A clinical system for three-dimensional extended-field-of-view ultrasound

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Objective: This work is concerned with the creation of three-dimensional (3D) extended-field-of-view ultrasound from a set of volumes acquired using a mechanically swept 3D probe. 3D volumes of ultrasound data can be registered by attaching a position sensor to the probe; this can be an inconvenience in a clinical setting. A position sensor can also cause some misalignment due to patient movement and respiratory motion. We propose a combination of three-degrees-of-freedom image registration and an unobtrusively integrated inertial sensor for measuring orientation. The aim of this research is to produce a reliable and portable ultrasound system that is able to register 3D volumes quickly, making it suitable for clinical use.

Method: As part of a feasibility study we recruited 28 pregnant females attending for routine obstetric scans to undergo 3D extended-field-of-view ultrasound. A total of 49 data sets were recorded. Each registered data set was assessed for correct alignment of each volume by two independent observers.

Results: In 77–83% of the data sets more than four consecutive volumes registered. The successful registration relies on good overlap between volumes and is adversely affected by advancing gestational age and foetal movement.

Conclusion: The development of reliable 3D extended-field-of-view ultrasound may help ultrasound practitioners to demonstrate the anatomical relation of pathology and provide a convenient way to store data.

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Three-dimensional (3D) ultrasound has become increasingly popular in recent years, providing an efficient framework for safe, low-cost volumetric imaging. The main clinical applications are at present in obstetrics, in areas such as assessment of cleft lip [1, 2]. Other research groups have explored 3D ultrasound in areas such as cardiac assessment [3, 4] and evaluation of the foetal brain [2, 5] using a single 3D volume. In our experience at a tertiary referral centre, 3D ultrasound is not routinely used despite it being readily available. The combination of 3D ultrasound and extended field of view may facilitate a wider use of 3D ultrasound by allowing ultrasound practitioners to image larger structures. It is well recognised that ultrasound image acquisition and interpretation is more operator-dependent than other imaging modalities such as CT or MRI. Providing 3D extended-field-of-view ultrasound may aid the interpretation of ultrasound images by demonstrating the anatomical relations of pathology pictorially to clinicians.

There are three ways to acquire 3D data [6]. The most obvious one is to have a 2D array of transducer elements in contact with the patient. This can record 3D data directly by using different combinations of elements to scan the volume of interest. Because dense arrays of 2D

transducers are difficult to build, alternative approaches have been developed using 1D transducer arrays to produce 2D B-scan images at known locations in 3D space. The second group of techniques, called freehand 3D ultrasound, involves letting the clinician sweep the 1D transducer array manually across the subject, coupled with some sort of a tracking device to measure its trajectory. The third group involves sweeping the 1D transducer array through a known trajectory using a mechanical device.

With the ability to obtain a 3D data set, the next evolutionary step in 3D ultrasound is to create an extended field of view by stitching several volumes together. This will potentially have many clinical advantages in visualising structures that are too large for a single volume. The mosaicing will also provide the ultrasound practitioner with a compounded volume of higher quality and the ability to visualise the anatomical structures from a variety of angles. A possible solution to stitching is to consider an automatic image-based registration technique that uses a matching algorithm to align overlapping data. In the majority of matching algorithms, the notion of a similarity measure is considered between two volumes, where the target volume is kept stationary while the position of the source volume is changed iteratively until an extreme value of the similarity measure designates a registration point. However, registering ultrasound volumes is a difficult task. The main difficulties associated with

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ultrasound data are that there is an irregular sampling of the acquisition space and the anatomy's appearance varies with the direction of insonification. Furthermore, there are acoustic artefacts relating to variations in the resolution, attenuation and the material properties in the propagation path of the sound waves. These all result in poor registration performance.

It is possible to register volumes using the image data alone [7]; however, in order to improve the speed and robustness of the registration algorithm, a simpler method using a reduced search space is often considered. In the literature, the majority of the mosaicing systems rely on an external position sensor as an input to the image-based techniques [8, 9]. However, image-based similarity measures can be very overlap-dependent [10]. This, combined with the fact that position sensors may not be well suited to routine clinical use, has led to the development of a modified version of 3D SIFT (scale-invariant feature transform) to globally register ultrasound volumes [11], which, rather than using an intensity-based approach, uses common features (blobs and corners) to match two volumes without using any position tracker. The disadvantage to this method is that it requires the pre-processing of volumes to remove artefacts.

The aim of this work is then to incorporate the benefits from the existing strategies, as previously described, to produce a clinical system that offers unique advantages in a clinical setting. This system does not require an inconvenient external position sensor attached to the probe as much of the information required to calculate the probe trajectory can be inferred by matching the 3D blocks of recorded data. However, we do consider a miniature inertial orientation sensor to guide the matching algorithms, increasing their speed and reliability. Such a sensor could eventually be incorporated in the probe housing; hence it will not be inconvenient in a clinical context. The complete solution is then determined from both the inertial sensor and the three-degrees-of-freedom (3-DOF) registration strategies discussed in this paper. The aim of our initial feasibility study is to assess the success rate of our proposed clinical system in a series of patients attending for obstetric scans.

Methods and materials

The study took place in a large teaching hospital. Ethics committee approval was obtained and all participants provided written consent. Over a 5-month period, 49 3D data sets were recorded from 28 pregnant females attending for routine obstetric ultrasound scans.

Initially a routine conventional ultrasound exam was performed using a ProSound Alpha 7 with a curvilinear 3–5-MHz probe (Tokyo, Japan) during which the routine anatomy checks and measurements were taken in accordance with local department protocol. Following this, multiple 3D volumes were recorded using an Ultrasonix Sonix RP scanner (Ultrasonix Medical Corporation, Richmond, Canada) running Stradwin software (<http://mi.eng.cam.ac.uk/~rwp/stradwin>) with a curvilinear 3–7-MHz mechanically swept 3D probe. A 3-DOF inertial orientation sensor

(InertiaCube3TM; Intersense, Billerica, MA) was used to record the orientation of the ultrasound probe continuously during the scan. This was rigidly attached to the probe, as illustrated in Figure 1, and was calibrated at the start of the study ([12] and Appendix A). Acquiring each sequence of 3D volumes took approximately 5 min and care was taken to ensure that each volume overlapped with the previous volume. Prior to acquiring the data set the sonographer selected the most appropriate scan depth and frames per volume (either 51 or 71 frames per volume). The acquired data were anonymised and stored on the ultrasound machine's hard drive for registration at a later date. Information about the participant's age, gestational age, body mass index (BMI), number of volumes, image depth and frames per sweep were also recorded.

Data sets containing fewer than four volumes were excluded before registration. In the remaining data sets the multiple volumes were registered to one another by an experienced sonographer with 6 years' clinical experience (Observer 1). The registration was carried out on a 3-GHz dual-core PC using the Nelder–Mead simplex algorithm approach [13, 14]. Figure 2 shows the user interface of the Stradwin software and an example of the registration process using a large test data set (<http://youtube/chIAenesKHk?hd=1>). The algorithm considers only a subset of the data on slices through the volumes. The implementation of the algorithm allows a fast (S1) and a robust (S2) option in which the slices are adjusted to cover a smaller or larger proportion of the volume, with a corresponding reduction or increase in registration time. Initially the algorithm was run on all volume pairs with the fast setup

Figure 1. Three-dimensional probes with jigs and attached orientation sensors.

Figure 2. Stradwin (<http://mi.eng.cam.ac.uk/~rwp/stradwin>) user interface displaying test scan.

(S1), where the slices covered just the central portion of each volume (Figure 3). Each pair was visually assessed for whether they had correctly registered and was deemed to be successful if sonographic image features were seen to line up correctly in the two volumes (Figure 4). For those pairs that did not register correctly, the algorithm was rerun in its more robust mode (S2) where the slices covered a larger portion of the data (Figure 3). The time taken to run each stage of the algorithm was recorded, as well as the total registration time for the data set, which included checking the volumes for correct registration and the storage of images. The registration of a data set was deemed successful if four or more consecutive volumes

registered. A cut-off of four consecutive volumes was chosen as this covered a reasonable area beyond what can be acquired in a single sweep. In addition, the Stradwin software was used to automatically measure the total distance imaged in the direction of probe motion for the longest registered sequence of volumes.

The subjective assessment of whether volumes registered correctly was repeated by a second non-clinical member of the research team (Observer 2), who was blinded to the previous results. Observer 2, although non-clinical, has been working in ultrasound research for more than 4 years. Using a second observer enables the repeatability of the subjective assessment to be assessed, thus mimicking the clinical setting when ultrasound

Figure 3. Image showing the difference between S1 (fast algorithm implementation option) and S2 (robust algorithm implementation option); S2 covers a larger area than S1.

Figure 4. Example of two images that are correctly aligned following registration.

images are reviewed by several different operators. From these data the agreement between the two independent observers could be calculated and the longest sequence of correctly registered volumes for which the observers agreed upon recorded. Pairs of volumes where both observers agreed registration had failed were then reviewed. Volumes which failed to register were examined for overlap and foetal movement; if there was good overlap and no foetal movement between the two volumes the failure to register was caused by the algorithm.

Results

In total, 49 3D data sets were recorded from 28 pregnant females. The average age of participants was 29 years (range 17–39 years); mean BMI for the 24 females whose BMI was recorded in their obstetric notes was 23.7 (range 17.6–33); and mean gestational age was 23 weeks and 4 days (range 19 weeks and 6 days to 35 weeks and 6 days). In total 22 females attended for second trimester scans and 6 for third trimester scans. Two data sets were excluded from further analysis because they contained fewer than four volumes of data. Of the remaining 47 data sets, 32 were recorded using 51 frames per volume and 15 were recorded using 71 frames per volume. The number of volumes per data set, scan depth and time taken to register the 47 data sets using algorithms S1 and S2 are summarised in Table 1. The number of data sets that met the criteria for successful registration (four or more consecutive volumes) are shown in Table 2. More than 50% of the data sets registered in 43/47 cases for Observer 1 and 40/47 cases for Observer 2. In 26/47 cases both observers completely agreed on whether or not the data had registered; agreement between the two observers was over 70% in 40/47 data sets. The longest registered sequence of agreement was $\geq 50\%$ of the data set in 35/47 cases. Figures 5 and 6 show examples of correctly registered data sets.

42/267 (16%) pairs of volumes for Observer 1 and 73/267 (27%) for Observer 2 in this pilot study did not register. There were 41 pairs of volumes where both observers agreed the volumes failed to register. Of these, four failed owing to no image overlap, seven failed owing to foetal movement and one owing to both no image overlap and foetal movement. 29 pairs of volumes failed to register owing to the algorithm and, of these, 16 were taken of fetuses during the third trimester of pregnancy. In total 22/41 failures were taken during the third trimester of pregnancy.

Discussion

The results of this pilot study have shown that 3D volumes of the foetus can be successfully registered to create 3D extended-field-of-view images in a high percentage of data sets (77–83%). This, however, relies on good overlap between volumes and a relatively still foetus. The success of the algorithm appears to decrease with gestational age, which was responsible for the most registration failures. This high failure rate in the third trimester of pregnancy is probably due to there being

Table 1. Summary of the number of volumes per data set and time taken to register each using S1 and S2 for each data set

Parameter	Mean	Range
Total number of volumes recorded	6.7 volumes	4–13 volumes
Scan depth	13.8 cm	9.8–18 cm
Total registration time including imaging	22 min	9–56 min
Number of volumes to which S1 was applied	6.7 volumes	4–13 volumes
S1 registration time	10 min 15 sec	3 min 16 s–12 min 36 s
Number of volumes to which S2 was applied	1.2 volumes	0–7 volumes
S2 registration time	2 min 43 s	0–18 min 47 s
Total distance imaged in longest registered sequence	15.51 cm	7.39–33.86 cm

S1, fast algorithm implementation option; S2, robust algorithm implementation option.

Table 2. Summary of the registration success for Observer 1, Observer 2 and agreement between the two observers (a successfully registered data set is one where four or more consecutive volumes registered)

Observer	Registered	Non-registered
Observer 1	39/47 (83%)	8/47 (17%)
Observer 2	36/47 (77%)	11/47 (23%)
Agreement between Observers 1 and 2	35/47 (74%)	12/47 (26%)

Figure 5. Example of a registered three-dimensional extended-field-of-view image taken of a 20-week-old foetus.

less amniotic fluid surrounding the foetus later in pregnancy, causing the edges of the foetus to be more difficult to distinguish.

Our results are less successful than previous results [14], where the Nelder–Mead simplex algorithm was tested on 16 two-volume data sets comprising 5 *in vitro* scans of a speckle phantom containing 5-mm inclusions and 11 *in vivo* scans of throat, calf and abdomen. A total of 1280 test runs of the algorithm were performed on these 16 data sets with varying numbers of image slices and initialisation depth. We found the success rate to vary between 85% and 97%, with a higher success rate (>95%) in those cases when we either considered a higher number of slice pairs or limited the search range

in the axial direction. The results are comparable with those of Ni et al [11], who demonstrated 83% success for non-clinical data and 85% for clinical data. These results were only based on one clinical data set, rather than a range of patients [11]. Our results, from 47 clinical data sets, are therefore more reflective of the clinical scenario when a registration algorithm would be required to work on a range of patients.

There was reasonable repeatability between observers for whether volumes had registered ($\geq 70\%$ agreement in 40/47 data sets). The clinical observer was, however, more lenient than the non-clinical observer (Table 2). This is perhaps due to the sonographer being more familiar with clinical data, which by definition are

Figure 6. Example of a registered three-dimensional extended-field-of-view image taken of a placenta at 20 weeks.

imperfect. The non-clinical observer, by contrast, is more familiar with phantom data, which are more precise and therefore may have had higher expectation of a registered image. Both observers had a similar amount of experience in their particular areas of ultrasound.

The short acquisition time of 5 min for 3D extended-field-of-view ultrasound makes its use in routine clinical practice feasible. The subsequent post-processing of the data and imaging required on average 22 min; although this is longer than the time taken to reconstruct CT and MR images, it is not an unrealistic in the clinical setting, and one would anticipate that as the technique develops reconstruction times will shorten. 3D extended-field-of-view ultrasound, in addition to imaging the foetus, is able to visualise the entire placenta in the second trimester, which may facilitate further research between the association of placental volume and pregnancy outcome [15]. In addition, visualisation of the entire amniotic sac may enable more accurate assessment of amniotic fluid volume, an indicator of foetal well-being.

In conclusion, this study has shown that 3D extended field of view in obstetric ultrasound is possible. Our preliminary data were taken from a small cohort of patients, and a further larger study is necessary. A future study should also include a greater breadth of gestational ages.

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Appendix A

Calibration of the inertial sensor

An inertial orientation sensor provides information about orientation changes relative to three axes passing through the electrical centre of the device. In order to determine the correct orientation of the blocks of acquired ultrasound data, we need to know changes in the orientation of the ultrasound probe. We therefore need to know the three fixed rotations that map from the co-ordinate system of the inertial sensor onto the co-ordinate system of the ultrasound probe. Once the sensor has been rigidly fixed to the probe these rotations are also fixed. It is therefore only necessary to determine them once (unless, of course, you remove the sensor from the probe). The process of determining these three rotations is generally known as spatial calibration.

We use a phantom-based technique to perform this calibration. The idea is to scan an object of known geometry and reconstruct a 3D shape from the resulting data. Only if the rotations used in the reconstruction are correct will the 3D reconstruction be valid. We choose to scan a flat plane and use a non-linear optimisation algorithm to infer rotational parameters that enable it to be correctly reconstructed as planar. A key requirement for success in this approach is that the scan be performed from a sufficient variety of directions.

Full mathematical details of the process and an explanation of how we are able to perform this technique in a semi-automated way are given by Housden et al [12].