Volume Imaging of Soft Tissues with Ultrasound

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Introduction

Acquiring and displaying 3D (volume) sets of ultrasound (US) data raises a number of interesting questions of both principle and practice. Some of these relate quite widely to soft tissue imaging in general, others such as the problem of what to do about 'speckle', are more specific to US. The practical problems are such that US has lagged very far behind techniques such as CT and MRI in establishing the practical utility of 3D visualisation. This paper will address some general questions in 3D imaging, which are well illustrated by some fairly new examples of US volume reconstructions, and on solutions to some of the practical problems. Some clinical results with 3D US justify the effort.

Much work in 3D imaging has to date focussed on hard tissues (eg in CT), where a small number of generally well-defined surfaces are embedded in a soft tissue matrix. Visualising and surgically manipulating this hard tissue has provided a number of now established techniques [1],[2]. In the case where soft tissues are to be investigated, there are few cases where a surface can be extracted simply by thresholding as with bone CT data. Soft tissue structures may have relatively low contrast and there may be competition from various artefacts, (particularly in CT and US) so that an organ within the body of tissue is immediately obscured by a clutter of 3D background structures.

Terminology

It is important to consider the terminology used in imaging 3 or more dimensions, which is still not very consistent. Ultrasound throws up some problems which highlight this. I would like to use the following: an *image* (including 3D images) is the set of data produced by an imaging machine; the image information is visualised by means of a *display*. There are several reasons for this. Most importantly, in order to perceive the large amount of information in a 3D data set, a large number of separate 2D displays are in general needed. Particularly in soft tissue imaging, the image data may not necessarily be amenable to the familiar shaded surface presentation. In US one may see a structure best by moving a cut-plane through the stored data, producing a series of slices interactively as if one were scanning a real subject. I would like to include this in 3D imaging techniques, and call the data set a 3D image. This is quite in keeping with the use of the term *image* in mathematics and removes the need to coin new terms like *'scene'*, as each extra dimension of data acquisition is added.

Acquisition Techniques

Our acquisition and display system is based on a video frame-grabber (Quintek Harlequin) with a network of transputers hosted by a PC, allowing video images to be stored at the update rates of most scanners (25 FPS). This represents less complete information than would be acquired by digitising the raw RF or even the rectified signal. However the problem of dealing with this large amount of information would hamper efficient processing. This video acquisition also offers the benefit of great flexibility, interfacing immediately to most standard clinical scanners. The slice information is written into the image using positional information from a sensor attached to the transducer, so that each component of the signal can be assigned a correct position in 3D image space. We have used both a linear position sensor in recording intraluminal probe positions and a magnetic 3D sensor attached to real time probes. In the latter case, positioning to within about

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1/2 mm within a 30 cm cube of object space appears to be achievable.

Reflections from tissue structures of a size large compared with wavelength are potentially very anisotropic. In order fully to extract the available acoustic information it is very desirable to scan tissue from a range of different angles. The information associated with a spatial voxel then corresponds to a function over at least 2 angles. We are therefore potentially dealing with a 5D image or 6D if we include the effect of time variation due to heartbeat or tissue movement. Clearly to record the full information for a 6D image is impractical, requiring storage of the order of hundreds of megabytes. What must be done instead at the present stage is to scan intelligently, recording some simple parametric representation of the acoustic structure for each voxel, for example average echo level; maximum; asymmetry and so forth.

Special Problems

The problem of setting a single threshold is that unlike the case in CT where a uniform tissue region generates a uniform image, US generates non-zero signal only on local gradients of the acoustic properties. It is a differential technique and the essential character of an US image is discontinuous. In addition to the differential outlines of particular structures, US images contain *'speckle'*, resulting from a variety of effects including interference. Speckle appears as a fine texture on 2D images, where a knowledge of tissue suggests that there is structure on a scale below the resolution of the acoustic system. There is undoubtedly structure present, but the true structure does not correspond to that seen in the image. The effect of speckle in 3D will certainly appear different from that in 2D images. There have been several important attacks on the problem of what to do with speckle in US images. These span a wide range: from trying to process it away, to saying that it conveys vital information which must be perceived by the clinician, and including the range of attempts to classify and interpret particular tissue types by means of texture [3],[4]. It would appear that any attempt to use or remove texture will benefit significantly from having the full 3D data available.

The effects referred to here mean that in principle rendering US images by simple thresholding is not ideal. Surfaces found by thresholding are in general not globally continuous, do not uniquely define a single interface and can be considerably rough. Attempts to create displays of shaded surfaces must take account of this.

Display Generation

As a result of the spatially dense image data, in common with other soft-tissue imaging, a range of different displays is necessary in order to display the full information. We have so far used: cine loop of the acquired (parallel) slices; interactively positioned cutplane; depth shaded surface view; dual threshold depth shaded view; and surface illuminated view. Other possible methods such as translucency are difficult as a result of the high density of echo signals. In each of the surface methods, further interaction is possible in the form of setting of the threshold level for the surface. In visualising a solid body of information, placing a *surface cut* through the data allows a 2D display to be generated interactively as is done as with real-time US, giving and immediate 'feel' for the position of the cut plane. The display is generated by means of interpolation an update rates of about 2-5 per sec are possible with a 2 transputer front-end interactive system.

Our main image workstation produces surface shaded images. It employs ray tracing combined with local surface normal extraction to define a surface, illuminated by a combination of a single light source with diffuse background and depth enhancement. Applied to smooth hard tissue surfaces this produces very good images. However with US, thresholding often produces very rough irregular surfaces, which do not shade well. We have found that in many cases use of simpler methods such as straightforward depth shading are preferable. The resultant images show surface with less emphasised surface roughness.

In the case of the surface views, it is rare to see the required surface unoccluded by overlying tissues. It is therefore required to 'surgically' remove this overlying tissue from the image. This image surgery forms an integrated part of the facilities on the Medical graphics workstation developed at UCH/UCL. Currently this is a manual operation which is both time consuming and difficult,

since on first view of the surface illuminated image little may be seen of the required organ. It is planned to develop techniques for automatic segmentation to circumvent this problem.

Clinical Studies

We have studied the following: intravascular images of iliac and femoral arteries and veins; rectum; prostate and oesophagus using rotating endoscopic system; and in-vivo foetal images, using a variety of real time arrays.

In intravascular studies a single pass of the narrow probe along an artery can record all the information needed for assessment in a subsequent review. Since the region in question may well be seriously diseased and prone to dissection, the limitation of the interrogation manipulation is a great advantage. The technique can show details of the structures of hard or soft plaque on and within the wall, normal layered structure in the vessel wall and presence of dissection flaps. Intravascular US is relatively new and not yet established as a clinical technique. Use of 3D appears to greatly enhance its clinical potential.

In foetal imaging we have so far seen about 6 normal volunteers at between 12 and 20 weeks gestation. These have so far been acquired without the benefit of the recently acquired position sensor, relying on uniform manipulation of transducer by the operator. In spite of this the results have been very encouraging. We have so far been able to see surface details of the soft tissues of foetal head and face; to selectively visualise hard tissues to define clearly the sutures and fontanelles of the skull; to see the structures associated with the brain midline and ventricles and to verify normal anatomy of the limbs. The system facilitates foetal measurements in 3D although up to now the lack of true dimensions has limited this.

Development of these techniques could transform US into a scan and review type of procedure similar to that of CT and MRI, where specialist interpretations could be sought after the examination, using the 3D image data.

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