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The accuracy of a new system for estimating organ volume using ultrasound

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Abstract
A new system is described for estimating volume from a series of multiplanar 2D ultrasound images. Ultrasound images are captured using a personal computer video digitizing card and an electromagnetic localization system is used to record the pose of the ultrasound images. The accuracy of the system was assessed by scanning four groups of ten cadaveric kidneys on four different ultrasound machines. Scan image planes were oriented either radially, in parallel or slanted at 30° to the vertical. The cross-sectional images of the kidneys were traced using a mouse and the outline points transformed to 3D space using the Fastrak position and orientation data. Points on adjacent region of interest outlines were connected to form a triangle mesh and the volume of the kidneys estimated using the ellipsoid, planimetry, tetrahedral and ray tracing methods. There was little difference between the results for the different scan techniques or volume estimation algorithms, although, perhaps as expected, the ellipsoid results were the least precise. For radial scanning and ray tracing, the mean and standard deviation of the percentage errors for the four different machines were as follows: Hitachi EUB-240, −3.0 ± 2.7%; Tosbee RM3, −0.1 ± 2.3%; Hitachi EUB-415, 0.2 ± 2.3%; Acuson, 2.7 ± 2.3%.

Keywords: 3D ultrasound, volume, CT, MR, fetal livers, fetal organ volume

1. Introduction

The measurement of volume is becoming increasingly important in medicine as abnormal volume is often an indication of abnormal function or pathology. For example, smaller than average fetal organ volume (expressed as a fraction of total fetal volume) may be associated with inter-uterine growth retardation (IUGR) and therefore prematurity. Larger than average prostate volume in men is indicative of benign hyperplasia of the prostate (BPH) or cancer. Increased volume of a transplanted kidney can indicate rejection.

Prior to the advent of tomographic imaging modalities the only way of estimating the volume of an internal organ was to obtain the major dimensions from plain x-rays and assume that the organ conformed approximately to a geometric shape such as an ellipsoid (Dodge *et al* 1960, Keats and Enge 1965, McLachlan *et al* 1968, Austin and Gooding 1971). An obvious disadvantage of geometric modelling is that organs are rarely regular in shape, and there may be significant differences in shape between normal and abnormal organs.

More accurate volume estimates can be obtained by acquiring a sequence of transaxial images through an organ. This may be achieved using x-ray computed tomography (CT), magnetic resonance imaging (MRI), positron emission tomography (PET) and ultrasound (US). The edge of an organ must be outlined on successive images and the points scaled using the in-plane calibration factors and the separation of adjacent image planes. For parallel images, the volume of an organ can be calculated, for instance, by multiplying the area of each region of interest (ROI) by the distance between image planes and taking the sum. However, ultrasound images are generally acquired by hand and so tend to be multiplanar, i.e. unevenly spaced, non-parallel and possibly sheared. In order to be able to calculate volume using ultrasound the position and orientation (pose) of each image in relation to adjacent images must be known. The transformed ROIs can then be used to estimate volume.

Various devices have been constructed for recording the location and orientation of an ultrasound transducer relative to a fixed reference point. Parallel scanning rigs have been used, for example, by Klein *et a*l (1993) for the fetus, Blankenhorn *et al* (1983) for blood vessels, Matsumoto *et al* (1981) for the heart and Hastak *et al* (1982) for the prostate. Mechanical arms are described by Sawada *et al* (1983) and Teicholz *et al* (1974).

More or less total freedom of movement is only possible with remote localization systems, or purpose-built volume scanners which have a small motor which translates or rotates the transducer within a housing (Gilja *et al* 1995). Remote acoustic systems are described, for example, by Brinkley *et al* (1978), King *et al* (1990), Moritz *et al* (1983) and Levine *et al* (1989), and remote electromagnetic localization systems have been described by Gardener *et al* (1991) and Hodges *et al* (1994). A range of techniques has been reported in the literature for calculating the volume of ROIs on multiplanar images. The ellipsoid method is frequently used on hearts (Dodge *et al* 1960) and prostates (Geirsson *et al* 1982). Planimetry is another commonly used method, which involves multiplying the area of each ROI by what might be termed the local slice thickness (Watanabe 1982). Another technique is tetrahedral decomposition in which the ROI points are connected to form a sequence of surface triangles joined to the centroids of the ROIs (Cook *et al* 1980).

In this paper we describe the use of an electromagnetic localization system connected to a PC-based video capture system. The details of operation have been previously described (Hughes *et al* 1996); the accuracy of the system was assessed on cadavaric fetal livers, kidneys and water-filled balloons on a single ultrasound machine. This study compares the overall accuracy of the system for measuring cadavaric kidney volume on four different ultrasound machines of varying quality. Three different scanning techniques were used to simulate how organs might be scanned *in vivo*, and four different volume calculation algorithms assessed.

2. Materials and methods

Ultrasound images were captured using a PC (486 DX2-66) video capture card (Win/TV, Hauppauge Computer Works, Hauppauge, NY, USA, distributed in the UK by ODT Europe, London), and the pose of the ultrasound probes recorded using an electromagnetic localization system (3Space Fastrak, Polhemus, Colchester, VT, USA, distributed in the UK by Virtual Presence, London). In a metal-free environment, the static accuracy of the system is specified as 0.8 mm RMS and 0.15° (2.618×10^{-3} rad) RMS, and the resolution 0.005 mm/cm range and 0.025° (2.181×10^{-3} rad). These values apply when the receiver is within 76 cm of the centre of the transmitter. Our own tests confirmed these results when the Polhemus was tested on a scanning couch.

Cadaveric porcine kidneys were scanned using a Toshiba Tosbee RM3/SSA-240A (3.75 MHz), a Hitachi EUB-415 (5 MHz), an Acuson 128XP/3 (5 MHz) and a Hitachi EUB-240 (3.5 MHz). All the probes used were of curvilinear design except for the Hitachi EUB-240, which had a linear array. Of the four machines used, the Acuson was the newest and the Hitachi EUB-240 the oldest.

The Fastrak receiver could be placed directly onto the transducer casings of the Hitachi EUB-415 and Tosbee RM3 scanners without any noticeable interference in the results. However, the receiver could not be placed directly onto the transducer case of the Acuson so a polyethylene offset rod was used (figures 1 and 2). For convenience, the displacement rod was also used on the Hitachi EUB-240. In each case, the vectors were oriented as in figure 1 and were aligned by eye and the lengths of ***v***1 and ***v***2 measured using a ruler.

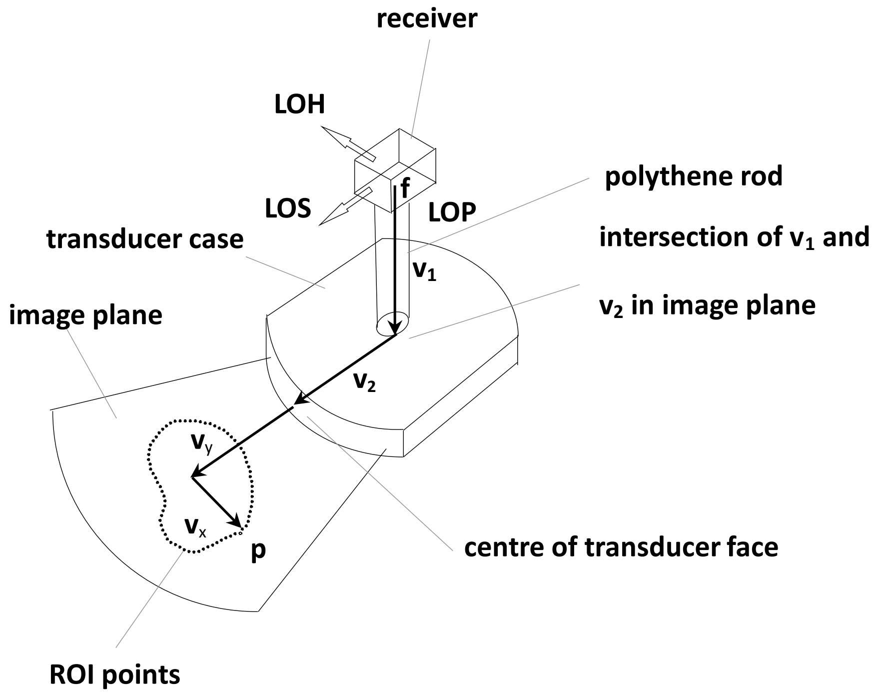


Figure 1. A schematic diagram of the attachment of the Polhemus receiver to the US probes. The receiver was attached directly to the transducer case for the Tosbee and Hitachi-415 and by means of a displacement rod for the Hitachi-240 and Acuson machines. The receiver has three vectors projecting from its centre: LOS, line of sight; LOH, line of hear and LOP, line of plumb. Each of the vectors $\boldsymbol{v}_1$, $\boldsymbol{v}_2$, $\boldsymbol{v}_x$ and $\boldsymbol{v}_y$ is aligned with one or other of the receiver vectors. The position of an image point is found by summing vectors $\boldsymbol{v}_1$, $\boldsymbol{v}_2$, $\boldsymbol{v}_x$ and $\boldsymbol{v}_y$.

The porcine kidneys (obtained from a local supermarket) were placed on a Perspex column in a Perspex tank (figure 2) filled with 26 l normal saline (0.9% NaCl solution). Normal saline was used in an attempt to reduce osmotic volume changes. The ultrasound transducer was placed in a gantry running across the top of the tank and moved by hand in order to acquire images oriented radially, parallel and slanted at 30° to the vertical (figure 3). The US transducer was stationary during image acquisition. About 5 minutes were required to scan each kidney using the three techniques.

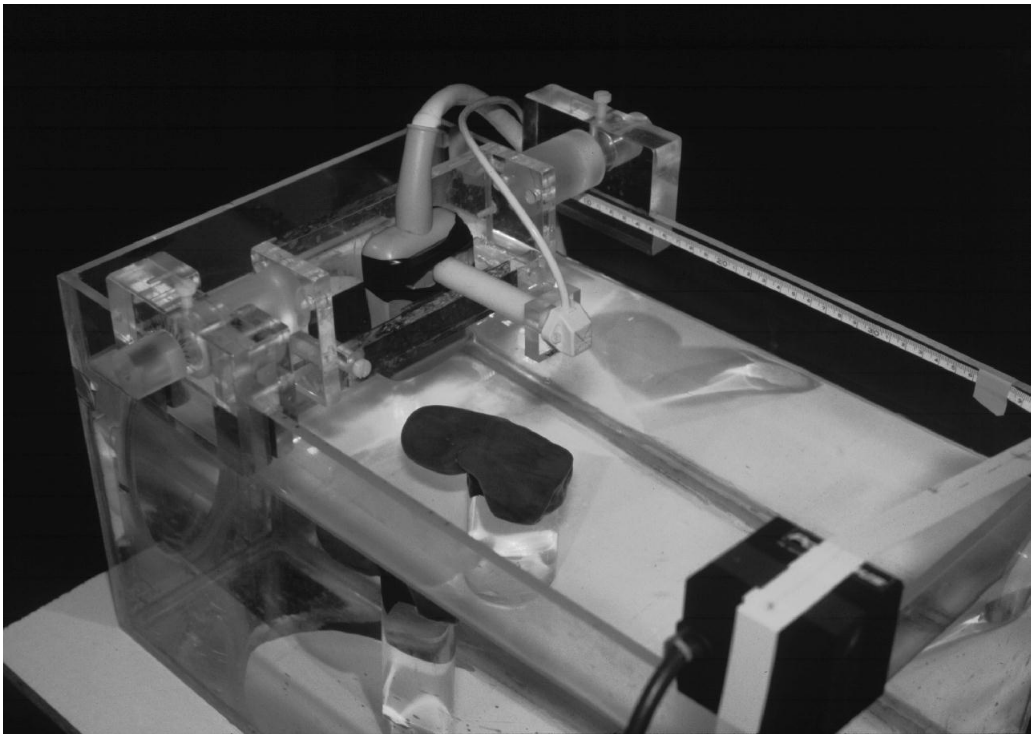


Figure 2. A kidney on a Perspex plinth within the scanning tank. The Polhemus receiver can be seen attached to the ultrasound probe by a plastic rod. The transmitter is at the bottom right of the picture.

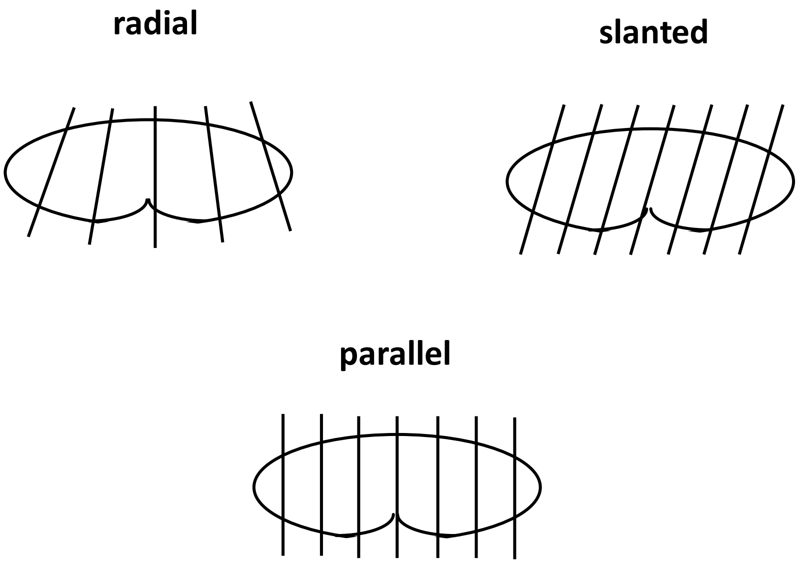
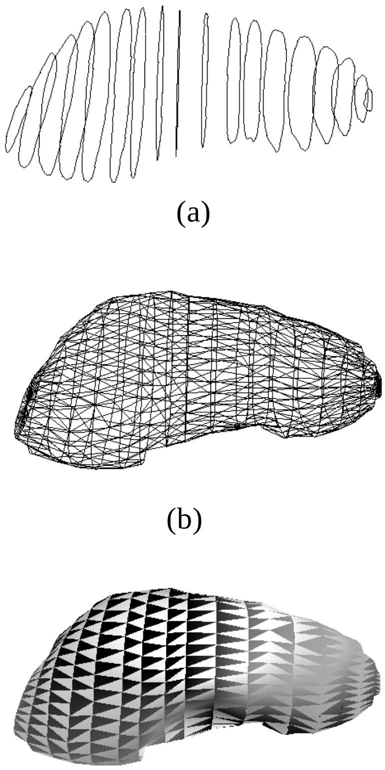


Figure 3. Three scanning techniques, radial, parallel and slanted, were chosen to simulate how an organ might be scanned *in vivo*.

Immediately after scanning, each kidney was placed in a 2000 ml graduated cylinder to measure the volume of water displaced. Five accurately machined Perspex cylinders ranging in volume from 30 and 260 ml were used to assess the accuracy of the water displacement method. The velocity of sound in the kidneys was assumed to be 1540 m s^{-1} within the kidneys, and errors in the path lengths of the *A*-scan lines caused by differences between the velocity of sound in water and tissue were ignored.

The Fastrak transmitter was placed on a corner of the tank. The receiver was always within 0.5 m of the transmitter. Between 15 and 20 (328 × 228 matrix, seven-bit resolution) images were acquired on each run. The image files were temporarily stored on the hard disk of the PC in the Microsoft Windows device independent bitmap (BMP) format.



(a)

(b)

(c)

Figure 4. (*a*) Transformed kidney ROIs, (*b*) triangle mesh and (*c*) surface rendering.

After each scanning session, the image and position files were transferred via a local network to a Titan graphics supercomputer (Kubota Pacific Inc, Santa Clara, CA, USA) for analysis. The edges of each kidney were outlined using a mouse, and the ROI points transformed into the Fastrak coordinates (figure 4) as described above. Each sequence of images took about 10 minutes to outline. After transformation, a triangulation algorithm (Hughes and Brueton 1994) was used to connect the outline points to produce a surface. Surface triangles were shaded light and dark to facilitate visual checking of the meshing algorithm. The surface model can be viewed from different directions.

Volume was calculated using the ellipsoid, planimetry, tetrahedral and a new ray tracing algorithm (RTA). To obtain the ellipsoid volume, the major axis was taken as the length of the kidney, and the product of the two minor axes as the maximum ROI area divided by π.

Kidney length was taken as the distance between the centroids of the first and last ROIs. This technique is really a simulation of the standard ellipsoid technique in which long- and short-axis images of an organ are acquired to obtain the above measurements. The planimetry technique involves multiplying the area of each ROI by the local slice thickness (Watanabe 1982). The tetrahedral volume is calculated by decomposing each pair of ROIs into a collection of tetrahedra (Cook *et al* 1980). The RTA method involves casting rays, arranged in a rectangular grid, through the surface triangular mesh.

The four sets of experiments were carried out in four separate locations: the Acuson and Hitachi EUB-240 were operated in an electrically noisy physics laboratory, the Tosbee in small room dedicated to obstetric scanning and the Hitachi EUB-415 in a spare room on a ward.

To assess the variability in scanning and tracing ROIs, the mean distance between ROI centroids and the mean distance between outline points were also calculated. Agreement between measured (water displacement) and calculated volumes were assessed by Bland–Altman plots (Bland and Altman 1986). These plots are produced by plotting the mean of each pair of measured and calculated volumes against the difference between the volume pair. The 95% limits of agreement are calculated as the mean of all the pair differences plus or minus twice the standard deviation of the differences.

3. Results

As there are so many permutations of the results only a few combinations are presented here to exemplify the overall results. The Tosbee was chosen to compare the four methods of calculating volume using radial scans, and the tetrahedral volumes were used to compare the three scan techniques. The radial tetrahedral volumes were used to compare machines.

Figure 5 shows the results for the Tosbee machine for the four methods of calculating volume with radial scans. In figure 5(*a*) the planimetry, tetrahedral and ray volumes all lie fairly close to the line of identity. The ellipsoid method has the greatest variability and appears to underestimate volume. This is borne out by figure 5(*b*) which shows the actual differences between the measured and calculated volumes. There appears to be little difference between the limits of agreement for the planimetry, tetrahedral and ray volumes.

As there is little difference between the planimetry, tetrahedral and ray volumes, the tetrahedral method can be taken as representative of the other two methods. Figure 6(*a*) shows the variation in tetrahedral volume with scan technique for the Tosbee. Figure 6(*b*) shows that the mean differences are fairly close together and the limits of agreement are similar in extent. All of the scan techniques result in slight overestimation of volume on average.

Figure 7 shows the radial tetrahedral results for all the machines. As there appears to be little difference between the scan techniques, the radial technique can be taken as representative of the other two and used to compare machines. Figure 7(*a*) shows that there is little to choose between the machines. The Bland–Altman plot in figure 7(*b*) shows that the limits of agreement are fairly similar although the limits of agreement for the Acuson and Tosbee are slightly narrower than for the Hitachi machines. The Acuson has the largest mean difference.

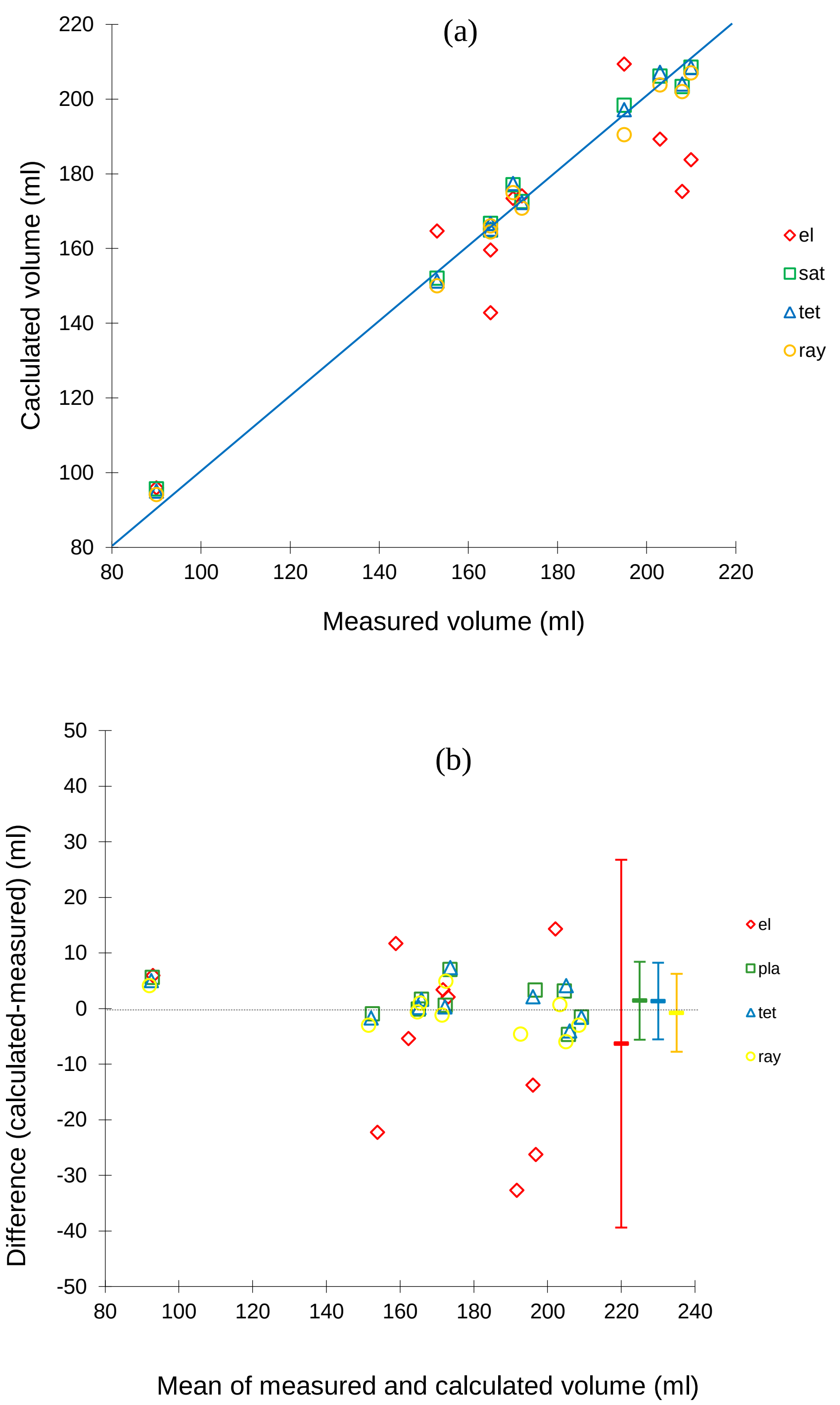


Figure 5. Radially scanned Tosbee kidney volumes calculated using the ellipsoid (el), planimetry (pla), tetrahedral (tet) and ray tracing (ray) methods. (*a*) Calculated versus measured volumes. The line of identity is shown. (*b*) The difference between calculated and measured volumes versus the average of the measured and calculated volume. The 'error bars' represent the 95% limits of agreement (±2 standard deviations) as prescribed by Bland and Altman (1986) and are designated by the first letter of the method.

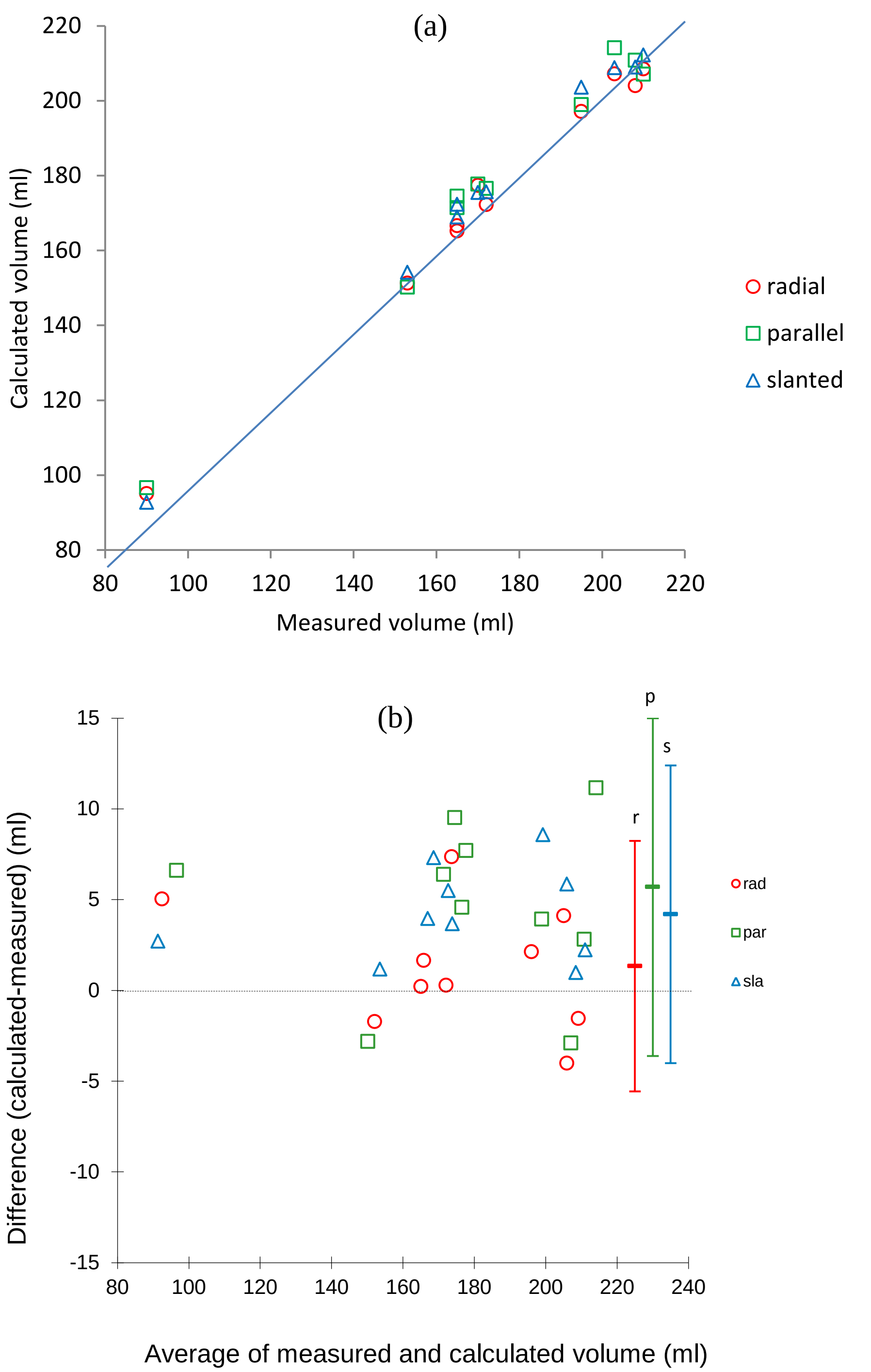


Figure 6. A comparison of the Tosbee kidney tetrahedral volumes for radial (rad), parallel (par) and slanted (sla) scans, (*a*) and (*b*) as above.

Table 1 shows the mean and standard deviation of the percentage error for the four volume calculation methods for radial scans for all machines. Note that the standard deviations of all the volume calculation methods are very similar within and between machines, the only exception being the standard deviation for the Hitachi EUB-415 radial ellipsoid volumes, which is comparable to the 3D methods.

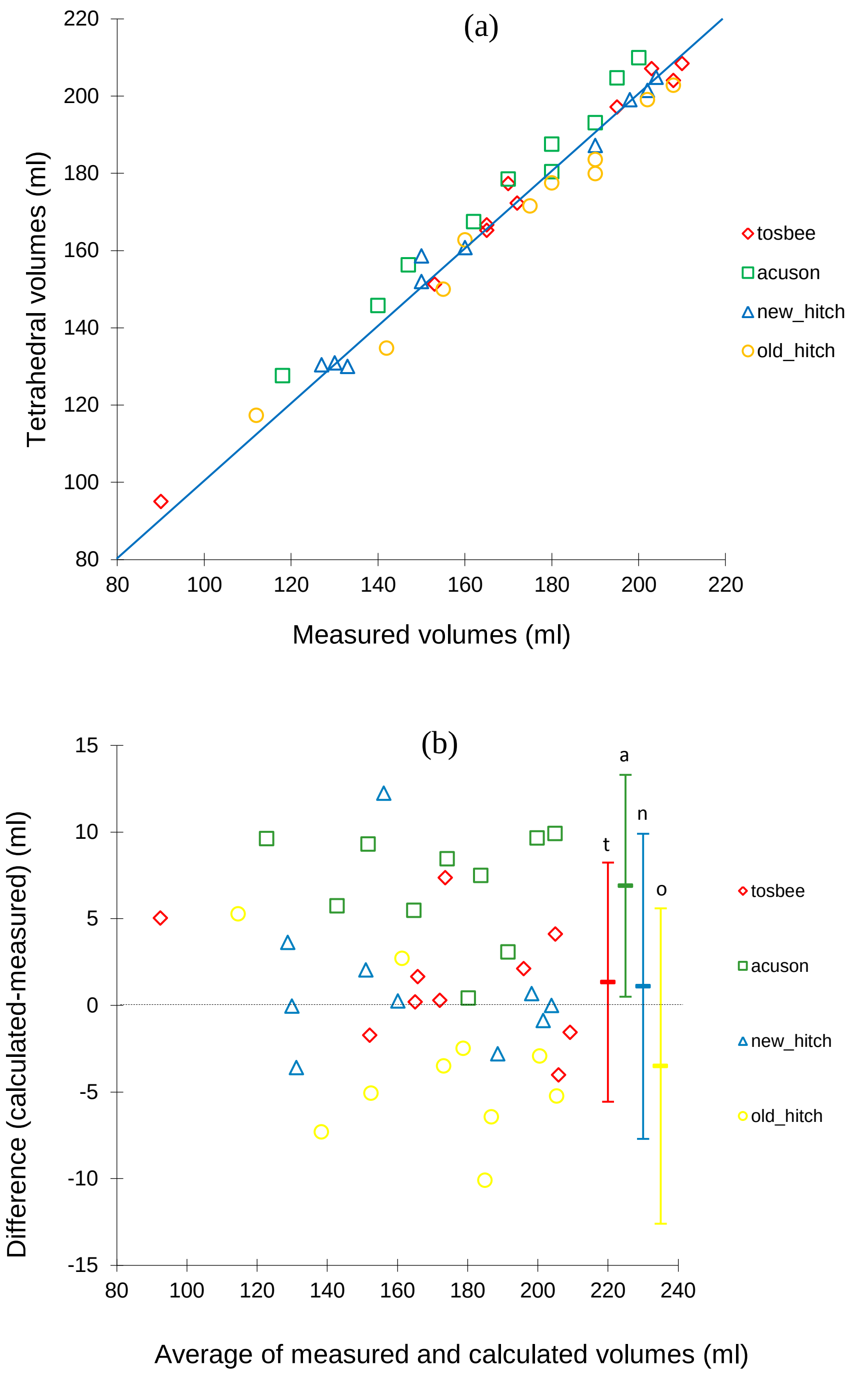


Figure 7. Radial scan tetrahedral volumes for the four ultrasound machines, (*a*) and (*b*) as above.

All kidneys appeared to be scanned consistently with little difference in the mean distance between ROI points and mean separation between image planes (table 2). The accuracy and precision (mean SD) of the water displacement technique was $-0.9 \pm 1.8\%$ (probably because the kidneys were, in this case, roughly ellipsoidal in shape).

Table 1. The mean and standard deviation of the percentage error for all four machines for the radial scans.

	el	pla	tet	ray
Acuson 128XP	5.3 ± 10.6	5.9 ± 2.1	4.3 ± 2.2	2.7 ± 2.3
Hitachi EUB-415	-8.9 ± 2.4	0.7 ± 1.9	0.8 ± 2.2	0.2 ± 2.3
Tosbee RM3	-2.7 ± 9.0	1.1 ± 2.5	1.0 ± 2.4	-0.1 ± 2.3
Hitachi EUB-240	-9.9 ± 6.9	-0.4 ± 3.5	-1.8 ± 3.0	3.0 ± 2.7

Table 2. The mean and standard deviation of the slice separation and the distance between outline points for the kidneys scanned on the Tosbee machine.

	Slice separation (mm)	Point separation (mm)
Radial	7.6 ± 2.3	5.3 ± 2.8
Parallel	7.0 ± 2.1	5.0 ± 2.6
Slanted	7.6 ± 2.5	5.1 ± 2.7

4. Discussion

The results show that the system produces acceptable results for a variety of ultrasound machines, scan techniques and volume estimation algorithms. The 3D volume algorithms (planimetry, tetrahedral and ray) are significantly better than the ellipsoid method (a standard deviation of ~3% compared to ~10%) except for the Hitachi EUB-415.

Several factors influence the overall error, for instance image and scanner resolution, tracing error, number of slices through the object, number of ROI points, variation in acoustic propagation velocity and refractive index, calibration, errors in measuring the position and orientation of the Polhemus receiver relative to the transducer etc. The accuracy of the water displacement technique was significantly better than any of the other accuracies. The small, but insignificant, differences seen between the variances of the scanning techniques and volume calculation methods may be due to a combination of the water displacement, rescan and retracing errors. More work needs to be done on assessing intra- and inter- observer scanning and tracing errors.

The mean percentage error reflects the overall systematic error for each machine. The large number of factors that influence the overall error probably explains the differences between the mean percentage errors. The Acuson had a greater systematic error than the other machines, which could be due to a number of factors (alignment of the receiver etc). The variance is the most important value as this indicates the degree of confidence that can be attached to any particular measurement.

All the machines performed more or less equally well in spite of marked differences in image quality. However, differences in image quality are likely to be important when scanning organs *in vivo*—on the assumption that the edge of an organ can be seen more easily and therefore outlined more easily on a higher-quality image. In the clinical situation there may be other machine independent errors such as breathing movement, movement of the alimentary tract and general patient movement. Another very important factor is the skill of the operator in perceiving the boundary of the organ. In practice, this could be the largest error.

A major disadvantage of electromagnetic localization devices is their susceptibility to interference from nearby metal, most immediately from metal within the ultrasound transducer case and cable. However, such effects can be minimized by using a plastic stand-off rod. For volume measurements, the error in the relative position of each image plane are more important than errors in absolute position. If environmental conditions are such that errors in absolute position exist, errors in relative position are likely to be small as in practice the traverse of the Polhemus receiver is generally only a few centimetres when an organ is scanned. Errors can also be reduced by keeping the transmitter and receiver as close together as possible and ensuring that there is a clear line of sight in between. Stray electromagnetic fields, from the ultrasound machine or other equipment are another source of error. More research needs to be done on quantifying these errors and assessing possible methods of reduction.

This experiment is limited in other ways, for instance, the transducers were held in a gantry and were stationary during image capture, rather than being hand held and moving. However, as the receiver pose information is acquired very fast (4 ms) and the transducer generally moves slowly (a few centimetres per second) image registration errors caused by the time delay between the acquisition of the pose data and image capture will be very small. Another limitation of this work is that only one observer was used to acquire the data and outline ROIs. However, this work gives an indication of the *minimum* error obtainable. *In vivo* errors are likely to be larger due to the differences outlined above. Clearly more work needs to be done to quantify these errors.

Our results are comparable to the results of others, for instance Gilja *et al* (1994) measured pigs' kidneys to accuracies of 6.5 ± 2.9% (3.25 MHz) and 9.8 ± 3.1% (5 MHz) and a fluid-filled pigs stomach to an accuracy of 0.4 ± 6.9% (3.25 MHz).

For measuring volume, the system described in this paper could compliment other imaging modalities. US compares favourably with CT and MR in terms of speed and cost, and has the added advantage of portability. US scanning is fast enough to be able to obtain a complete sequence of images through an organ within a single breath hold. However, a potential problem is that it takes 10 min or so to trace round the ROIs. Automatic surface extraction algorithms cannot presently be used reliably with ultrasound images as tissue interfaces tend not to be uniformly distinct around the circumference of a structure. However, image quality continues to improve and so automatic segmentation may be possible in the future. Image quality could possibly be improved by scanning a structure from multiple directions and merging the images into a single data set. This facility would be particulary useful in cases where structures are too large to be scanned in a single sweep and in cases where edges cannot be seen because of acoustic shadows, or an interface is tangential to the US beam.

The system may have potential for calculating the volume of fetal organs (brain, lungs, liver etc), prostates and kidneys. The system could perhaps be useful in diagnosing early rejection of transplanted kidneys and monitoring a decrease in volume associated with effective therapy (Absy *et al* 1987). Another potential use is in the follow-up of tumour therapy.

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