IMAGE DATA BASED METHOD FOR 3D ULTRASONIC DATA POSITIONING

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Abstract: In the free-hand 3D ultrasonography, the position of the us. probe of a 2D ultrasound scanner is sensed by a special positioning subsystem so that the us. data are accompanied by the positional data. The resulting 3D data should permit flexible visualization and more accurate volume measurements than can be achieved using isolated 2D B-scans; however, the use of the position sensor may be inconvenient for the clinician and, moreover, the main disadvantage is that these position sensors are not accurate enough and calibration of the measuring equipment may be too complicated. The objective of the presented work is thus to replace the positioning sensor with a technique to estimate the probe trajectory based on the B-scan images themselves. Some publications on such techniques exist, based on speckle correlation analysis, e.g. [5]-[6]. In this paper, we propose a method of a different kind expected to enable accurate positioning using image-data registering based algorithm, including possible modifications.

Introduction

The 3D ultrasonic data resulting as a data block compiled from multiple 2D scans should permit flexible visualization and more accurate volume measurement than can be achieved using 2D B-scans. The 3D ultrasound data can be reconstructed from the standard 2D ultrasound images if the position and orientation of each 2D image is known. Mechanical systems have been used to acquire the position and orientation information [1]-[2], but these are rather impractical. Remote localization systems, such as electromagnetic ones, allow a freedom in image acquisition covering the desired field of view with good insonification of the structures of interest. Previous studies using a magnetic sensor system have demonstrated the feasibility of reconstructing the 3D images from 2D slices [3]-[4].

However, the use of the position sensing systems, besides being inconvenient for the clinician, has the disadvantage in not being accurate enough and also, its calibration may be too complicated. The objective of this work is thus to substitute the positioning system with a correlation based technique estimating the probe trajectory from the obtained B-scan images themselves. Approaches of this kind have been described, e.g. in [5]-[6].

The speckle-correlation based algorithms have still some problems; it is necessary to estimate some parameters of the expected correlation function before performing the curve fit, such as the mean ultrasound wavelength and the distance from the transducer to the imaged point. In this paper, a method is proposed for accurate positioning based on speckle-correlation algorithm, including some modifications.

Materials and methods

Elevational speckle correlation: Tuthill et al. propose an automated 3D US frame positioning system [5], where B-scan images are collected by means of a handheld transducer moving in the elevation direction (perpendicular to the scan plane) and frame spacing is computed by a speckle-correlation algorithm, without additional positioning hardware. The concept of this method consists in monitoring and evaluating the scan plane motion in the elevational direction via spatial correlation of the speckle texture. It has been found that the autocorrelation intensity can be approximated by a Gaussian function with respect to frame spacing.

Since the correlation algorithm applies only in the similar speckle regions, we need a speckle detector to find the speckle region. For the fully developed speckle, the intensity image is characterized by exponential distribution and the constant ratio 1 of the mean to standard deviation (SD). In [5], a detector based on the first-order statistics was developed. When the mean-to-SD ratio for a 3D region was between 0.8 and 1.2, the region is classified as a speckle region. Another method to detect speckle regions using the statistics of a homodyned *k*-distribution can be found in [7].

Only the pixels in the speckle regions are used to form the correlation (or better the autocovariance) function. The autocorrelation (R_A) for positions r_1 and r_2 is defined as

$$R_A(r_1, r_2) = \langle A(r_1) A(r_2) \rangle , \qquad (1)$$

while the autocovariance (C_A) is

$$C_A(r_1, r_2) = R_A(r_1, r_2) - \langle A(r_1) \rangle \langle A^*(r_2) \rangle , \qquad (2)$$

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where A means the intensity function and $\langle\ \rangle$ means the expected (mean) value.

The elevation profile of the normalized intensity correlation function can be fitted by a Gaussian function [6]

$$C(\Delta y) \approx \exp[-2a_0(\Delta y)^2], \quad (3)$$

where Δy is displacement in the elevation direction, $a_0 \propto 1/(\lambda_0 z)^2$ with λ_0 being the ultrasound wavelength at its central frequency, z is the distance from the transducer to the ultrasound field point. These coefficients are determined by the transducer properties. By fitting the covariance function for a set of consecutive frames with a Gaussian curve, the average frame spacing can be estimated.

The approximation (3) uses some transducer properties. In fact, some of these properties, such as the transducer's focus and the depth of ROI (region-of-interest, i.e. region drawn by a physician, witch includes object or objects of interest, e.g. tumor), are dependent on the scanning conditions. However, the ratio of the frame spacing can be easily obtained without any transducer properties, based on the correlation function [5]. According to equation (3), the profile correlation function values of frames f_i, f_j and frames f_j, f_k in a 3-D data set can be defined individually as follows:

$$C(f_i, f_j) \approx \exp[-2a_0 D(f_i, f_j)^2], \quad (4)$$

$$C(f_j, f_k) \approx \exp[-2a_0 D(f_j, f_k)^2], \quad (5)$$

where f_i, f_j, f_k are the frames in the 3D dataset, D is the distance between of the two frames, and C is the correlation function of the two frames. On the basis of these equations (4) and (5), we can get the ratio of all the frame spacing in the 3D dataset from equation (6). In fact, there is an assumption of parallel scans, so the rotation of the transducer is not allowed.

$$\frac{D(f_i, f_j)}{D(f_j, f_k)} = \sqrt{\frac{\ln(C(f_i, f_j))}{\ln(C(f_j, f_k))}}, \quad (6)$$

The proposed approach: In our modification, a technique based on image data matching to estimate the approximate position of each frame is used, without a need to estimate any coefficients. In fact, a reference frame is needed, which we obtain by scanning one frame in the direction perpendicular to the scan planes of the main data. The relationship between a reference image and the 3-D data set is depicted in Figure 1. Then, the intersection lines of the images with the reference frame can be found by registering the appropriate data in the reference- and data-images. Consequently, frame spacing can be directly computed from the distances of these lines [8]. The procedure to find the intersection lines is based on some assumptions. First, during the data-image acquisition, the transducer must be moved forward unidirectionally and monotonously. Second, the transducer must be moved along almost a straight line.

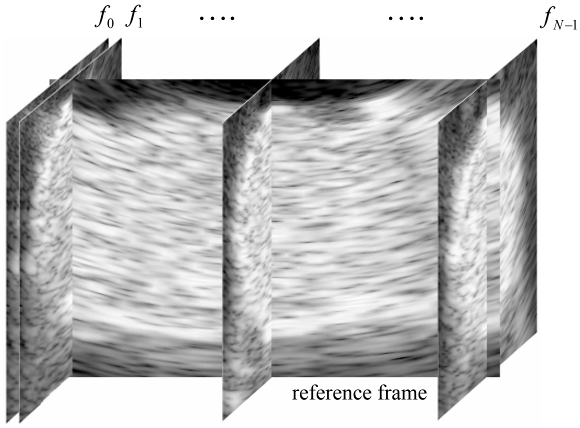


Figure 1:The relationship of the frames in the 3-D data set and the reference frame

Image registration: In case of finding intersectional lines, we tested one of the simplest voxel similarity measures, which is the sum of *square intensity differences (SSD)* between images [9]:

$$SSD = \frac{1}{N}\sum_{k=1}^{H}\left|A(i,k) - B(r,k)\right|^2, \quad (7)$$

where i is a line number in frames, r is a line number in reference frame, A and B are the intensity functions of pixels, H is the height of the ROI and k is the index of height in the ROI.

SSD measure is very sensitive to a small number of voxels that have very large intensity differences between images A and B, so we tested *sum of differences (SAD)* too:

$$SAD = \frac{1}{N}\sum_{k=1}^{H}\left|A(i,k) - B(r,k)\right|, \quad (8)$$

Finding principle intersectional line: Due to noise and blurry nature of ultrasonic images, it is hard to find the correct intersectional line, even using the SSD (7) or SAD (8) algorithm described above. Thus, we need some restrictions and assumptions before using the image registration algorithm. First, we must select the best reference frame to obtain a good result. Here, the best reference frame means that this frame should contain as much of the object of interest as possible. In the same way, choosing the best frame in the 3-D data set that contains as much of the object of interest as possible. Because the ROI is drawn by a physician, the center of the object of interest can be expected to approximate the center of the ROI. We can get the intersectional line using these two frames and this assumption. We search all of the combinations of the two lines, in which one is from the central region of ROI in the frame f_m and the other is from the central region of the ROI in the reference frame. Calculating the function SSD, SAD respectively, for each set of the two lines the optimum is found: the two lines that have the minimum distance will be considered the same line, i.e. the intersectional line of the frame f_m and the reference frame.

The range of the central region of the ROI in the reference frame and the frame f_m can be set to 10% of the width of the ROI. For the same object of interest, the corresponding lines of the ultrasonic images that contain the biggest region of this object in the reference direction and in the main direction may be very similar. However, if the search range is too large, we may get the wrong intersectional line. If we can ensure that the reference frame and the frame f_m that we chose are very accurate, then we can decrease the search range to increase the accuracy of the found intersectional line [8].

Finding other intersectional lines: Now, we explain how to derive the position of the frame f_{m-1} from the position of the frame f_m. If the frame f_m intersects the reference frame in line i of the frame f_m, then we will assume that all the other frames of the 3-D data set intersect the reference frame at their i -line. In the elevational direction, if the frame f_m intersects the reference frame at line j of the reference frame, then the frame f_{m-1} must intersect the reference frame at the line that is in the left side of line j of the reference frame. The line of minimum SSD function (SAD respectively) in the reference frame is the intersectional line of the frame f_{m-1} with the reference frame. Repeatedly using this method, we can obtain the intersectional lines of the frame f_{m-1} to the frame f_0 with the reference frame. In the right side of the frame f_m, the intersectional lines of the frame f_{m+1} to the frame f_{N-1} with the reference frame can be obtained in a similar way.

The search range of the reference frame when we perform the line matching is adaptive. If we do not limit the search range, it may happen that the distance of the two intersectional lines is too far, even if these two intersectional lines are formed by two consecutive frames with the reference frame. This is an obvious error because the frames in the 3-D data set are continuous, and this error will cause all the intersectional lines obtained after this intersectional line to be wrong. To prevent this error from happening, we set the search range to 10% of the width of the ROI initially. If the final result of the intersectional line of the frame f_0 with the reference frame was out of the ROI, we decreased the size of the search range. Alternatively, if the final result of the intersectional line of the frame f_0 with the reference frame was inside the ROI, we increased the size of the search range. This method was repeatedly used until the intersectional line of the frame f_0 with the reference frame was near the boundary of the ROI. The search range in the right side of line j of the reference frame can be set in a similar way. After finding the search range, all the intersectional lines of all the frames in the 3-D data set with the reference frame can be obtained, so we get a set P of the positions.

Speckle correlation for the ratio of frame spacings: From equation (6), the ratio of all the frame spacings in the 3-D data set can be computed without additional parameters. First, extract three frames (f_0, f_1, f_2) and get the ratio of the $D(f_0, f_1)$ and $D(f_1, f_2)$. Then extract three frames (f_1, f_2, f_3) and get the ratio of the $D(f_1, f_2)$ and $D(f_2, f_3)$ in the same way. Repeatedly using this method, the frame-spacing ratios in the 3-D data set can be calculated.

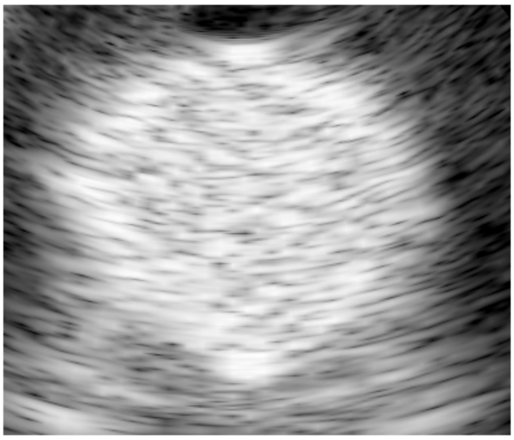
Because the speckle correlation algorithm only provides the ratio of frame spacings, we need to know the positions of the frame f_0 and f_{N-1}. When the positions of the start frame f_0 and the end frame f_{N-1} have been known, the other frame positions can be obtained through the ratio of frame spacings. Here, we directly use the positions of the frame f_0 and f_{N-1} from the set of positions P, obtained in previous steps by the image registration. Then we can get another set of positions P', which provides the approximate positions of all the frames in the 3-D data set calculated based on this speckle correlation algorithm.

Final position refinement: Because the set of positions P' is obtained based on the speckle correlation algorithm, the errors that are caused by this algorithm directly influence the accuracy of the frame positions P'. The original set of the frame positions P can assist in refining the positions and reducing this error. For each frame f_i between the frame f_0 and f_{N-1} in the 3-D data set, we refined the position of frame f_i in a small range. This small refinement range for frame f_i is between $(f_{i+1} - f_i)/2$ and $(f_{i+1} - f_i)/2$. In this range, the position closest to the position of the frame f_i in the set P is the new position of the frame f_i. In this way, all the frame positions of the 3-D data set can be refined.

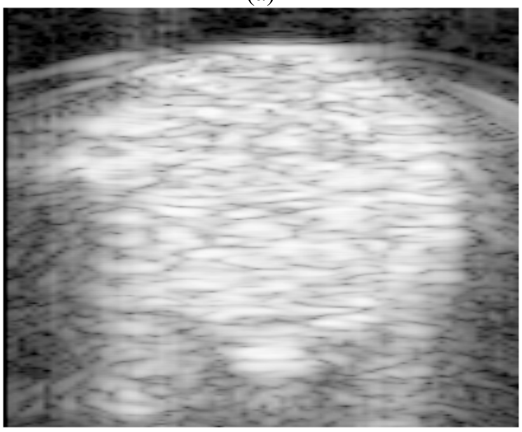
Experimental data acquisition

The image dataset of B-scans for our experiments was captured by VingMed System FiVe Echocardiography System (GE Medical System) modified for RF acquisition with a 2D transducer, mounted on the special holder together with the position sensor. The position and orientation information were obtained via electromagnetic MiniBird System (Ascension Technology Corporation). The information from the position system will be used just to subsequently compare our results with the position data provided by the use of the MiniBird positional system.

The dataset was captured via transducer movement in two directions, the transversal and the longitudinal direction. In fact, only one direction is needed to reconstruct the 3D volume; the other direction is used as the reference for reconstructing the 3D volume. We need only a single frame along the reference direction to be used as a reference for image matching. We assume that the transducer is translated in parallel to the body surface and the movement results in axis-aligned images. The rotation of the transducer is not allowed so that the data frames may be considered mutually parallel.



(a)

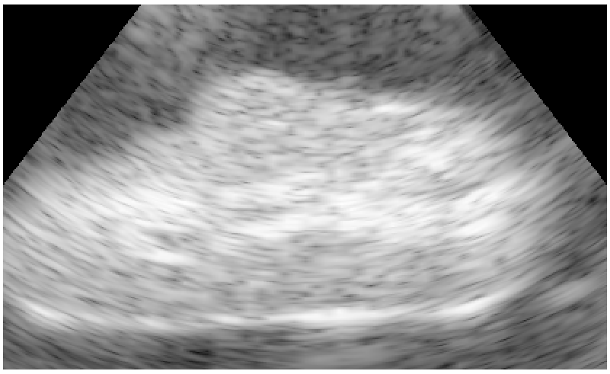


(b)

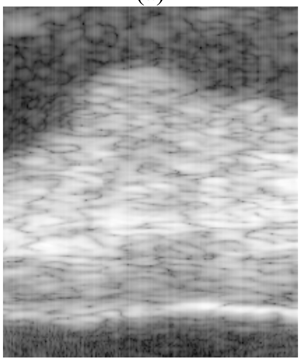
Figure 2: (a) The reference frame of white flour mixed with water in synthetic bowel dipped into the water – transversal view. (b) The reconstructed reference frame from the longitudinal frames

Results

In this section, we present some experimental results. For the present, we tested this approach on our experimental data sets which we aquired as described above. First data set is white flour mixed with water in synthetic bowel dipped into the water (Figure 2). The second one is a pig kidney placed also into the water (Figure 3).



(a)



(b)

Figure 3: (a) The reference frame of a pig kidney placed into the water – transversal view. (b) The reconstructed reference frame from the longitudinal frames

Conclusions

We have proposed a method which combines the speckle correlation algorithm and the image matching technique. This system is supposed to enable accurate compilation of the 3D dataset from the 2D ultrasonic B-scans acquired by free-hand scanning without any additional positioning hardware; in fact, just a reference frame is needed additionally for the compilation. There are some difficulties and limitations with our positioning system. We assume that all the frames in the 3-D data set and the reference frame are almost parallel or orthogonal. The rotation of the transducer is not allowed in the speckle correlation algorithm or the image registration method. In the present, the work is focused on implementation, verification and further improves of this approach.

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