THREE-DIMENSIONAL MEDICAL ULTRASOUND

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ABSTRACT: A number of groups around the world are working in the field of three dimensional (3D) ultrasound (US) in order to obtain higher quality diagnostic information. 3D US, in general, involves collecting a sequence of conventional 2D US images along with information on the position and orientation of each image plane. A transformation matrix is calculated relating image space to world space. This allows image pixels and region of interest (ROI) points drawn on the image to be displayed in 3D. The 3D data can be used for the production of volume or surface rendered images, or for the direct calculation of ROI volumes.

1. INTRODUCTION

The purpose of this paper is to briefly introduce the reader to the field of three dimensional medical ultrasound. Ultrasound, as a medical imaging modality has developed since the end of the second world war, and grew out of developments in SONAR (SOund Navigation And Ranging). Ultrasound images from within the body are effectively 'sonar maps'. Ultrasound is used extensively in medicine, especially in obstetrics (care of the unborn baby), gynaecology and cardiology. Ultrasound examinations now account for at least 25% of all of medical image examinations.

Ultrasound images are tomographic in nature, i.e. they provide a cross-sectional images of patient anatomy. In this they are similar to X-ray computed tomography (CT) and magnetic resonance (MR) images. However, ultrasound does have significant advantages over CT and MR, for example, ultrasound is inherently safer than CT or MR, especially when imaging the fetus. Also, US machines are many times cheaper to purchase and maintain than CT or MR machines.

2. TWO DIMENSIONAL ULTRASOUND

Before we can discuss 3D ultrasound, we need to briefly describe the formation of 2D images. Images are formed by measuring the time for echoes to return from inside the body and therefore are sometimes called 'echograms'. Sound is reflected back towards the transducer wherever a change in the acoustic impedance is encountered. Conventional 2D images are produced by the emission of ultrasound pulses from an array of piezoelectric elements. The time for echoes from inside the body to return to the transducer elements is recorded and the depth of echogenic structures calculated assuming a velocity of sound of 1540 m/s in tissue.

Medical ultrasound probes typically operate in the range 3 - 7 MHz. The maximum resolution of course depends on the wavelength of the ultrasound. The wavelength of 3 MHz ultraound is about 0.5 mm in tissue. Axial resolution is given as half the spatial pulse length. For example, if a 3 MHz probe emits pulses four cycles in length, the axial resolution will be $(4 \times 0.5 \text{ mm})/2 = 1.0$ mm. Attenuation increases with frequency, therefore there is a trade-off between resolution and depth of penetration.

3. THREE DIMENSIONAL ULTRASOUND

3D ultrasound is a logical extension of 2D ultrasound, and a number of groups around the world are currently working in the field of 3D ultrasound (for a comprehensive review see Nelson and Pretorius 1999). 3D US involves acquiring a sequence of conventional 2D ultrasound images through a volume of interest (VOI) within the body. Some means is required to register the acquired images to a three dimensional coordinate system. This can be achieved by means of a 3D tracker connected to the ultrasound probe, or by a rotational or translational mechanism within the probe housing that sweeps the image plane through the VOI (Hamper et al, 1994, Gilja et al, 1995, Blaas et al, 1995). The two methods are depicted schematically in figure 1.

Transducers with an internal mechanism tend to be quite bulky and have to be held stationary for a number of seconds to allow a sufficient number of images to be acquired. Probes with an external localiser attached can be scanned free hand. An advantage of 'dedicated' 3D probes is their compactness—no external localiser is required. However, a disadvantage of dedicated 3D probes is that they have a narrow field of view close to the surface of the probe and also cannot capture extended structures (blood vessels in the leg, for example). In contrast, transducers with an external localiser can be used to capture extended structures.

3D image sequences can be acquired from within blood vessels using an intravascular ultrasound probes (Ennis et al, 1993, Rosenfield et al, 1992). This is effectively a very small array of piezoelectric elements on the end of a narrow tube (catheter) that can be inserted into an artery. The catheter is pulled back at a constant rate and images are acquired at regular intervals in time and therefore position. In this case it is assumed that tip of the catheter has moved in a straight line down the centre of the lumen. Rotation of intravascular probes has also been tried, in which case rotation angle is used to determine image orientation (Kok-Hwee et al, 1994).

Martin et al (1993) have developed a 2D phased array probe small enough to fit down an oesophageal catheter. The probe is deployed adjacent to the heart, providing clear images through the chambers. A pulley system sweeps the image planes through the heart, enabling cardiac volumes to be obtained during anaesthesia.

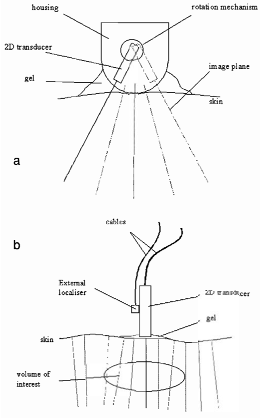

Figure 1. Different techniques for acquiring a 3D image dataset. (a) In a dedicated 3D probe some kind of mechanism rotates or translates a conventional 2D probe. (To improve clarity, only a few image planes are shown). (b) Free hand acquisition of images.

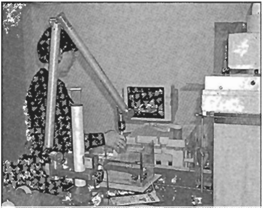

Figure 2. Mechanical Faro Arm attached to a 2D ultrasound transducer (a water-filled plastic test object is seen in the water tank).

Many different types of external space trackers have been used to record the position and orientation of ultrasound probes. The most commonly used are electromagnetic (EM) devices, for example those manufactured by Polhemus Inc, 3Space Fastrak (Gardener et al, 1991, Hodges et al, 1994, Hughes et al, 1996, Blass et al, 1999) and the Ascension Technologies Flock of Birds (Leotta et al, 1997, Gilja et al, 1998, Berg et al, 1999). These have a transmitter and receiver each containing three orthogonal wire coils. The transmitter coils are energised giving rise to signals, which are detected by the receiver coils. The relative strengths of the field picked up by the receiver enables the position and orientation to be calculated relative the coordinate system of the transmitter. The transmitter and receiver connect to a systems electronic unit, which usually interfaces to a PC via a RS 232 serial connection (although Ascension Technologies produce a system (pcBIRD) that plugs into a PCI bus in a PC).

EM trackers are prone to interference by nearby metal, in the case of the Polhemus Fastrak, studies have shown that this is minimal if the receiver is at least 7 cm away from the probe and 20 cm above a metal framed patient couch (Gardener et al, 1993). The flock of birds sensor can be attached directly to an ultrasound probe with no ill effects.

Mechanical arms have also been used (Sawada et al, 1983). Figure 2 shows a Faro Arm part of a system being developed by the author to quantify organ movement for radiotherapy purposes. Although mechanical arms are very cumbersome compared to EM trackers, they do have the advantage that they can be used in the presence of large quantities of metal, near a radiotherapy linear accelerator, for example.

Some groups have experimented with acoustic trackers comprising spark gaps placed on a structure attached to the ultrasound transducer and microphones placed on a stationary bar (Levine et al, 1989). These systems are fairly cumbersome and require the continual monitoring of temperature and humidity, which affect the velocity of sound in air.

There is a third method used by some companies in which there is no direct "localiser" but the operator does a freehand sweep of the scanning probe over the area of interest. The machine assumes a uniform movement and uses some image processing techniques to "stitch" the 2D scans together, in sequence, into a 3D data block

4. CONVERTING POINTS FROM IMAGE TO REAL WORLD COORDINATES

Central to any 3D US system is an algorithm to convert the coordinate of a point in a 2D US image (either an ROI points drawn on the image by the user, or a pixel) into world coordinates. The tracking device will give the spatial coordinates and orientation of the moving sensor relative to the device coordinate system (for the Faro Arm shown in figure 2, the origin of the coordinate system is the metal ball to the left of the base).

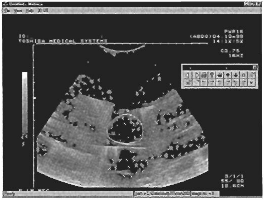

Figure 3. Ultrasound image through test object.

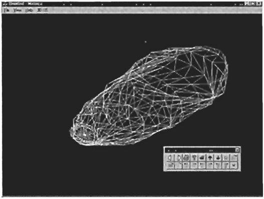

Figure 4. 3D reconstruction of test object as a triangle mesh.

The centre of the sensor is connected to a reference point in the image (for example the centre of the transducer face) via a series of 3D vectors. These vectors can be used to construct a (3×3) transformation matrix to convert points from image to world space (and vice versa if necessary). Calculation of the transformation matrix can be automated by scanning a test object of known dimensions.

5. ACQUISITION OF IMAGES

All ultrasound machines have a video output (for an auxiliary monitor for example), and so images can be acquired by a PC based video frame grabber. Dedicated 3D systems tend to use the video data available within the ultrasound machine. Ideally, the position and orientation data should be acquired at the same instant as each image. In reality there will be a small delay, which might need to be taken into account.

The number of images acquired and scan technique depends on what kind of processing is to be performed on the images. If volumes are to be calculated then only a few images, 10-20 for example, are required to adequately sample the structure or organ. Regions of interest (ROI) are traced on the relevant images (figure 3) and the points transformed from

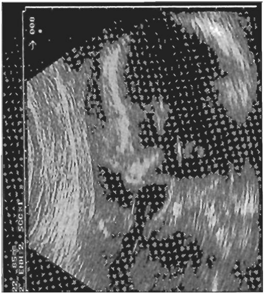

Figure 5. 2D ultrasound image through a fetal face.

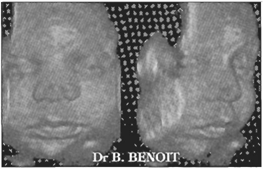

Figure 6. Surface rendered fetal face (not the same fetus as shown in figure 5). (Courtesy of Dr. B. Benoit).

image space to world space. The multiplanar ROIs can be used to calculate volume directly, or the ROI points can be connected into a triangle mesh (figure 4) prior to volume calculation. For volume measurements, it is best if the acquired image planes do not intersect as many volume calculation algorithms presume non-intersecting ROI planes.

If volume or surface rendering of the 3D image data is to be performed then many more images need to be acquired—in some cases as many as 10,000 (Barry et al, 1997). When very long sequences are acquired some means of removing the effect of cardiac and respiratory movements. Barry et al. have developed a system for quantifying plaque thickness in carotid arteries (the blood vessels that carry blood up the neck to the brain). Around 10,000 2D ultrasound images are stored on video tape. The audio channels are used to store the position and orientation data and ECG data (for cardiac gating). Images are digitised by computer after acquisition. Maybe in the future it will be possible to store this amount of video data directly onto a computer hard disc or RAM.

After the image data has been captured the next stage is to consolidate the data into a regular grid. A regular 3D grid is constructed, and a pixel intensity calculated for each node by interpolating between image pixels proximate to the node. Either a surface extraction algorithm can be applied to the data, or rays cast through the data to produce a volume rendered image. Surface rendered images can be used to calculate volume (assuming that a closed surface is generated). In spite of their name, volume rendered images cannot be used to calculate volume. Figure 5 shows an example of a fetal head in profile and figure 6 shows a surface rendering of a fetal head (although not the same one as in figure 5).

6. VOLUME ALGORITHMS

Prior to the advent of 3D ultrasound, organ volumes were calculated from maximum dimensions obtained from roughly orthogonal 2D images. These dimensions would then be used to obtain an ellipsoid volume. As can be imagined, significant errors arise if the organ is not nearly ellipsoidal in shape.

A number of different algorithms have been devised for calculating volume from multiplanar ultrasound images. Watanabe (1982) has developed a system, which utilises only multiplanar ROIs. The area of each ROI is multiplied by what we might term a local slice thickness. This technique is commonly used for calculating the volume of a structure imaged using X-ray computed tomography (CT) or magnetic resonance (MR) which produce parallel images of a precise and known spacing. In the case of images acquired using a hand held probe, images will not be exactly parallel and will not be exactly evenly spaced (as shown in figure 1b for example).

If the surface of an organ is tessellated into triangles, truly 3D volume algorithms can be used. For example a central point within the object can be connected to each triangle vertex to fill the space with tetrahedra (the volume of which are given by one sixth of the scalar triple product of the edge vectors connecting the centroid to the triangle vertices. This technique assumes that there is a clear line of sight between the centroid and each triangle vertex (i.e. does not have deep concavities). If the object is more complicated in shape, then tetrahedral decomposition can be performed within successive pairs of ROIs (Cooke et al, 1980).

An adaptation of Gauss' theorem (Hughes et al, 1997) can also be used. This involves multiplying the x, coordinate of each triangle centroid by the x component of the triangle normal and then by area of the triangle. This is repeated for the y and z centroid and normal components. The three volume calculations are then weighted according to the area of the surface projecting in the x, y and z directions respectively. Studies have shown that volumes can be measured to an accuracy of down to 2% (Hughes et al, 1996).

7. DOPPLER

When an ultrasound pulse reflects off a moving structure, a red blood cell (RBC), for example, it undergoes a shift in frequency. The magnitude of the frequency shift depends on velocity of the RBC relative to the ultrasound beam, and the sign of the shift depends on the direction of blood flow relative to the ultrasound beam. A colour image can be produced, overlying the grey scale anatomical image, showing variations in blood flow. If a sequence of registered 2D Doppler images is acquired then a 3D map of blood flow can be generated (Pretorius and Nelson, 1992, Picot et al, 1993).

8. THE FUTURE

A lot of work is currently being carried out to assess the advantages of 3D ultrasound compared to 2D (for example, Hamper et al, 1994). Some studies have already shown that 3D US is better able to detect fetal abnormalities such as cleft lip and palate (Ulm et al, 1999). Another advantage is that patients and non ultrasound literate medical staff are able to comprehend 3D images better than 2D US images.

Work is underway to develop 3D phased array probes. 2D phased array probes are commonly used in cardiac imaging. A linear array of piezoelectric elements is excited in a certain temporal pattern resulting in ultrasound beams propagating away from the face of the probe at various angles. An advantage of a phased array probe is that the field of view at depth is very much wider than the face of the probe. Hence it is possible to image the heart via the space between the ribs. If the linear array of piezoelectric elements is extended into a 2D array then we have a 3D phased array probe. However, 3D phased array probes have the same limitations as mechanical 3D probes, i.e. restricted field of view, especially close to the transducer. At present there are difficulties in developing viable 3D phased array probes. The major problem is one of bandwidth—i.e. all of the returning echoes cannot be processed quickly enough. Cross-talk between the closely spaced elements is another problem.

One can perhaps envisage a kind of mat placed across the abdomen of a patient that has a number of piezoelectric elements embedded within. The associated electronic circuitry would be able to process return signals in parallel thus producing a real-time 3D image of patient anatomy. Developments in 3D ultrasound rely on improvements computer technology.

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