Performance of ultrasound based measurement of 3D displacement using a curvilinear probe for organ motion tracking

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Abstract: Three-dimensional (3D) soft tissue tracking is of interest for monitoring organ motion during therapy. Our goal is to assess the tracking performance of a curvilinear 3D ultrasound probe in terms of the accuracy and precision of measured displacements. The first aim was to examine the depth dependence of the tracking performance. This is of interest because the spatial resolution varies with distance from the elevational focus and because the curvilinear geometry of the transducer

causes the spatial sampling frequency to decrease with depth. Our second aim was to assess tracking performance as a function of the spatial sampling setting (low, medium or high sampling). These settings are incorporated onto 3D ultrasound machines to allow the user to control the trade-off between spatial sampling and temporal resolution. Volume images of a speckle-producing phantom were acquired before and after the probe had been moved by a known displacement (1, 2 or 8 mm). This allowed us to assess the optimum performance of the tracking algorithm, in the absence of motion. 3D speckle tracking was performed using 3D cross-correlation and sub-voxel displacements were estimated. The tracking performance was found to be best for y displacements and poorest for z displacements. In general, the performance decreased with depth, although the nature of the depth dependence was complex. Under certain conditions, the tracking performance was sufficient to be useful for monitoring organ motion. For example, at the highest sampling setting, for a 2 mm displacement, good accuracy and precision (an error and standard deviation of $<0.4\text{mm}$) were observed at all depths and for all directions of displacement. The trade-off between spatial sampling, temporal resolution and size of field of view is discussed.

1. Introduction

The goal of our research is the real-time three-dimensional (3D) tracking of abdominal organ motion, which is currently the subject of much investigation within the fields of radiation therapy and HIFU therapy (Shimizu *et al* 2000, Pernot *et al* 2004, Webb 2006). Ultrasound is ideally suited for imaging tissue motion because of its real-time functionality and its soft tissue imaging capability. It has been used in

radiation therapy to position patients prior to treatment and a number of different commercial 2D or 3D systems now exist (Chandra *et al* 2003, Cury *et al* 2006). Previously, we demonstrated the feasibility of imaging using B-mode ultrasound during radiotherapy (Hsu *et al* 2005). We showed that a 1D linear array transducer was able to measure the 2D displacement of a tissue phantom with sub-millimetre precision, regardless of electronic noise due to the operation of the radiotherapy equipment. The next steps are to develop the technique for 3D displacement estimation and to combine it with an absolute positioning system so that the measured motion of the internal anatomy can be placed in the treatment frame of reference. This paper deals with the first of these issues. It assesses the feasibility of 3D displacement estimation using a commercially available 3D abdominal ultrasound probe.

Previously, Morsy and von Ramm (1998) performed 3D correlation-based speckle tracking. To collect 3D volume-images, a piston transducer was moved in two dimensions across the surface of a stationary phantom. A small volume ($8\text{mm} \times 8\text{mm} \times 8\text{mm}$) was imaged by acquiring a radio-frequency (RF) echo line at positions on a grid of 42×42 points. The RF data were envelope detected prior to speckle tracking. Correlation tracking of the phantom was demonstrated for small ($\sim 1\text{mm}$) displacements in the axial and azimuthal directions by moving the phantom in between acquisitions. Chen *et al* (2005) demonstrated 3D correlation based speckle tracking using simulated RF data. They assessed the feasibility of using the technique to overcome decorrelation due to out-of-plane motion in 1D and 2D elasticity imaging. They showed that 3D correlation based tracking is more accurate than 2D tracking. They also introduced optimisation schemes to reduce the computational cost of 3D correlation based speckle tracking and discussed a hardware solution based on

FPGAs (field programmable gate arrays) that would facilitate real-time tracking. Pernot *et al* (2004) proposed a method using four sub-apertures, each comprising seven elements, which were selected from a hemi-spherical phased array containing 200 elements. Each sub-aperture created a 1D beam of ultrasound that travelled in a unique direction. RF echo lines were acquired using each sub-aperture and 1D axial displacement was measured along the four directions using 1D RF speckle tracking. 3D displacements were then resolved from the four directional measurements. The technique was shown to be effective in monitoring tissue motion at a point a few millimetres in front of a HIFU heating zone and was successful in guiding HIFU treatment delivery to moving samples of *in vitro* bovine tissue. To date, the authors have only attempted to measure the motion of a small region of tissue ($<1\text{cm}^3$), as HIFU is a highly localised treatment. They have concentrated only on translational motion, although potential approaches for dealing with rotation are discussed.

Unlike the previous techniques, which used custom-built scanning equipment, our approach is to use a commercially available real-time 3D ultrasound scanner. Primarily, we are interested in tracking abdominal organs such as the liver, bladder, pancreas and prostate. To image these organs, we require a low frequency transducer with a large imaging depth. Commercially available probes that meet these criteria are curvilinear to generate a wide field of view at large imaging depths. 3D abdominal probes consist of a curvilinear transducer that is mechanically swept in a fan shape. A large field of view is particularly useful in the context of guiding therapy, as it facilitates patient set-up and would allow tracking of large tumours as well as adjacent organs that may be at risk. However, there are a number of potential disadvantages to using a curvilinear array for our purpose (displacement estimation). Firstly, the angle

of incidence of the ultrasound beam changes with azimuthal and elevational position, which will cause the speckle pattern to decorrelate when there is movement in these directions. The effect of this decorrelation on tracking performance has not previously been studied, so the first aim of this paper was to determine whether displacement estimation with a 3D curvilinear probe is feasible given the angular decorrelation. A second potential disadvantage of curvilinear arrays is that due to the probe geometry, the spatial sampling in the azimuthal and elevational directions decreases with depth. In addition, as with all 1D ultrasound arrays, the elevational beam width decreases with distance from the elevational focus (which typically occurs at a depth of ~6cm). Therefore, assuming that displacement estimation with a curvilinear probe is shown to be feasible, the second aim of this work was to characterise the accuracy and precision of displacement tracking as a function of imaging depth.

The third aim of this paper was to assess the tracking performance as a function of the spatial sampling setting on the scanner (“low”, “medium” or “high”). This is different from the issue mentioned above of a depth-dependent sampling due to curvilinear geometry. Instead, we refer to the fact that manufacturers of 3D ultrasound scanners allow the user some degree of control over the trade-off between spatial sampling and acquisition speed via front panel settings. The temporal resolution for both mechanically swept 1D arrays and 2D array-based 3D ultrasound imaging systems is fundamentally limited by the speed of sound (Fenster *et al* 2001). The time taken to acquire a volume of data, t_{vol} , is proportional to the product of the imaging depth (ID), the number of lines (L), and the number of frames (F). To track organ motion due to respiration in real-time, assuming a breathing cycle of 3-5 seconds, we require a volume acquisition rate, $1/t_{vol}$, of ~10 Hz (Keall *et al* 2004). To achieve this volume

acquisition rate without restricting the field of view to a few cubic centimetres, spatial resolution has to be sacrificed using techniques such as parallel beam forming in receive mode (Smith *et al* 1991) or by reducing the spatial sampling frequency (so as to reduce L or F). As mentioned above, the latter can be achieved using front panel settings. Therefore, it would be worthwhile investigating the effect of reduced spatial sampling on the accuracy and precision of displacement estimation.

If the organ is in motion during data acquisition, this will cause the speckle pattern to blur and decorrelate, which will reduce tracking performance (Pernot *et al*, 2004). The extent of decorrelation due to motion will depend on the speed, direction and variability of that motion. In this initial study, volume-images of a phantom were acquired before and after the transducer had been moved by a known displacement, a method that we refer to as “step and shoot”. This allowed characterisation of the system under optimum conditions so that we could determine and understand the measurement errors in the absence of motion. A future paper will examine the additional degradation in tracking performance due to object motion during imaging.

2. Methods and materials:

2.1 Experimental set-up

A 3D curvilinear ultrasound probe with a broadband frequency of ~4-7 MHz (Model: 3D4-7EK, Medison Co., Ltd, Seoul, S. Korea) was used to image a phantom. The probe acquires a volumetric image by sweeping a 1D transducer array about an axis parallel to the transducer face whilst capturing 2D images at regular intervals so that they form a fan shape. The scanner can record up to 50 volumes of B-mode data

(Accuvix, Medison Co., Ltd, Seoul, S. Korea). These are stored in 8-bit binary format, the size of the matrix being defined in rows $\times$ columns $\times$ frames.

The probe was clamped to a mechanical articulated arm, which was mounted on a three-axis of motion translational stage (Nioutsikou *et al.* 2006). The position of the transducer was controlled by three stepper motors driven by commands from a PC. To assess tracking accuracy, we compared measured displacement vectors with known displacements. This required knowledge of any rotation between the coordinate system of the translational stage and that of the transducer. We attempted to attach the transducer to the translational stage such that visually, there was no rotation between the two systems. However, any residual misalignment was determined as described in Appendix A.

2.2 Tissue phantom experiments

A homogeneous phantom, approximately 20cm $\times$ 10cm $\times$ 8cm in size and made of a water-based gel (14% by weight porcine skin powdered gelatine) loaded with 2% by weight polyethylene scatterers, was used to simulate the ultrasound speckle pattern characteristic of homogeneous tissue, such as liver parenchyma. The speed of sound in the phantom was ~ 1580 m/s. The phantom was placed at the bottom of a large tank of water on top of ultrasound absorbent material (a polyurethane rubber, Aptflex F28, Precision Acoustics, Dorset UK) to reduce the intensity of ultrasound reflections from the bottom surface of the tank. The phantom and absorbent material were fixed securely in place and remained stationary throughout the experiments.

Rather than altering the position of the phantom, known displacements were achieved by moving the transducer using the translation stage. The transducer was moved over distances of 1, 2 and 8mm in each of three independent directions (x , y and z), as well as in all three directions simultaneously. A minimum displacement of 1mm was chosen as we do not require sub-millimetre accuracy for our application and a maximum displacement of 8mm was chosen to represent a typical maximum instantaneous organ displacement (Balter *et al* 1995; Nederveen *et al* 2002). At each position, and for each of the three sampling settings (low, medium and high), up to 20 consecutive volume-images were acquired. The height of the phantom was 10cm. To obtain measurements at depths greater than 10cm, two experiments were performed, the first using an imaging depth of 13cm and the second using an imaging depth of 20cm. In the first experiment, the front face of the probe was initially positioned at a height of approximately 3cm. This was changed to 10cm for the second experiment.

The imaging (or probe) geometry is shown in figure 1. For a point P (x , y , z) in the Cartesian co-ordinate system, x , y and z are defined by

$$\left. \begin{aligned} x &= (b+r)\sin\theta \\ y &= -D\cos\varphi \\ z &= D\sin\varphi \\ \text{where} \\ D &= (b+r)\cos\theta - a \end{aligned} \right\} \quad (1)$$

where r is the axial distance of point P from the face of the transducer, θ is the azimuthal angle, which is the angle between point P and the central line of the frame, φ is the sweep angle, which is the angle between the imaging plane containing P and the central axis, a is the distance between the physical pivot point of the ultrasound

transducer and the virtual apex of the sector described by the ultrasound field and b is the radius of curvature of the curvilinear transducer (see figure 1). The azimuthal angle and sweep angle were set to 69° and 65° respectively, for all experiments.

Assuming a perfectly aligned system, from equation set 1, an x motion within the Cartesian geometrical system produces an axial (r) and azimuthal (θ) motion in the probe's geometrical system, a z motion produces an axial (r) and elevational (φ) motion, and a y motion produces a combination of all three motions.

A volume-image is a 3D matrix in which each element represents one image voxel. For fixed sweep and azimuthal angles, the size of the matrix is determined by the imaging depth (no. of rows (R)) and the sampling setting (no. of columns or lines (L) and no. of frames (F)). The syntax $[R, L, F]$ will be used to denote the size of an image matrix or sub-regions thereof. Table 1 gives the number of rows and the axial sampling frequency (defined as the number of samples per millimetre) for the two imaging depths used in this study. Table 2 gives the number of scan lines and number of frames for each sampling setting. It also gives the spatial sampling frequencies in the azimuthal and elevational directions for two different axial depths, 7cm and 15cm.

2.3 3D displacement estimation algorithm

The 2D displacement estimation program written in MATLAB (The Mathworks Inc., Natick MA, USA) and described in Hsu *et al* (2005) was rewritten to track in 3D. The program estimates a 3D displacement vector $\hat{d} (d_x, d_y, d_z)$ at a point $P_I(r, \theta, \varphi)$ within a volume-image of the tissue phantom by performing a 3D cross-correlation search on the local speckle pattern. The program selects a *reference volume*, A , centred on point

P_1 , from a volume-image acquired at an initial start position. The r, θ, φ co-ordinates of P_1 were pre-programmed according to the position of the ultrasound field that was under investigation. A *search volume*, B , of equal size is selected from a user defined *search region* positioned in a second volume-image acquired after the transducer had been moved. The 3D normalised correlation coefficient, ρ , between A and B was calculated according to (Morsy and von Ramm 1998)

$$\rho = \frac{\sum_{r=1}^R \sum_{\theta=1}^L \sum_{\varphi=1}^F (a_{r\theta\varphi} - \bar{a})(b_{r\theta\varphi} - \bar{b})}{\sum_{r=1}^R \sum_{\theta=1}^L \sum_{\varphi=1}^F (a_{r\theta\varphi} - \bar{a})^2 \sum_{r=1}^R \sum_{\theta=1}^L \sum_{\varphi=1}^F (b_{r\theta\varphi} - \bar{b})^2} \quad (2)$$

where a is a voxel value within reference volume A (containing R rows, L lines and F frames), b is a voxel value within search volume B (of equal size), and $\bar{a}$ and $\bar{b}$ are the mean voxel values of volumes A and B respectively. ρ was calculated for all possible positions of the search volume as it was successively shifted in increments of one voxel throughout the user defined search region. The size and position of the search region were chosen according to the magnitude and direction of the known displacement. The algorithm outputs a 3D matrix that describes the sampled cross-correlation function (*ccf*), which is the spatial distribution of ρ . The next step is to locate the sub-voxel position of the peak of the *ccf*, $P_2(r, \theta, \varphi)$, which indicates the most likely position that the reference volume has moved to. Peak localisation was performed by firstly finding the element within the correlation matrix with maximum ρ (ρ_{max}), giving the integer location of P_2 . Sub-voxel location was then found using a 3-point 1D Gaussian fit to the correlation function, in all three dimensions. This was implemented using programs based upon code written for 3D particle tracking in

particle imaging velocimetry called URAPIV (Gurka *et al* 1999). Finally, the Cartesian coordinates $P_1(x_{P1}, y_{P1}, z_{P1})$ and $P_2(x_{P2}, y_{P2}, z_{P2})$ were determined using equation 1, and the displacement vector $\hat{d}$ found using

$$\hat{d} = R^{-1} * \left(\begin{bmatrix} x_{P1} \\ y_{P1} \\ z_{P1} \end{bmatrix} - \begin{bmatrix} x_{P2} \\ y_{P2} \\ z_{P2} \end{bmatrix} \right) \quad (3)$$

where R is the rotation matrix between the coordinate system of the transducer and that of the translational stage (see Appendix A).

Displacement estimation was performed for at least 10 of the 20 pairs of volumes, giving rise to at least 10 displacement measurements. The first few volumes of each data acquisition (acquired in the first few seconds) were unusable, as artefacts were observed in the volume-images due to initial juddering of the probe when it began its sweeping motion. Switching the probe on a few seconds prior to motion tracking will avoid this problem for future application to organ tracking during therapy. The first usable volume in the start position was cross-correlated with the first usable volume in the finish position, the second usable volume in the start position was correlated with the second usable volume in the finish position, and so on.

2.4 Parameters investigated

As explained above, the main objectives of this study were to assess the feasibility of the technique and to investigate the tracking performance as a function of axial depth and scanner sampling setting. Other parameters that might be expected to influence

the accuracy and precision of displacement estimation include the size of the reference volume used in tracking, the distance moved and direction of movement, and the azimuthal and elevational positions within the field of view. A complete list of the parameters investigated is shown in Table 3. We expect that the larger the reference volume, the greater the measurement precision. However, for increased reference volume size, the spatial resolution of the technique decreases (Hsu *et al* 2005, Smith and Fenster 2000). We therefore investigated precision and accuracy as a function of reference volume size in order to determine the minimum size that would achieve adequate tracking. The number of distances investigated was restricted to three in order to limit the amount of data acquired. The choice of minimum and maximum distances was discussed above. A third distance of 2mm was chosen because we are primarily concerned with tracking small distances. This is because when tracking organ motion during therapy, we wish to detect small movements out of the planning target volume, as this will improve treatment delivery accuracy.

3. Results

Accuracy and precision were investigated as a function of the parameters presented in Table 3. For one unique set of parameters, e.g. low sampling setting, 1mm movement, x -direction, reference volume size [9, 9, 9] and position P_1 , the measured displacement is the mean tracked displacement averaged over 10 to 20 measurements. The standard deviation across these data is due to noise associated with the transducer, i.e., thermal noise, electronic amplifier noise and any possible error in the elevational position of the 1D transducer array (i.e., if the mechanical sweep is not perfectly reproducible). The standard deviation was found to be less than 0.05mm for all measurements. Therefore, for results presented herein that were obtained using one

unique set of parameters, the standard deviation is not presented. The error on the known displacements, i.e., the positional uncertainty associated with the translation stage, is $\pm 0.004\text{mm}$ and therefore is also not considered further.

3.1 Reference volume size, distance and direction of displacement

The dependence of accuracy and precision on reference volume size (expressed as the number of voxels in one dimension, i.e., the number of rows, columns or frames) is shown in figures 2 to 4 for displacements in the x , y and z directions respectively. The mean displacement was obtained by averaging over a series of measured displacements with P_l positioned at different depths on the central axis (between 4cm and 13 cm in intervals of 0.8cm). The precision is defined as the standard deviation across this data set and therefore includes random variation due to different speckle patterns as well as any potential systematic variation with depth.

For x -displacements (figure 2), good accuracy ($>95\%$) and low standard deviation ($<3\%$) are obtained for all sampling settings, provided the size of the reference volume is at least $[7, 7, 7]$. In the y -direction (figure 3), very good accuracy and precision are obtained for all displacements, all sampling settings and all reference volume sizes with the exception of the smallest tracking volume ($[3, 3, 3]$). Accuracy and precision are considerably poorer for z -displacements (figure 4). With high and medium sampling settings, good accuracy and precision are obtained for a reference volume size of at least $[11, 11, 11]$. For the low sampling setting, accuracies are poor for 1mm and 2mm z -displacements; even for the largest reference volume size, the error is $\sim 0.5\text{mm}$, corresponding to an accuracy of 50% and 75% for 1mm and 2mm

distances respectively. From the above, it was decided that a reference volume of 5 rows, 7 columns and 11 rows ([5, 7, 11]) would be used to obtain subsequent results.

For application to soft tissue tracking during therapy, we define adequate accuracy to be either 80% or an error of 0.5mm, whichever is lower (i.e., for 1mm, 2mm and 8mm displacements, adequate accuracy means an error less than 0.2mm, 0.4mm and 0.5mm respectively). Adequate precision is defined to be a standard deviation $\leq 5\%$. Using these definitions and considering all data presented in figures 2 to 4, it can be shown that 67%, 75% and 59% of 1mm, 2mm and 8mm movements respectively were adequately tracked. This shows that tracking is poorest for 8mm displacements. In some cases, the mean displacements differ from the known displacements by amounts greater than can be expected from the precision of the measurements. This is due to a systematic bias in the displacement measurement, which is discussed in section 4.6.

Figure 5 gives the mean measured displacement and precision (error bars) for the x , y and z directions for a volume tracked over an x , y and z movement (xyz), where $d_x = d_y = d_z$. Comparing the accuracy and precision in figure 5 with the results for pure motion (d_x , d_y , and d_z) in figures 2, 3 and 4 respectively, it is apparent that tracking performance has deteriorated for all displacement vector components.

3.2 Correlation coefficient

The peak of the correlation coefficient function, ρ_{max} , is a measure of the degree of similarity between the reference volume before and after displacement. The greater the decorrelation, the lower the probability that the tracking algorithm can correctly identify the new position, meaning that ρ_{max} is an index of tracking precision (Morsy

and Von Ramm, 1999). If this relationship can be confirmed for our probe, it suggests the potential for using ρ_{max} as a measure of the reliability of the tracked movement.

Mean precision (as defined above), for a reference volume of size [5, 7, 11], was calculated over all displacements (x , y , z , and xyz , 1mm, 2mm and 8mm) and plotted as a function of ρ_{max} for the three different sampling settings (figure 6). Precision is plotted on a logarithmic scale. A separate logarithmic line of best fit was calculated for each sampling setting (figure 6). Pearson correlation coefficients between the data points and the best fit line were 0.63, 0.84 and 0.79 for high, medium and low sampling settings respectively, showing that there is a weak correlation between precision and ρ_{max} . In figure 6, the data points with standard deviation values of greater than 1mm (i.e., poor precision) are for 8mm z -displacements imaged with low and medium sampling settings. Data points with the lowest standard deviation are for y -displacements. The considerable difference in the tracking performance for these two directions is the most likely explanation for the logarithmic trend in the data.

3.3 Depth (axial position) dependence

Depth dependence of tracking performance was assessed by positioning the reference volume at different axial positions along the central A-line of the volume image (i.e., for P_I , θ and ϕ were kept constant at 0° and r was varied) for both 13cm and 20cm imaging depth data. Figures 7 to 9 give mean measured displacement, for 1mm, 2mm and 8mm displacements respectively, as a function of depth. For xyz motions, only the z -component of displacement is shown, denoted by $z(xyz)$, as this has the poorest accuracy and precision. In figure 9, for the low sampling setting, the z - and $z(xyz)$ -displacements are not included as they are not accurate (most exceed 15mm). Lines

joining the d_z data points have been drawn to aid the eye. Results for y -displacements are not given, as all were adequately tracked; the worst tracking performance was for a 2mm movement at a depth of 18cm, for low sampling setting, where the accuracy was 95% (an error of 0.1mm) and the standard deviation was 0.5% (or 0.01mm).

We observe a decrease in accuracy as depth increases. This is more evident in the high and medium sampling data, as the low sampling accuracy is poor at all depths. For x -displacements, the decrease in accuracy with depth is likely to be due to an increase in the azimuthal beam width and an increase in spatial sampling. The decrease in accuracy with depth is even more apparent for the z - and $z(xyz)$ -displacements. This corresponds with the greater depth-dependent decrease in the elevational spatial resolution (relative to the azimuthal spatial resolution). In general, high sampling settings produce more accurate results than medium sampling at all depths and for all displacements. This indicates that increasing the spatial sampling at depth has improved the tracking performance, implying that a depth-dependent increase in elevational and azimuthal beam width cannot be the only cause of the degradation of tracking performance with depth. It is worth noting that the azimuthal focal depth is set automatically to a depth that is equal to the nearest integer depth in centimetres to the middle of the field of view. Hence, for these data, where the imaging depths were 13cm and 20cm, the azimuthal focus occurred at 6 or 7 cm and 10 cm respectively. There was no obvious improvement in accuracy at these depths.

In summary, adequate tracking performance (denoted by the shaded areas in figures 7 to 9) was obtained for all x -displacements, using the high and medium sampling settings. Only the 8mm x -displacement was adequately tracked on the low sampling

setting. Z-displacements were adequately tracked, using the high sampling setting, to depths of approximately 130mm, 160mm and 110mm for 1mm, 2mm and 8mm movements respectively. For $z(xyz)$ displacements, adequate tracking was obtained up to depths of 100mm, 120mm and 110mm respectively.

Examination of the displacement tracking results reveals systematic depth-dependent trends in the x - and z -displacement data. An example of this behaviour is seen in figure 8, for the measurement of a 2mm x -displacement acquired using low sampling; the measured displacement shows an almost linear increase between 1.65mm and 2.5mm as depth increases. Investigation of the results has shown that is a direct result of bias in the sub-voxel displacement estimation.

3.4 Azimuthal and elevational position

Dependence of measured x -, z - and $z(xyz)$ -displacements on the azimuthal position (θ) and elevational position (ϕ) of P_I was investigated. Azimuthal and elevational sampling is symmetrical about the central axis and therefore we only investigated one half of the field. When varying the azimuthal position, the axial and elevational positions were fixed at 112.5mm and $\phi = 0^\circ$ respectively and when varying the elevational position, the axial and azimuthal positions were fixed at 112.5mm and $\theta = 0^\circ$ respectively. No dependence of tracking performance on the azimuthal or elevational position was observed.

4. Discussion

The goal of this study was to investigate the feasibility of the measurement of 3D displacements, in the absence of motion, using a commercially available curvilinear

3D ultrasound probe. Firstly, we investigated the effect of the axial variation of spatial sampling and beam width on the accuracy and precision of tracked displacements and secondly, we assessed tracking performance as a function of the spatial sampling setting on the scanner. In this section, we discuss the results of these investigations and expand the discussion to show how the spatial sampling will affect the temporal resolution and hence, the maximum speed of organ motion that can be tracked.

4.1 Direction of displacement

Of the x , y and z directions, tracking performance was poorest in the z direction. For z movements, elevational motion is the major displacement component. Performance was expected to be worst in this direction because for 1D transducer arrays, the elevational spatial resolution is poorer than the azimuthal and axial spatial resolutions. The elevational direction also has the poorest spatial sampling frequency. However, if the ratio of sampling interval to beam width were the same in the elevational direction as in the azimuthal and axial directions, we would not necessarily expect poorer sampling to have any additional adverse effect on tracking performance.

Error in the elevational position of the tracked volume translates to an error in both d_y and d_z . Using equation set 1, it can be shown that for a depth of 10cm and $\theta = 0^\circ$, an error in the elevational position of 1° will lead to an error of 2.2mm in the z -direction and an error of 0.2mm in the y -direction. This is particular to the curvilinear geometry of the probe and shows the importance of good tracking performance in all directions.

Tracking accuracy and precision were poorer for xyz movement than for z movement. This is due to the additional decorrelation of the speckle pattern as the probe moves in all three dimensions.

4.2 Distance moved

Overall, the accuracy and precision was worst for 8mm displacements. This is due to decorrelation of the speckle pattern. As the tracked volume moves away from the central axis, the angle at which the ultrasound beam is incident on the tracked volume changes and thus the speckle pattern also changes; for greater displacements, the angular change is larger. This effect is more pronounced at shallower depths because for a fixed Cartesian displacement, the azimuthal and/or elevational angular change is greater. Thus, when comparing the accuracy of 2mm and 8mm displacements in figures 8 and 9, the poorer accuracy for 8mm is more evident at shallower depths.

Angular decorrelation has implications for the maximum speed of organ motion that can be tracked, i.e., it is more probable that tissue can be tracked if it is displaced by 2mm between consecutive volume acquisitions than if it is displaced by 8mm. Therefore, it may be necessary to ensure that the temporal resolution is sufficient to limit the movement of the tissue between acquisitions to just a few mm. The temporal resolution of the probe is dependent on the spatial sampling and the size of the image volume; these relationships are discussed further in section 4.7. It is worth noting that even if angular decorrelation were not present, it would still be necessary to use a temporal resolution that restricts the inter-volume movement. This is because the purpose of motion tracking during therapy is to detect small movements of the organ before it moves out of the PTV (planning target volume). As we discussed previously

(Hsu *et al* 2005), the principal motivation for developing methods of monitoring the inter- and intra-fraction organ position is to reduce the geometrical margin below the level currently used in radiotherapy of abdominal organs (1-2 cm). A reasonable aim is to reduce the margin (and hence the inter-volume movement) to a few mm.

4.3 Sampling setting

Tracking performance was better for the high sampling setting than it was for medium, and medium was better than low. There was poor accuracy for z -displacements using medium and low spatial sampling settings and therefore, at present, these settings are not useful for 3D tracking (see figures 7 to 9). However, a significant contributing factor to the poor accuracy at these lower spatial sampling settings is bias error in the sub-voxel displacement estimation (see section 4.6). Therefore, if this problem can be solved, tracking performance may improve for these settings. It should be noted that we also found that accuracy is adequate to track x -displacements using both high and medium sampling settings. Therefore, an optimum setting would combine the medium azimuthal sampling frequency with the high elevational sampling frequency, which would increase temporal resolution (relative to the high spatial sampling setting) whilst maintaining adequate tracking performance.

4.4 Position

The tracking accuracy was observed to decrease with depth, as was expected due to the increase in azimuthal and elevational beam width with depth beyond the azimuthal and elevational focal depths. The azimuthal focus occurred at depths of ~7cm and 10cm for imaging depths of 13cm and 20cm respectively whilst the elevational focus occurred at 6cm. There was also a decrease in spatial sampling frequency with depth

and it is difficult to determine which of these two factors was more important in the decline of tracking performance. However, at greater depths an improvement in tracking accuracy was seen when changing from medium and low sampling settings to high, implying that the two lower spatial sampling settings reduce tracking performance.

The rate of decrease in tracking accuracy with depth was displacement-dependent. It was worst for 8mm (i.e., a more rapid decrease with depth was observed). As mentioned above, depending on the speed of organ motion, to maintain tracking performance it may be necessary to limit the distance moved between image volume acquisitions and therefore this will become increasingly necessary at greater depths.

The reference volume size used to investigate tracking performance as a function of depth ([5, 7, 11]) was determined using data obtained at depths of 4cm to 12cm. It is possible that, at greater depths, increasing the size of the reference matrix would increase tracking performance (Smith and Fenster 2000). For a linear transducer array with axially invariant sampling, if the size of the reference volume is fixed, then, as beam width increases as a function of depth, the amount of information contained in the reference volume will decrease and tracking performance will also decrease. However, for a curvilinear array, the spatial sampling frequency also decreases axially and therefore provided the rate of change of spatial sampling frequency is similar to the rate of increase in beam width, there will not be a significant reduction in the amount of information in the reference volume as a function of depth (i.e., in Cartesian space, the reference volume size increases with depth). Consequently,

increasing the reference volume size at greater depths may not improve tracking performance.

The tracking performance did not vary as a function of azimuthal or elevational position. This was as expected because the spatial sampling and spatial resolution only vary as a function of axial position. If another mechanism for causing elevational and azimuthal dependence was present, it was too small to be observed. At the very azimuthal edges of the field, i.e., $\theta \sim 34^\circ$, we might see some change in accuracy and precision due to defocusing of the beam at these points (Goldstein 2000). However, due to the size of the search volume in the azimuthal direction (7 voxels), the midpoint of the reference volume did not approach such a high value of θ .

4.5 Correlation coefficient

The precision of the displacement estimates decreased with decreasing ρ_{max} (figure 6). It is therefore possible that in the implementation of real-time 3D tracking, ρ_{max} could be used as a measure of the reliability of the tracked positions. For example, if ρ_{max} falls below a predetermined threshold, the displacement estimates could be “rejected” until the tracking system locates the tracked volume with greater precision. This is particularly important for any automated method of controlling patient treatment.

In this study, where the applied motion was translational, the main sources of decorrelation were electronic noise and change in the incident angle of the ultrasound beam. It has been shown for a linear array transducer that decorrelation of the speckle pattern would also occur due to tissue rotation and deformation (Meunier and Bertrand 1995). Although these have not been investigated in the present study, it is

worth noting that rotation is in fact equivalent to a change in imaging angle. For example, an 8mm x -movement at an imaging depth of 4cm produces a 5.7° change in the angle of incidence of the ultrasound beam and this is adequately tracked (see figure 9). Typically, it is not expected that abdominal organs will rotate to a greater extent than $\sim 10^\circ$ (Aubry *et al* 2004) and therefore the amount of rotation could perhaps be restricted in a similar fashion to the amount of movement, i.e., by using a high temporal resolution. Note that this would also restrict the inter-volume speckle decorrelation due to tissue deformation.

The feasibility of tracking in the presence of rotation and deformation will be the subject of future work. The previous paragraph addressed the issue of how to deal with decorrelation caused by rotation and deformation. However, it may also be of interest to measure these types of motion, as this would provide more detailed information on the location of the target organ than a simple measure of global translation (Hsu *et al*, 2005). To measure non-translational motion, the reference volume would need to be successively moved to different parts of the ultrasound field to obtain the spatial distribution of displacement. However, this would increase the computation time and may make the procedure difficult to implement in real time.

4.6 Error due to bias in the sub-voxel interpolation

In some instances, a strong depth-dependent trend across all depths was observed in the measured displacement data (figures 7 to 9). This can be attributed to bias in the sub-voxel location of the peak of the cross-correlation function (*ccf*). The magnitude of the bias error was greater for low and medium spatial sampling settings, which have larger sampling intervals. The bias error was also greater at greater depths where

the sampling intervals are larger. The bias error is an absolute error and therefore it is more deleterious to the tracking accuracy of smaller displacements. For example, for the low sampling setting, a bias error of 0.4mm was observed at a depth of 15cm. If we are tracking a 1mm displacement, this introduces an inaccuracy of 40%.

The Gaussian peak fit is based on the notion that the correlation peak has an approximately Gaussian shape; anything that does not approach this will lead to non-optimal sub-pixel interpolation. This behaviour has been investigated by Nobach *et al* (2005), who demonstrate with the aid of simulated imaging functions, such as an airy function, that a non-Gaussian imaging function yields larger (up to a factor of 10) bias errors when using the Gaussian fit. From the results it is evident that, if we wish to track with medium and low sampling settings, the bias error will have to be reduced. Furthermore, as discussed above, we may wish to restrict the inter-volume movement and measure large (several centimetre) movements by summing up consecutive displacement estimates. In such a case, there is a theoretical possibility that the bias error would accumulate to produce a significant error on the total distance moved. Future work will investigate possible methods to reduce the bias error (Scarano 2002).

4.7 Implications for temporal resolution

As mentioned in sections 4.2 and 4.4, we may wish to restrict inter-volume movement, especially at greater depths, by increasing temporal resolution. The time it takes to acquire one ultrasound volume is

$$t_{vol} = L \times F \times 2 \times \frac{ID}{c} \quad (4)$$

where L is the number of lines per frame, F is the number of frames, ID is the imaging depth (cm) and c is the speed of sound (cms^{-1}). For a given imaging depth, the temporal resolution can be improved by reducing L or F , i.e., by reducing the azimuthal sampling, elevational sampling, the azimuthal angle or the sweep angle. There are limitations on the extent to which these can be reduced. If sampling is too low, we will be unable to track with sufficient accuracy (section 4.3) and if the angles are too small, the field of view (FOV) will be reduced and the tissue will move out of the imaging volume during a treatment session. The maximum temporal resolution that can be achieved will determine the maximum organ speed that can be tracked. The following analysis uses the results of this study to examine maximum trackable organ speed as a function of depth.

We calculate the maximum trackable organ speed as a function of depth for two FOVs (5cm and 10cm), where FOV is defined as the size of the image matrix in both the x and z directions at the depth of the reference volume. These FOVs would be sufficient to track organ motion due to respiration, where the total distance moved is on the order of 3cm (Balter *et al* 2001). The imaging angle needed to provide a particular FOV will decrease with depth. For example, for a 10cm FOV, at a depth of 10cm, we require azimuthal and sweep angles of 39° and 44° respectively, whereas at a depth of 15cm, we only require azimuthal and sweep angles of 29° and 32° respectively. The imaging angles will be reduced by a factor of two for the 5cm FOV.

From the results discussed above, we identified that we require the high sampling setting to be able to accurately track 2mm displacements (in the absence of object motion) in all directions to depths of 15cm. The high spatial sampling setting uses

elevational sampling of 1° per frame and azimuthal sampling of 0.56° per line. We also showed that the medium sampling setting, which has azimuthal sampling of 0.66° per line, is adequate to track x -displacements. As discussed above, it may be desirable to use a combination of “high” elevational sampling and “medium” azimuthal sampling. Therefore, the maximum trackable organ speed has been calculated for two situations, the high spatial sampling setting and a theoretical combination of the high and medium settings (“high/medium”). Assuming a 10cm FOV and an imaging depth of 10cm, we will have 70 lines ($L = 70$) and 44 frames ($F = 44$) for the high spatial sampling setting and 59 lines ($L = 59$) and 44 frames ($F = 44$) for the high/medium spatial sampling setting. Using equation 4, this results in t_{vol} being 0.4s and 0.33s for the high and high/medium sampling settings respectively. The maximum trackable organ speed is then given by the product of $1/t_{vol}$ and the maximum distance that can be tracked, which is assumed to be 2mm (section 4.2).

Figure 10 gives the maximum trackable organ speed as a function of depth up to 15cm, which is the maximum depth at which 2mm z -displacements could be tracked. Previous studies have shown that as a result of respiratory motion, abdominal organs can move with speeds of up to 15mm s^{-1} (Pernot *et al* 2004). From figure 10, we can see that using the high spatial sampling setting, it is feasible to track organs at these speeds, and at depths of 5 to 15 cm, using a 5cm FOV but not a 10cm FOV. For a combination of high and medium sampling settings, which produces a 19% increase in maximum trackable organ speed, an organ speed of 15mm s^{-1} can be tracked at all depths up to 15cm using a 5cm FOV.

Considering that a small field of view may be required to increase temporal resolution, this suggests that the principal advantage of a curvilinear probe (its wide FOV) may not be relevant for this application. Therefore, a 3D linear probe might be more appropriate, especially as it would not be subject to the reduced tracking performance caused by angular decorrelation and poor spatial sampling at depth. However, commercially available 3D linear probes use high frequency transducers and would only be suitable for imaging superficial abdominal anatomy (up to ~8cm in depth).

Figure 10 is for one specific result. Further work may show this to be conservative, i.e., it may show that displacements greater than 2 mm (but less than 8mm) between volumes can be tracked. Furthermore, the spatial sampling requirements could be reduced if the bias error is reduced, further increasing temporal resolution. However, figure 10 is helpful in understanding the trade-offs between the maximum trackable organ speed, depth, and FOV and shows that as depth increases, the maximum trackable organ speed increases. This is due to the fact that for a fixed FOV in Cartesian space, the temporal resolution can be increased by reducing the azimuthal and elevational angles. This analysis has been based on results from measurements made using the “step and shoot” method of acquisition and therefore does not include any potential reduction in the maximum trackable organ speed due to motion during volume acquisition. This will be the subject of future work.

4.8 Real-time implementation

The displacement calculations reported in this study were computed off-line. The eventual goal is to perform 3D displacement estimation in real-time. As discussed

previously (Hsu *et al* 2005), real-time 2D elastography, which requires a 2D cross-correlation search at many different locations in the 2D ultrasound image, is commercially available. Our cross-correlation is performed in 3D but is for one location only, therefore the number of computations is comparable and it is expected that real-time implementation is feasible. Optimisation schemes to reduce the computational cost of 3D correlation based speckle tracking and a hardware solution based on FPGAs are discussed in Chen *et al* (2005).

Conclusions

This study has shown that a commercially available 3D curvilinear probe can be used to track 3D speckle patterns that undergo pure x , y , z and xyz motions. Accuracy and precision are adequate to track small displacements (1mm to 2mm) for application to motion tracking in therapy using the scanner's "high" sampling setting. Correlation coefficient was shown to decrease with precision and therefore may have the potential to be used as a measure of confidence in the tracking system. We have shown that larger displacements result in greater decorrelation of the speckle pattern due to angular change. Therefore, we require sufficient temporal resolution to limit the distance moved between volume acquisitions. Given the observation of good tracking performance for 2mm displacements with the high sampling setting, we have estimated maximum trackable organ speeds as a function of depth.

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Appendix A

The transformation between the coordinates of a point in the ultrasound frame of reference, M_U , and the coordinates of the same point in the frame of reference of the translational stage, M_S , is given by:

$$M_U = RM_S + T \quad (A1)$$

where R is a 3×3 rotation matrix and T is a 3×1 translation matrix. Equation A1 can be solved by finding the coordinates of a set of three non-coplanar points in both frames of reference. The use of more than three points increases the precision with which R and T can be determined. Using N points, such that the matrices M_U and M_S each contain $3 \times N$ elements, the rotation and translation matrices are obtained by minimising ϵ^2 in the following least-square fit equation (Bouchet *et al.* 2001):

$$\epsilon^2 = \sum_{n=1}^N \left\| M_{U,n} - (RM_{S,n} + T) \right\|^2 \quad (A2)$$

The mutual points were obtained by embedding a 0.4mm-diameter ball-bearing in a gelatine phantom and acquiring volume-images of the ball-bearing at 50 different transducer positions ($N = 50$).

Our aim was to compare measured and known displacements. The displacement between two points depends only on their relative (not absolute) positions, so the translation matrix, T , did not need to be determined. If all points are transformed such that they are referred to the centroids of the sets of points in ultrasound and stage space respectively (the transformed coordinates being denoted by M_{UC} and M_{SC}), R can be determined independently of T , by minimising (Bouchet et al 2001):

$$\mathcal{E}_R^2 = \sum_{n=1}^N \left\| M_{UC,n} - RM_{SC,n} \right\|^2 \quad (\text{A3})$$

Equation A3 was solved in MATLAB, using singular value decomposition, with the line of code $[U, W, V] = \text{svd}(M_{SC} * M'_{UC})$, where the superscript ' denotes the matrix transpose. The rotation matrix was then given by the line of code $R = V * U'$.

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Figure 10. Maximum trackable organ speed as a function of depth for 5cm field of view and a 10cm field of view and using the high and high/medium sampling settings.

Figure 1. Schematic diagram of the imaging geometry. x , y and z directions (translation stage motions) are shown relative to elevational, azimuthal and axial directions. Distances a and b are 14.5mm and 40mm respectively, where b is the radius of curvature of the transducer. The central axis is coincidental with the y -axis.

Figure 2. Mean measured displacement and mean precision (error bars) as a function of reference volume size (expressed as the number of voxels in one dimension) and spatial sampling setting, for 1, 2, and 8mm displacements in the x direction. Data points have been staggered along the horizontal ordinate to separate the error bars.

Figure 3. Mean measured displacement and mean precision (error bars) as a function of reference volume size (expressed as the number of voxels in one dimension) and spatial sampling setting, for 1, 2, and 8mm displacements in the y direction. Data points have been staggered along the horizontal ordinate to separate the error bars.

Figure 4. Mean measured displacement and mean precision (error bars) as a function of reference volume size (expressed as the number of voxels in one dimension) and spatial sampling setting, for 1, 2, and 8mm displacements in the z direction. Data points have been staggered along the horizontal ordinate to separate the error bars.

Figure 5. Mean measured displacement in x , y and z directions for xyz displacements averaged over displacements measured at 8mm intervals between depths of 4cm and 13cm at the central axis for the high, medium and low spatial sampling setting using reference volume size [5, 7, 11].

Figure 6. Correlation coefficient as a function of precision for 1, 2 and 8 mm displacements, x , y , z and $z(xyz)$ motions. Logarithmic trend lines have been fitted to high, medium and low data sets with correlation coefficients of 0.63, 0.84 and 0.79 respectively.

Figure 7. Measured displacement as a function of reference volume depth for 1mm x , z and $z(xyz)$ displacements for the high, medium and low spatial sampling settings using reference volume size [5, 7, 11].

Figure 8. Measured displacement as a function of reference volume depth for 2mm x , z and $z(xyz)$ displacements for the high, medium and low spatial sampling settings using reference volume size [5, 7, 11].

Figure 9. Measured displacement as a function of reference volume depth for 8mm x , z and $z(xyz)$ displacements for the high, medium and low spatial sampling settings using reference volume size [5, 7, 11].