and minimum of each cell—are sorted in a tree structure for efficient searching). A drawback with span space and interval trees is the high storage overhead of the data structures used to sort the cells. Brodoloi and Shen [40] use compression in order to reduce this overhead, but there is a cost in terms of false positives in the search and in added complexity. Recent work [41] describes a simple but effective span space approach based on bucket sorting of the cells.

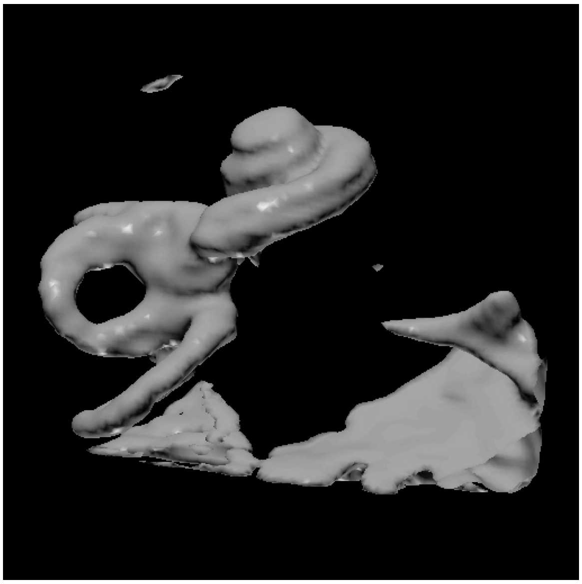

Figure 1: *Isosurface of a human cochlea generated from MRI data (image courtesy of Paul Hanns and the Manchester Visualization Centre, University of Manchester).*

Note that for medical applications isosurfacing is most effective for large bony and vascular structures imaged using CT, where the structure is sufficiently large and well defined in order to be extracted as a surface from the data. For smaller structures, isosurfacing is more problematic because a significant number of voxels will lie on the boundary of the structure, and there is a 'partial volume' effect where a voxel is composed of more than one material. Similarly, many structures such as soft tissue or lesions cannot be adequately extracted by isosurfacing, and indeed, it can be difficult to obtain meaningful results using isosurfacing on MRI data.

In (direct) volume rendering, as opposed to explicit surface extraction, voxel-based data are directly visualized without extracting an intermediate surface representation. Figure 2 shows an example of a volume rendering obtained from CT data.

A 'transfer function' is used to map physical fields of a given volume data set to optical properties, such as opacity

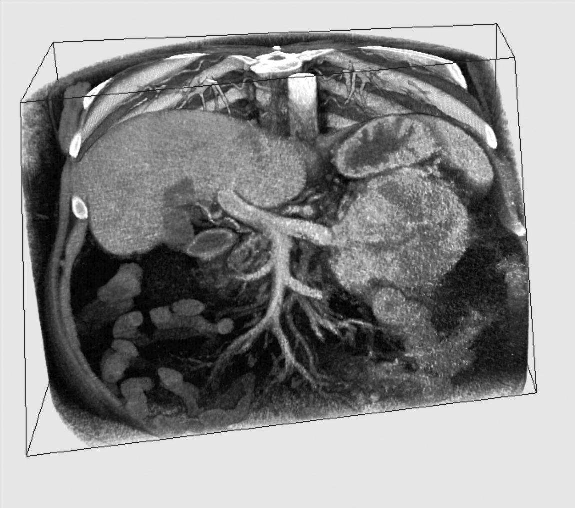

Figure 2: *Volume rendering generated from CT Data Scan showing the relationship between the confluence of the superior mesenteric, splenic and portal veins with a large tumour.*

and colour (although typically just grey scale values are used). The resulting entity can be thought of as a multicoloured gel.

There are four basic approaches to direct volume rendering. High quality, at the cost of compute time, is provided by ray casting and splatting; lower quality, but faster, is provided by shear-warp rendering and texture-based methods. Ray casting creates an image by casting a ray for each pixel into the volume, and compositing the light reflected back to the viewer from a set of samples along the ray, taking into account the colour and opacity at each sample. The seminal paper is that of Levoy [42]; a modern reference indicative of the state of the art in ray casting is the paper by Grimm *et al.* [43]. In splatting voxels are 'thrown' at the image in a forward projection, forming a footprint, and the result is accumulated in the image plane (see [44] for original paper, and [45] for more recent algorithmic efficiencies). In the shear-warp technique, speed-up is achieved by first aligning (using a shear) the volume and the viewing direction so that a line of voxels can project directly to a pixel, and secondly, compensating for the first transformation by an image-warp transformation (see [46] for original paper and [47] for a recent reworking). In texture-based methods, the volume is seen as a 3D texture and slices are projected and composited using graphics hardware (see [48] for initial ideas on texture-based volume rendering). Indeed, the increasing power of graphics hardware is a major influence on algorithm design—see, for example, combined ray casting and textures [49]. A detailed review of this area (including a section on dealing with large data volumes) is contained in the SIGGRAPH tutorial [50]. This area is discussed further in Section 5.8.

However, despite the potential of volume rendering, it is arguably more commonly used in the research laboratory than in everyday clinical practice. Research continues therefore to make volume rendering more effective. One strand of research is transfer function design. Rather than make colour and opacity depend only on one variable (voxel data value), multidimensional transfer functions have been proposed. Indeed, even in the early work on volume rendering, the transfer function depended on both value and gradient, in order to highlight boundary surfaces [51]. More recently, curvature-based transfer functions have been proposed [52], but the usability of multidimensional transfer functions remains an issue—particularly, as they introduce an extra layer of complexity. As a reaction to this, recent research has sought to return to a simpler histogram-based approach to transfer function design [53].

A variation on volume rendering has proved successful in medical visualization, particularly for the visualization of blood vessels from MR imaging. This is maximum intensity projection (MIP). It is based on images obtained after a contrast agent is introduced into the blood stream. The data values of vascular structures are then higher than surrounding tissue. Such examinations are referred to as either CT or MR angiography. By modifying the ray casting technique to pick out the voxel of maximum intensity (rather than compositing contributions from all voxels as would normally be done), we obtain a fast, effective and robust technique [54]. MIP does have some drawbacks however. As the image contains no shading information, depth and occlusion information is missing. Small vessels can tend to disappear in front of large bright vessels, due to the partial volume effect mentioned earlier. In addition, bones can have similar (or higher) intensity values than contrast-enhanced blood vessels, and so it is necessary to segment out the bones before applying MIP. A variation on MIP is closest vessel projection [55,56], or Local MIP [57], where along each ray the first local maximum above a user-defined threshold is used.

A number of special techniques have been developed for the visualization of anatomical tree structures. Hahn *et al.* [58] derive a symbolic representation of vascular structures and create a visualization of the structures as a smoothed sequence of truncated cones. Oeltze and Preim [59] likewise start with the assumption of a circular cross-section and create convolution surfaces from a skeletal representation of the vessels.

A strand of research has sought to improve our perception of volume rendered images. For example, the 'volume illustration' work of Ebert and Rheingans has applied non-photorealistic principles to improve perception through illumination effects and enhancement of features [60]. More recent work in this direction includes that of Hadwiger *et al.* [61], where different objects within a volume can be rendered in different ways; Bhagavatula *et al.* [62] where settings for volume rendering algorithms are input in an expressive manner; and Bruckner *et al.* [63] where opacity is reduced in less important data regions.

Virtual endoscopy has stimulated much research in volume rendering [64]. A particular rendering technique that has proved successful in virtual endoscopy is the first-hit ray casting approach of Neubauer *et al.* [65].

3.3. Image fusion

With the abundance of medical imaging technologies and the complementary nature of the information they provide, it is often the case that images of a patient from different modalities are acquired during a clinical scenario. To successfully integrate the information depicted by these modalities, it is necessary to bring the images involved into alignment, establishing a spatial correspondence between the different features seen on different imaging modalities. This procedure is known as *Image Registration*. Once this correspondence has been established, a distinct but equally important task of *Image Fusion* is required to display the registered images in a form most useful to the clinician and involves the simultaneous rendering of different modalities. Multimodal rendering is an active research area that includes 3D direct volume rendering of multimodal data sets, analysing how and where data fusion should occur in the rendering pipeline, and how rendering algorithms can be adapted and optimized for multimodal data sets. A possible framework for fusion methods and multimodal rendering can be found in [66].

Image registration techniques have been applied to both monomodality and multimodality applications. Registration of images from multiple modalities is generally considered a more difficult task. Nevertheless, there are significant benefits to the fusion of different imaging modalities. For example, in the diagnosis of epilepsy and other neurological disorders, it is common for a patient to undergo both anatomical (MRI, CT, etc.) and functional (PET, fMRI, EEG) examinations. Image guided surgery also makes extensive use of pre-operatively acquired imaging data and intra-operatively generated surface imaging and/or ultrasound for procedure guidance.

Various classifications of image registration methods have been proposed [67–69]. Classification criteria include dimensionality (2D-to-2D, 3D-to-3D, 2D-to-3D), domain of the transformation (global or local), type of the transformation or degrees of freedom (rigid, affine, projective, nonlinear), registration basis (extrinsic, intrinsic, nonimage based), subject of registration (intra-subject or inter-subject) and level of automation (automatic, semi-automatic).

A large body of the literature addresses 3D-to-3D image registration, and it is certainly in this area where techniques are most well developed. 3D/3D registration normally applies to tomographic data sets, such as MRI or CT volumes. Examples include the registration of time-series MR images in the early detection of breast cancer [70,71]. Registration of separate 2D tomographic slices or intrinsically two-dimensional images, such as X-ray, are possible candidates for 2D-to-2D registration. While the complexity of registration is considerably reduced compared to higher dimensionality transformations, examples of 2D-to-2D registration in the clinical domain are rare [72]. This has been attributed to the general difficulty in controlling the geometry of image acquisition [69]. There are generally two cases for 2D-to-3D registration. The first is for the alignment of a 2D projection image with a 3D volume, for example, when aligning an intra-operative X-ray or fluoroscopy image with a pre-operative MR volume. Secondly, establishing the position of one or more 2D tomographic slices in relation to an entire 3D volume is also considered 2D-to-3D registration [73].

Rigid registration generally describes applications where the spatial correspondence between images can be described by a rigid-body transformation, amounting to six degrees of freedom (three rotational, three translational). Rigid techniques apply to cases where anatomically identical structures, such as bone, exist in both images to be registered. Affine registration additionally allows for scaling and skewing between images, yielding twelve degrees of freedom. Unlike rigid-body transformations that preserve the distance between points in the object transformed, an affine transformation only preserves parallelism. Affine techniques have proven useful in correcting for tomographic scanner errors (i.e. changes in scale or skew). Nonrigid registration approaches attempt to model tissue deformation which can be the result of pathological (e.g. tumour growth), normal (e.g. aging), external (e.g. surgical intervention) or internal (e.g. development) processes. It entails transformations with considerably larger numbers of degrees of freedom, allowing nonlinear deformations between images to be modeled [74–76] and deformation fields such as those resulting from intra-operative brain shift to be estimated [77]. A popular approach is to obtain an initial coarse approximation using rigid registration followed by a local nonrigid refinement step. Nonrigid registration is highly relevant for medical virtual environments dealing with unconstrained highly deformable organs (e.g. heart, liver, etc.) affected by intrinsic body motion (breathing, bowel, etc.) and/or extrinsic motion (tool–tissue interactions). In general, nonrigid registration is a more complex problem, with fewer proven techniques and difficult to achieve in real time with sufficient accuracy and reliability. Novel approaches employing acceleration techniques based on advanced hardware architectures are being explored to tackle these issues [78].

Landmark-based registration techniques use a set of corresponding points in the two images to compute the spatial transformation between them. Landmark registration approaches are divided into *extrinsic* methods, based on foreign objects introduced to the imaged space, and *intrinsic* methods, i.e. based on the information provided by the natural image alone. Extrinsic approaches are common in image-guided surgery [79–81]. Intrinsic methods [82–84] instead rely solely on anatomical landmarks which can be elicited from the image data.

Rather than develop correspondence between sets of points within two images, surface- or segmentation-based methods compute the registration transformation by minimizing some distance function between two curves or surfaces extracted from both images. Segmentation-based registration methods have been used for both rigid and nonrigid applications [85]. The iterative closest point (ICP) algorithm is one of the best known methods for 3D-to-3D registration [86,87]. It is based on the geometry of two meshes and starts with an initial guess of the rigid-body transform, iteratively refining it by repeatedly generating pairs of corresponding points on the meshes and minimizing an error metric.

Voxel similarity-based algorithms operate solely on the intensities of the images involved, without prior data reduction by segmentation or delineation of corresponding anatomical structures. Common similarity measures include correlation coefficient, joint entropy and variants of mutual information [88–91]. Voxel property-based methods tend to use the full image content throughout the registration process.

As noted in [68,69], while there exists a large body of literature covering registration methods and a number of open-source toolkits [92,93] offering a variety of registration tools, the technology is presently used very little in routine clinical practice, with one or two major exceptions that tend to rely on the more basic landmark-based and ICP approaches. This can be attributed to several likely factors. Firstly, image registration is a relatively new research area. Secondly, image registration is only one component of an entire medical image analysis framework, whose other components (such as segmentation and labelling) in many cases may be insufficiently reliable, accurate or automated for everyday clinical use [69]. Also, there are numerous logistical problems involving the integration of analogue and digital data obtained from different imaging modalities.

3.4. Soft-tissue modeling

The purpose of soft-tissue modeling is to enable the simulation of tissue behavior. This is required in a variety of applications including surgical simulators for training, intra-operative deformation modeling and surgical planning. In general, soft-tissue modeling algorithms can be classified as either *geometrically-based* or *physically-based*. With geometrically-based modeling, the shape of an object is adjusted by changing the position of some control points, or by adjusting the parameters of an implicit function defining the shape. A typical example of this type of technique is free-form deformations (FFD) [94], where the object is embedded into a lattice of a simple shape. Deformation of the lattice causes a consequent deformation of the object. These methods are often fast, but the object deformation is carried out indirectly and may bear little or no resemblance to the physically plausible deformation. Recent research has focussed on improving the user interaction with the objects to allow direct manipulation—see [95] for example.

In contrast, physically-based modeling methods attempt to mimic reality by embedding the material properties and characteristic behavior of the object into a suitable model. Physics laws govern the movement of object elements. These methods can explicitly represent the real life, dynamic behavior of objects. Object deformation is applied in an intuitive manner, typically by interacting directly with the object. One of the most widely used approaches in physically-based modeling is the mass-spring technique. The pioneering work was by Terzopoulos and co-workers in the context of facial modeling [96,97] and this has been extended to medical and surgical applications [98–100]. The objects are constructed as a set of mass points, or particles, connected by damped springs. The particles move based on the forces applied. This is a relatively fast modeling technique and easy to implement. The method has been applied to cutting as well as deformation [101]. Fast numerical techniques for solving the differential equations which occur in mass-spring (and FEM, see below) are described by Hauth and Etzmuss [102]. Reported difficulties with the mass-spring technique [103] include unrealistic behavior for large deformations, and problems with harder objects such as bone.

A different approach is a fast deformable modeling technique, called 3D ChainMail [104–107]. A volumetric object is defined as a set of point elements, defined in a rectilinear mesh. When an element of the mesh is moved, the others follow in a chain reaction governed by the stiffness of the links in the mesh. Recent work has extended the 3D ChainMail algorithm to work with tetrahedral meshes [108]. The resulting algorithm is fast enough to be used for interactive deformation of soft tissue, and is suitable for a Web-based environment.

Compared to the mass-spring technique, the finite element method (FEM) is a physics-based approach [109] capable of producing results of greater accuracy as it aims to simulate the real underlying physics. An FEM model subdivides the object into a mesh of simple elements, such as tetrahedra, and physical properties such as elastic modulus, stresses and strains are associated with the elements. The equilibrium equations for each element, which govern their movement, are solved numerically. Due to its highly computational nature, this method is typically used in applications such as surgical planning [110], where real time interaction is not required; it is not presently practical for most interactive applications, but solutions are starting to appear, for example, in suturing [111].

While the ability to interactively simulate the accurate deformation of soft tissue is important, a system that does not include the capability to modify the simulated tissue has limited utility. Considerable efforts have been made by the research community to improve on the modeling and

Figure 3: *Adaptive soft-tissue modeling with cutting and grasping.*

Figure 4: *Capacitance sensor with conditioning unit.*

interactivity of soft tissue. Most of these efforts have concentrated on pre-computing deformations or simplifying the calculations. There have been several attempts at locally adapting a model in and around the regions of interest [112,113]. However, these models rely on a pre-processing phase strongly dependent on the geometry of the object, with the major drawback that performing dynamic structural modification becomes a challenging issue, if not impossible. More recently, a volumetric model that adapts the resolution of the mesh by re-tesselating it on-the-fly in and around the region of interaction was presented in [114]. This adaptive model does not rely on any pre-processing thus allowing structural modifications (Figure 3).

An important issue in surgical simulation is the detection of collision between deformable objects—for example, collisions between deformable organs, or between surgical instruments and organs. A review of this area can be found in [115].

There are two excellent surveys on physically based modeling in computer graphics generally: the SIGGRAPH tutorial by Witkin and Baraff [116] and the Eurographics 2005 STAR report by Nealen *et al.* [117]. A gallery of applications in surgical planning can be seen at [118], with a recent application involving soft tissue modeling in craniomaxillofacial surgery being described in [119].

4. Physical Properties of Organs, Tissues and Instruments

The skills of laparoscopic surgery are mostly visuo-spatial and the sense of touch or haptics is of lesser importance than in Interventional Radiology (IR) which relies heavily on it. Haptics is therefore likely to be of greater importance in attaining 'reality' in simulation of IR procedures, though this is as yet, unproven. The force feedback ('feel') delivered in VE simulator models is generally an approximation to a real procedure, as assessed by experts. In virtual environment training systems, haptics based on real procedural forces should allow a more authentic simulation of the subtle sensations of a procedure in a patient. The nature of tissue deformation, is however, nonlinear and this introduces a need to acquire information from direct study [120,121] of static and dynamic forces in tissues. In evaluating needle puncture procedures, for example, *in vitro* studies are essential for detailed study of the physical components and effects of overall summated forces. Thus the biomechanical properties, and consequently much of the 'feel', of a needle puncture derive from steering effects, tissue deformation, and from resultant forces at the proximal end of the needle representing integration of forces from the needle tip (cutting, fracture) and the needle shaft (sliding, friction, clamping, stick-slip) [121,122]. Due to the different physical properties of living tissues, *in vitro* data require verification by *in vivo* measurements [123]. While there is a dearth of literature on measurement of instrument–tissue interactions *in vivo*, important studies have been performed during arthroscopy [124] and in needle puncture in animals [123]. Until recently there were few devices available for measurement of instrument forces *in vivo* in humans, unobtrusively: flexible capacitance pads (Figure 4) present a novel opportunity to collect these data *in vivo* and calibration *in vitro* has shown that output is stable and reproducible [125]. *In vivo* work has gone on to show the feasibility of mounting capacitance pads on an operator's fingertips, beneath sterile surgical gloves, to obtain real-time output data during sterile procedures in patients [126].

Needle guidance by ultrasound imaging during procedures in patients affords a method of correlating the integrated force values obtained from the sensors with the anatomical structures traversed by the needle [127] providing a verification of, published *in vitro* studies and also of simulator model output forces, enabling re-evaluation of underlying mathematical models [120]. Ultimately, combination of visuo-spatial realism in VEs based on patient specific data, with haptics derived from *in vivo* studies, will deliver an authentic learning environment to the trainee.

5. Applications and Innovative Projects

5.1. Educational tools

Traditional teaching of human anatomy involves dissection. The benefits include a spatial understanding which is difficult to glean from textbook demonstrations. Traditional methods are now less commonly used, having been largely replaced by problem-based learning scenarios and pre-dissected specimens. Yet there are developing deficiencies in the anatomical knowledge of today's medical graduates, and such information is vital to their future practice in specialties such as surgery and radiology. A substitute for the traditional, interactive methods exists in the application of virtual environments where anatomy can be explored in 3D, providing new educational tools ranging from three-dimensional interactive anatomical atlases to systems for surgery rehearsal [128]. The facility to develop these methods has been drawn from data sets such as the Visible Human Project [129]. The Visible Human male is around 15 GB and the female 45 GB in size. There are now a number of Chinese [130] and Korean [131] variants with up to 1.1 TB of data, yielding vastly improved image resolution (0.1 mm) but some challenges for data manipulation and segmentation processes. The exemplar product in this category is the VOXEL-MAN family of applications developed by the University of Hamburg [132], based on the original male data set from the Visible Human project. Organ relationships, gunshot tracks, related physiology and histology can be learnt in a highly interactive manner. Anatomical structures can be explored and labelled, and can be made available in cross-sectional format [133], segmented, isolated organ structure, or can introduce functionality, as in the contractility of muscle. Structures can be removed sequentially, returning to the learning advantages of traditional dissection. In addition, feedback to the trainee can be provided as to developing knowledge levels.

5.2. Diagnostic aids

CT and MR scanners can provide 3D data sets of 1000 image slices or more, but only around 20 to 60 hard copy images can be displayed on a typical light box. Consequently, thorough assessment of a complete volume data set requires mental reconstruction to provide the viewer's own 3D interpretation from those relatively few 2D images. Today's radiologists, when reporting medical imaging, will generally use a proprietary workstation, with much reporting work based on review of axial images, often using interactive movie scrolling to rapidly evaluate the multiple slices. Exploring off-axial (multiplanar) 2D reconstruction can also provide valuable demonstration of anatomical relationships. Other visualization methods can be deployed, such as sliding thin-slab maximum intensity projection (STS-MIP) [134], which can be considered as part way between a 2D slice and a true 3D view. However, it is the latter that has the significant advantages of being able to present all of the image data to the clinician at the same time and in an intuitive format. A volume presentation or a maximum intensity projection, also allows measurements to be performed in other than the axial plane.

Moreover, this 3D representation could be significant to a nonspecialist in radiology, such as a surgeon. The techniques used are described in Section 3. They allow the clinician to see the internal structure and the topology of the patient's data. Figure 2 is a good example of a volume rendering generated from a CT data scan showing the relationship between the confluence of the superior mesenteric, splenic and portal veins with a large tumour. Figure 5 is another interesting example, which was obtained by segmenting and volume rendering data obtained from a Siemens Sensation 16, Multidetector Array CT scanner with intravenous contrast enhancement for maximum arterial opacification The study was post-processed using a Leonardo workstation with Vessel View proprietary software. The anatomy depicted is unusual and shows separate origins of the internal and external carotid arteries from the aortic arch.

Note that all of the major scanner manufacturers (Philips, GE, Siemens, etc.) already provide 3D visualization support and the value of 3D as a diagnostic aid is being demonstrated with excellent results [135]. Companies such as Voxar (Edinburgh, UK) and Vital Images (Plymouth, MA) have been successfully marketing 3D volume rendering software technologies for several years. Such displays are particularly valuable in the planning of the endoluminal management of aneurysm where accurate lumen length measurements of the aorta are required to determine the correct size of endograft needed for a particular patient's anatomy. In such cases, the use of 3D representation replaces invasive catheter-based measurement, reducing risk, discomfort and cost. Nevertheless, use of 3D visualization techniques today is still largely confined to the workstations in radiology departments.

5.3. Virtual endoscopy

In video or optical endoscopy [136], an endoscope (made from a fibre optic tube) is inserted into the patient body through a natural or minimally invasive opening on the body. It is an invasive process and only a limited number of structures can be viewed by the camera positioned at the tip of the endoscope. Virtual Endoscopy (VEnd) is a visualization technique that provides the same diagnostic information to the clinician by creating a 3D model from a medical scan of the patient. VEnd is not invasive (except for the radiation produced by a CT scan) and enables visualization of any structure, providing for interactive exploration of the inner surface of the 3D model while being able to track the position of the virtual endoscope. The clinician can also record a trajectory through the organ for later sessions, and create a movie to show to other experts.

A disadvantage of VEnd is the lack of texture and colour of the tissue being examined. However, if VEnd is combined with other tools to highlight suspicious areas [137,138], the clinician can have access to additional information during the diagnostic phase. Other potential limitations of VEnd include the low resolution of images, the creation of a 3D model can also be time consuming and the use of 3D software is often nonintuitive and frustrating. Most of these limitations are being overcome with higher resolution scanners, improved 3D software and experienced operators.

Clinical studies have shown that VEnd is useful for surgical planning by generating views that are not observable in actual endoscopic examination and that it can also be used as a complementary screening procedure or control examination in the aftercare of patients [139–142]. Of particular note are studies that have investigated the effectiveness of virtual endoscopy [143,144], which augur well for adoption of this technique. Virtual colonoscopy has received most of the attention and research effort [145,64] with a recent study finding that its cost can be up to half of that of standard colonoscopy [146].

The steps involved in building a VEnd system include: data acquisition, pre-processing and detailed segmentation, calculation and smoothing of the path for fly-through animation, and volume/surface rendering for visualization. Different approaches to each of these steps address different VEnd applications such as colonoscopy, bronchoscopy or angiography. Bartz [64] provides an excellent overview of clinical applications of VEnd and highlights the current research topics in the field. With improvements currently being made in the spatial and temporal resolution of imaging modalities, and new techniques for automated path definitions [147], improved tools for user orientation during the virtual endoscopy, and improved reconstruction speeds allowing real time fly throughs at high resolution, VEnd will undoubtedly play an increasing role in the future of whole body imaging.

5.4. Treatment planning aids

The use of medical imaging and visualization is now pervasive within treatment planning systems in medicine. Typically such systems will include a range of visualization facilities (e.g. virtual resection [148]), fusion of images, measurement and analysis planning tools and often facilities for the rehearsal of operations and risk analysis. For most image guided surgery systems (see Section 5.5 for neurosurgery planning), except for some trauma surgery where surgical planning is based on intra-operative imaging, surgery is planned preoperatively. Treatment planning systems are essential to such areas as hepato-pancreatic surgery, spinal surgery, maxillofacial surgery, radiotherapy treatment planning, neurosurgery, etc. Liver surgery planning is one of the most advanced fields in this area (including use of augmented reality—see Section 5.7) and is gaining clinical importance, with examples of both pre-operative [149,150] and intra-operative systems [151]. Within orthopaedic surgery planning and navigation, aids can be classified as CT-based, 2D fluoroscopy based, 3D fluoroscopy-based and image-free. The special issue of the Injury journal titled 'CAOS and the integrated OR' explains these approaches and gives details of treatment planning aids for tibial osteotomy, total hip replacement, spinal surgery, long bone fractures, anterior cruciate ligament (ACL) reconstruction, etc. [152]. To date, the focus has been on providing greater precision when targeting treatment. The next challenges will be to provide more effective user interfaces [153], real time systems and to inclusion of functional data.

The benefits of using 3D visualization techniques for radiotherapy treatment planning have been reported for many years [154,155]. A well-used technique is called 3D conformal radiation therapy (3DCRT), in which the high-dose region is conformed to be close to the target volume, thus reducing the volume of normal tissues receiving a high dose [156]. More recently, 3DCRT has evolved into *intensity modulated radiotherapy treatment* (IMRT) of cancer tumours [157] which relies on a sophisticated visualization planning environment. In IMRT the tumour is irradiated externally with a number of radiation beams (typically 5–9). Each beam is shaped so that it matches the profile of the tumour. The intensity of radiation is varied across the beam using a multileaf collimator (MLC). The shape and intensity of the beam is computed by an inverse planning optimisation algorithm. The goal of this algorithm is to provide a high radiation dose to the tumour while sparing normal tissue surrounding it. Particular attention is paid to reducing radiation to critical organs (e.g. spinal cord, heart, pituitary glands, etc.). Planning involves visual identification of tumour(s) and critical organs, selecting the beam directions and defining the objective function and penalties for the planning optimisation algorithm. The radiation plan is then checked by viewing overlays of radiation dose on patient anatomy and various radiation dose statistics. Revised plans are produced by adjusting the parameters to the planning optimisation algorithm until a satisfactory solution is obtained. Comparisons of conventional techniques, 3DCRT and IMRT for different clinical treatments have been made [158] and verify that the techniques that use 3D visualization do indeed reduce dose. Researchers at the University of Hull have used a virtual environment to further enhance radiotherapy planning (Figure 6) by producing a full-scale simulation of a real radiotherapy room used for IMRT with visualization of patient specific treatment plans displayable in stereo-vision

Figure 5: *Volume rendering of unusual anatomy with separate origins of the internal and external carotid arteries from the aortic arch (indicated by arrow).*

Figure 6: *Virtual environment simulation of 3D radiation therapy treatment at the University of Hull.*

on a large immersive work wall. This simulation has been further developed as a tool for training radiotherapists in skin apposition treatment where a trainee uses an actual handheld pendant to control the virtual linear accelerator in the virtual environment [159]. A study has recently been completed to determine the benefits of immersive visualization for training radiotherapists.

5.5. Guiding aids

In the last twenty years, surgical techniques have undergone radical change. A growing number of procedures are now conducted using a *minimally invasive* approach, in which surgeons operate with instruments passed through small holes in the patient, not much larger than a centimetre in diameter. There are significant benefits to minimal invasion, such as reduced patient trauma, reduced blood loss and pain, faster recovery times and, as a result, reduced cost. However, by keeping the surgeon's hands out of the patient, we incur the cost of occluding the view of the surgical field. A high resolution miniature video camera inserted through an access port is depended upon to act as the eyes of the surgical team, who view the operation on monitors in the operating theater. A number of minimally invasive surgical systems now provide colour and dual-camera configurations for stereo-vision capability. There are limits to this approach, however. Traditional optical imaging technology cannot be used to see within organs, or around corners, and so the range of treatments that can be accommodated in this way is restricted. If minimally invasive techniques are to prevail over their more traditional techniques across more surgical interventions, new and more capable means of guidance are necessary [160].

Image-guided surgery is the term used to describe surgical procedures whose planning, enactment and evaluation are assisted and guided by image analysis and visualization tools. Haller *et al.* describe the ultimate goal of image-guided surgery as 'the seamless creation, visualization, manipulation and application of images in the surgical environment with fast availability, reliability, accuracy and minimal additional cost' [161]. Surgical systems incorporating image guidance for neurological or orthopaedic interventions are available from a number of manufacturers [162–164].

All neurosurgical interventions require considerable planning and preparation. At the very first stage, anatomical MR or CT images are taken of the patient's head and brain, and transferred to a supporting IT infrastructure. Specialist hardware is used to reconstruct the scanned images both as two-dimensional and three-dimensional views of the data. Pre-surgical planning tools allow a thorough review of all images obtained. Furthermore, the surgeon is able to interact with a three-dimensional visualization of the data by changing the viewing angle, adjusting the transparency and colouring of different brain structures, removing skin and bone from the image and so on. In addition, virtual instruments can be manipulated within the simulation, and their position and trajectory through the skull and brain displayed in real time.

These features provide considerable assistance in the general decision-making process of surgical planning.

In the operating theater, a different type of multimodal registration is employed. Intra-operative guidance systems work by establishing a spatial correspondence between the patient on the operating table and pre-operative anatomical data, enabling the localization of surgical instruments and the patient's anatomy within the three-dimensional space of the operating theater. With registration between tool, patient and theater established, when an instrument is placed on the patient, the exact location of the instrument tip is projected onto anatomical images displayed on a computer monitor. Herein lies the essence of image guidance, enabling the surgeon at any time to 'see' the exact location of his instruments in relation to both the anatomy and the surgical plan. Just as in the planning stage, critical functional information obtained pre-operatively can be added to these images. In addition, intra-operative ultrasound images registered with the original MRI data can be used to update the position of the tumour and surrounding anatomy [165]. While a coarser representation than that provided by intra-operative MRI [166], ultrasound data obtained during the intervention can be used to detect and account for tissue shift that might otherwise render the registration dangerously inaccurate, and at relatively little extra cost [161,166]. Reliability, accuracy and the invariable presence of organ deformation affecting any initial alignment are key challenges of intra-operative registration and thus of image guidance.

5.6. Skills training aids

The teaching of medical knowledge has for eons been a tradition where a major part of this process has been the teaching of clinical skills in an apprenticeship. Training on patients is associated with discomfort and provides limited access to the range of training scenarios required for full experience [167]. This creates difficulty training in a time efficient manner [168]. In surgery, and in particular, laparoscopic surgery, the requirement for completely new, specialist clinical skills, has driven the development of novel solutions to training. Skills boxes and other bench models [11] have been shown to accelerate the acquisition of skills. Fixed models and mannequins have also been shown to be of value in training skills in interventional radiology (IR), for example, using vascular models produced by rapid prototyping to train in catheterisation skills [169]. There are, however, few models simulating the broader scope of IR procedures. The cost of rapid prototyping is a limiting factor and such models lack realistic physiology: once constructed, their anatomical content cannot be easily altered. For needle puncture procedures, fixed models lack robustness and are destroyed by repeated needle punctures. While animal models are used to train IR skills [170], and also incorporate physiology, they are expensive, with anatomical differences [171], a lack of pathology and, in the UK, of political acceptability.

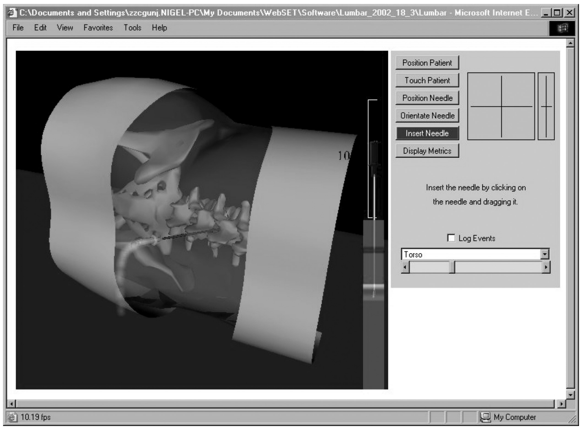

Figure 7: *Simulator for lumbar puncture training (from the WebSET project).*

VEs have a major role to play in the future of skills training. The value of a VE in this context was first demonstrated with the development of a laparoscopic simulator model to train skills using, initially, simple geometric shapes, with subsequent development of more realistic visual displays and haptics. This showed for the first time that VE could be a highly effective training tool [13].

Underpinning the acceptance of VE in teaching clinical skills is the process of validation, where objective studies are used to demonstrate resemblance to a real world task (face validity) and measurement of an identified and specific situation (content validity). Simple tests, such as construct validation, may show that an expert performs better than a novice in the model under test, while more complex, randomized controlled studies (concurrent validation) correlate the new method with a gold standard, such as apprenticeship [172]. Ultimately, predictive validity will evaluate the impact on performance in a real procedure. Clearly the reliability of the validation process, across evaluators (inter-rater reliability), and between repeated tests (test-retest reliability), must also be confirmed. There is a great deal of activity in this area. One example is the lumbar puncture courseware developed in the WebSET project (Figure 7) [173]. Validation studies have shown that the multimedia component of the lumbar puncture courseware aids in the acquisition of knowledge while the VR simulation further aids in the development and practise of technical skills [174].

At present the state of the art in VEs [173,175–179] and haptics has led to considerable work in surgical simulation and some work in IR. Some academic and commercial models simulate catheterization [180–183,15] and needle puncture procedures [182,184,185]. In some simulator models, development has been based on a Task Analysis [10,182] and this is essential in the development of clinically valuable, simulator models of interventional radiological procedures. Virtual environments will ideally be based on patient-specific imaging data with sem-iautomatic, or perhaps automatic, segmentation of vascular anatomy, organs and ducts. While many skills can be acquired in simple geometrical shapes, the clinical 'reality' needed to simulate the patient environment will entail incorporation of realistic physiology, physiological motion and physics-based tissue deformation characteristics. Within a decade, challenging virtual training scenarios will be reproduced in an *ad-hoc* manner, using real cases, simulating complications and critical events and introducing a facility for pre-treatment procedure rehearsal. Scenarios will be viewed by a trainee as simulated fluoroscopy, CT and ultrasound and, using haptic interfaces, procedures of graded complexity will be practiced. For the first time, it will be possible to experience the consequences of technical mistakes, and complications, in complete safety. Virtual environments will confer an opportunity to train more rapidly, in a variety of case scenarios, maintaining higher skill levels, even where throughput is low. Metrices derived by subject matter experts from the fundamental task analysis, will provide the basis of objective assessment of performance and feedback to the trainee. On a cautionary note, uptake of the training aspects must be focussed within curricula to avoid the risk of divorcing technique from clinical context and the hazards of ionizing radiation.

Incorporation of objective measures of performance introduce reproducible skills assessment which could form a part of examination, certification and revalidation. The objective assessment methods that are available in relation to the assessment of surgical training [6–9,12] are absent from IR. Given that IR training is undergoing significant change, in both its ethos and the scope of its technology, there is a pressing need for the development and validation of a specific assessment methodology. The metrices derived will not only be applicable to objective assessment of performance in simulator models, but also in the clinical scenario, and may ultimately be incorporated into certification. It is essential, however, that this process is validated and that simulator development is documented and transparent. Currently, much of the activity in simulator model training of catheter procedures is industry driven and led. The metrices required for assessment of performance may, however, differ between training curricula and countries. Training organizations will therefore require, at the least, sight of the methodology, procedure descriptions, task analyses and metrices which constitute the development process of a simulator to be used in their curriculum. Even better would be the full involvement, or a lead role, of the credentialing and training organization in the process of simulator design and development.

5.7. Augmented reality

Augmented Reality (AR) superimposes computer-generated artifacts onto an existing view of the real world, correctly orientated with the viewing direction of the user who typically wears a suitable *head-mounted display* (HMD), or similar device [186]. AR is a technology growing in popularity, with applications in medicine (from as early as 1988 [187]), manufacturing, architectural visualization, remote human collaboration, and the military [188]. However, AR makes the already onerous systems integration tasks associated with VE systems even more difficult and so acutely highlights present limitations associated with spatial tracking and other VE peripherals.

Video cameras can be attached to the front of an otherwise normal (scene occluding) HMD and the video feeds mixed with the computer-generated data before being displayed to the user via the HMD's active display panels and optics. Such systems are referred to as 'Video-through' AR systems. The alternative is to use 'See-through' displays in which the synthetic computer-generated images are injected into the display to impinge on the user's view, for example, the Varioscope AR [189,190]. Such systems allow uninterrupted and unaltered views of the real world, which is necessary for user acceptance in medical applications. The central problem in AR is the correct alignment of the viewer's coordinate system (i.e. the position and orientation of the viewer's eyes, e.g. the endoscope) with the virtual coordinate system of the augmented images. Without establishing the spatial transformation between object and observer coordinate frames, it is impossible to correctly augment the view of a scene. This in itself is a registration problem (see Section 3.3) and, for AR to be effective, the registration must be both accurate enough to keep discrepancies between the real and virtual visual stimuli to a minimum, and fast enough to be used in real time. Latency or 'jitter' of the display of virtual objects in the real environment can degrade the quality of the AR experience severely. There are additional problems which must be addressed, such as occlusion, and radiometry i.e. matching the lighting models of real and virtual worlds.

AR is a relatively new technology, and as a result there are few commercial or clinical AR systems in widespread use. Of those that are currently available, interesting systems that attempt to fuse the visual and haptic displays from the user's perspective are being developed at CSIRO [191] and ReachIn Technologies AB [192], and at EVL [193]. It is likely that with increasing robustness of the technology, demonstrable clinical value and availability of improved HMD or other suitable hardware such as semi-transparent displays [194], use of AR will grow in those areas where the technology can succeed. In particular, we can expect an integration of AR with systems already being utilized for image guided surgery [195] (see also Section 5.4). Developments such as these are making use of commodity hardware and so reducing the cost of AR solutions. Unfortunately for many clinical applications, particularly those involving deformable anatomical structures, the registration problem still proves to be a difficult barrier to overcome. Many research groups are actively working on this and the other problems in AR identified above. For example, techniques have been developed to map intra-operative endoscopic video to 3D surfaces derived from pre-operative scans for enhanced visualization during neurosurgery [196]; and the correct occlusion handling of virtual objects with real scene elements [197].

In surgery, AR offers a natural extension to the image-guidance paradigm and offers great potential in enhancing reality in the operating room [198]. This 'X-ray vision' quality of AR is especially useful to the restricted-view realm of minimally invasive surgery. From the outside, we could augment an internal view of the patient based on pre-operative or intra-operative data, presenting the surgeon with a large and information-rich view of the surgical field while entirely preserving the benefits of minimal invasion. Clearly, the enhanced visualization characteristic offered by AR can be beneficial to a wide range of clinical scenarios.

One of the first medical AR systems was designed to display individual ultrasound slices of a foetus onto the pregnant patient [199,200]. Ultrasound data were combined with video from a head-mounted display. Initially, due to inaccurate tracking, the alignment between the ultrasound and video was poor and the foetus lacked 3D shape. However, the research group at UNC has improved the technology and is now developing the system for visualization during laparoscopic surgery [201]. The innovative work of this group also included an AR system for ultrasound-guided biopsy of breast lesions [202].

Many medical AR systems concentrate on the head. This is because the task of alignment and tracking of the patient is minimized, as the motion of the patient and organs is rigid and constrained. One such system is the Microscope-Assisted Guided Interventions (MAGI) [203,204]. The system projects a reconstructed 3D vessel tree from pre-operative data onto the surgical microscope view. The aim is to allow the surgeon to view structures that are beneath the observed surface, as if the tissue were transparent. Figure 8 is an image from MAGI being used to guide the surgeon when removing a tumour (acoustic neuroma). A moving texture map is used here as the display paradigm and very good 3D perception was reported by the surgeon. Recent work to develop the MEDical Augmented Reality for Patients (MEDARPA) workstation [205] is also of note as it uses augmented information that is located on a free-positionable, half transparent display, the 'virtual window" that can be positioned over the patient.

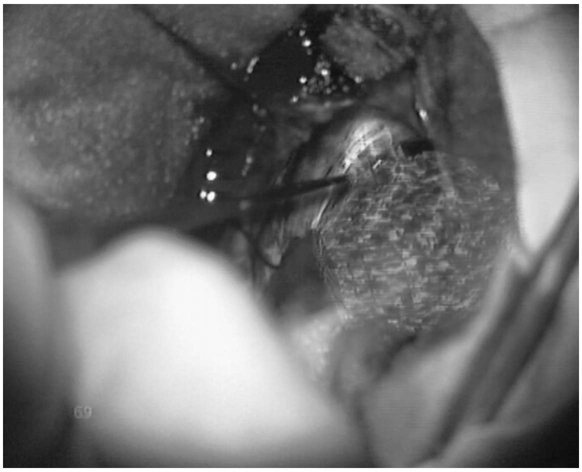

Figure 8: *Overlay of an acoustic neuroma after resection, showing how it had entered the internal auditory meatus—from the MAGI project (image courtesy of David Hawkes, UCL Centre of Medical Image Computing).*

5.8. Trends in high performance computing for medical applications

The ever increasing computational power and very modest cost of desktop PCs make them the ideal platform for running the majority of the applications discussed in this paper. The evolution of volume rendering hardware illustrates this point well (see also Section 3.2). A well- known hardware acceleration technique for volume rendering is to use texture mapping hardware [48]. Originally, this technique was available only on SGI workstations where the data are loaded directly into 3D texture memory. Subsequently, it has been adapted for the Graphics Processing Unit (GPU) found on all modern PC graphics cards, and often combined with use of per pixel, or fragment, shaders. Many groups are publishing work in this active research area [206–208], and a good overview of the field can be found in [50]. Texture data must be fetched via the Accelerated Graphics Port (AGP) or the PCI Express bus (for state of the art PCs) from the main memory of the PC and this can be a bottleneck. New developments in PC hardware will soon solve this, however, together with the use of a new breed of software algorithms that are addressing the problem of processing very large data sets on a PC architecture [209,50]. Special purpose hardware has also been designed for PCs [210], for example, the VolumePro 1000 card implements shear warp rendering and can deliver up to 30 frames per second for 512 cubed voxel data [211,212]. This performance can now be matched by the GPU on a high-end graphics card, and even an inexpensive PC can achieve real time for a 256 cubed voxel data set.

Despite their growing power, there are still instances where the computational demands of the application are too great for PC systems—for example, advanced image registration methods. A possible alternative that some research groups are investigating is the use of Grid computing, designed to allow end users transparent access to a wide variety of high performance computing facilities and data acquisition devices. A good medical example is from the CrossGrid project, which has demonstrated a Grid-based prototype system for planning vascular interventional procedures such as the treatment of arteriosclerosis [213]. The user manipulates a 3D model

Figure 9: *Remote volume visualization of patient data delivered to the operating theater (from the Op3D project) [216].*

of a patient's arteries to insert a bypass. The effects of this adaptation are then analyzed using an efficient mesoscopic computational haemodynamics solver for blood-flow simulations. This can be achieved in real time by using Grid resources. Remote radiation treatment planning has also been implemented by using high-speed communications and a supercomputer resource [214].

Also related to Grid computing is the idea of remote visualization, where an application is run on a visualization server and the results delivered in real time across the computer network to a client workstation. This is achieved by streaming the contents of the frame buffer on the server (where the graphics primitives are rendered) in a similar fashion to how MPEG movie files are streamed across the Internet. OpenGL Vizserver [215] from SGI was the first product to support remote visualization, and similar products are also available from IBM, and HP. Remote visualization has already been exploited for medical applications such as the example shown in Figure 9. Here, a volume visualization application is being used to aid with hepato-pancreatic surgery [216]. OpenGL Vizserver is being used to deliver the volume renderings of the patient data to a laptop client in the operating theater. The data sets being used in this application are too large to be processed on just the laptop (although with pre-processing this may soon become possible, for example, by converting the volume data into a hierarchical wavelet representation as described by Guthe *et al.* [209]). Visualization on the Grid will provide the opportunity to combine 3D visualization with advanced flow simulation, soft-tissue modeling, image fusion, etc. in real time.

6. Future Challenges and Conclusions

The above review demonstrates that medical applications using computer graphics techniques are becoming commonplace for education, diagnosis, treatment planning, treatment delivery and procedures training. Early attempts were hampered as graphics hardware was too expensive, which prevented widespread use, or too slow, which did not make real time applications possible. Today, however, simulating realism at acceptable levels is becoming possible and the hardware and price are becoming less of an issue. We note that there are still many research challenges remaining, however, including:

- Further and better integration of human, machine and information. User interfaces must be improved, integration in clinical routine is difficult without a detailed analysis of the clinical workflow.
- Continued striving for improved algorithms, e.g. better realism, replacing tedious and unreliable segmentation procedures, etc.
- New application areas will emerge.
- Reducing cost will continue to be a priority.
- Validation of training simulations and computer graphics algorithms must be demonstrated. Currently, there are many claims as to what the technology is good for but relatively few papers providing evidence that this is actually true.
- Metrices used for performance assessment must fit with the curriculum in which the simulator is to be used.
- Integration of computationally intensive tasks with Grid resources and high performance computing may be needed.
- Moral and ethical issues will need to be addressed further.

One area that we have examined in detail is the use of VEs for surgical training. Current commercial solutions here focus on the field of laparoscopy surgery. The next few years will also undoubtedly see more variety of solutions, for such as interventional radiology, open surgery techniques, and other specialist domains such as dental surgery. Similar expansion of applications relying on computer graphics will occur in other medical areas too. While we have no doubt that as these items are progressed the impact will be significant, a sea change in medical opinion will still be required. A significant learning curve has to be overcome by today's medical profession if the advances already achieved, never mind those to come, are to be translated into realities in health care and benefit to patients. The positive collaborations between computer scientists, clinicians, medical physicists and psychologists that are present in the majority of the projects described in this survey paper augur well that this will happen to the future benefit of patients.

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