Image Based Calculation of Elevational B-Scan Separation

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Abstract: A speckle decorrelation technique was employed to calculate the elevational (out of plane) separation of B-Scan images. To calibrate the relation between frame separation and speckle decorrelation a phantom was scanned uniformly. The calibration data was used to calculate the frame separations in uniformly and non-uniformly spaced phantom and breast scans. The error relative to a mechanical reference position encoder increased with frame separation (25 μm to 300 μm). Using a 300x100 pixel subimage, the frame position was tracked with an average error of 3% in the phantom and 5% in the breast scan.

INTRODUCTION

In 3-D ultrasound, the position of each frame within a volume scan must be determined accurately. Current position measurement techniques require additional hardware, such as an articulated scan arm, electromagnetic positioner or acoustical time-of-flight position encoder. We proposed an alternative method that is based on the speckle decorrelation between parallel frames to estimate the separation distance (1). If extended to full 3-D position tracking, this technique would then allow the use of ordinary, freely moving, handheld transducers for volume imaging.

MATERIALS AND METHODS

In this study, the elevational separation of scan planes was estimated from frame-to-frame speckle decorrelation using second-order speckle statistics. It can be shown (2) that for a rectangular linear array probe the echo intensity autocovariance near the focus is a Gaussian. The normalized intensity elevational covariance for a pixel in the focal region then also has a Gaussian shape. For an elevational transducer velocity with respect to the insonified tissue of v_x [mm/frame], the intensity covariance, also referred to as the correlation function, can be written

$$C(\Delta x, z) = \exp(- (v_x \, \Delta x)^2 / (2 \, \sigma_v^2(z)))$$

where Δx is the frame lag, z is depth [mm], and $\sigma_v(z)$ is the depth-dependent beam correlation width which can be obtained by a transducer calibration. Samples of the correlation function are calculated from pure speckle regions in a subset of typically 10 images. The average frame spacing in this subset can then be obtained by curve-fitting of the calculated data to a Gaussian. The underlying assumption of a constant transducer velocity during acquisition of the subset is justified if the frame rate is high enough so that the frame-to-frame change in transducer velocity is small.

To determine regions of pure speckle, first-order speckle statistics were employed (3). For fully developed speckle, the ratio of the mean intensity to the standard deviation (MSD) is 1.0. Thus a speckle selecting algorithm was created that calculated the average MSD for a moving 3-D volume intensity region. A binary mask was formed displaying regions where the MSD lay in a small acceptance range around 1.0. Window size (21×21 pixels by 10 frames) and acceptance range (0.9 to 1.1) were selected as an acceptable compromise for both spatial resolution and rejection of non-speckle regions. Only pixels in the mask were then used to calculate the correlation function.

A LOGIQ 700 (General Electric Medical Systems, Milwaukee, WI) scanner with a 13 MHz 1.5-D array probe was used for the acquisition of the scans. For calibration, a focal lesion phantom (CIRS, Norfolk, VA) was scanned in a region with a high density of subresolvable particles and very few specular reflectors. The transducer was mounted on a mechanical micro-positioner to obtain a set o f uniformly spaced (25.4 μm) frames in an elevational sweep. The speckle mask was calculated for a central ROI (300×100 pixels) in each frame. The average covariance was computed for each row in the ROI and curve fit to a Gaussian. The depth-dependency of their standard deviations was approximated by a polynomial and scaled by the step size to determine the beam correlation width $\sigma_v(z)$. In the following scans, the same ROI and scanner settings were used as for the calibration.

From the calibration scan, sets with larger frame spacings were created by taking subsets containing every 2nd, 3rd, etc. frame of the original. These subsets were used to verify the scaling of the estimated spacings and to find the maximum spacing to which accurate separation estimations are possible.

The speckle selection and frame spacing computation were then tested on uniformly spaced scans from portions of the phantom containing small, spherical regions of weak scatterers (3 mm "voids"). Next, the algorithms were applied to non-uniformly spaced elevational scans of the phantom and to breast scans. A linear free-scan system with electronic position readout (accuracy ± 18 μm) was used to obtain sets of images and reference position data. During the scans, the transducer was fixed perpendicular to the scan direction. Frame spacings and total distances were computed using the calibration data and compared to the reference values.

RESULTS AND DISCUSSION

The scaling of the uniformly spaced calibration set showed excellent agreement between calculated and reference spacings with less than ±3% error for frame-to-frame separations up to 125 μm. For a 300×100 pixel ROI vertically centered around a 30 pixel long focal region formed of two separate transmit foci, the error increased steadily to -6.1% at 280 μm spacing. For a smaller ROI entirely in the focal region the standard deviation of the estimations increased due to smaller statistics, but the overall error was reduced since the theoretical relationship between speckle decorrelation and frame spacing only holds in the focal zone.

Scans of phantom regions containing speckle and voids demonstrated the efficacy of the speckle detector that recognized the specular reflections of the boundaries of the voids and did not include them in the mask to be used for the spacing estimation.

Free-hand linear scans of the phantom were then obtained where the operator moved the transducer at variable speeds during each scan. In "pure" speckle regions as well as in regions containing voids the algorithm tracked the actual frame spacing well. The average error was 7% for the frame-to-frame spacing and 3% for the total distance from the beginning of the scan. Fig. 1 shows a typical example of calculated and reference spacings where the velocity of the transducer changed considerably during the course of manual translation.

Finally, a free-hand linear scan of a breast containing a cancerous lesion was obtained. The area marked as true speckle, particularly in and around the tumor, was less than 5% of the total ROI as opposed to roughly 40% in the phantom scans. Accordingly, the error in the separation estimation increased but the position data was still tracked with a 5% accuracy (Fig. 2).

The technique described here can now be extended to multidimensional transducer motion, ultimately providing pure image-based registration of frame positions in 3-D ultrasound scans.

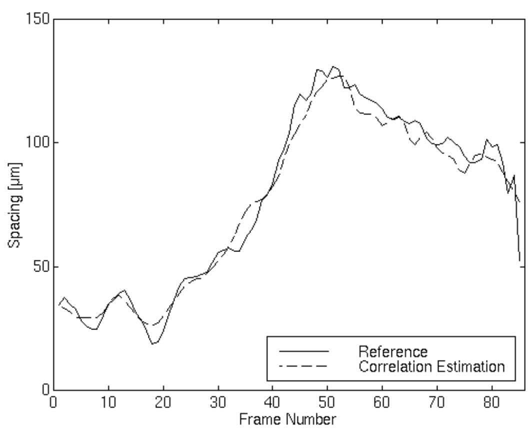

FIGURE 1. Calculated and reference frame spacing in a two-speed scan of a phantom containing "pure" speckle.

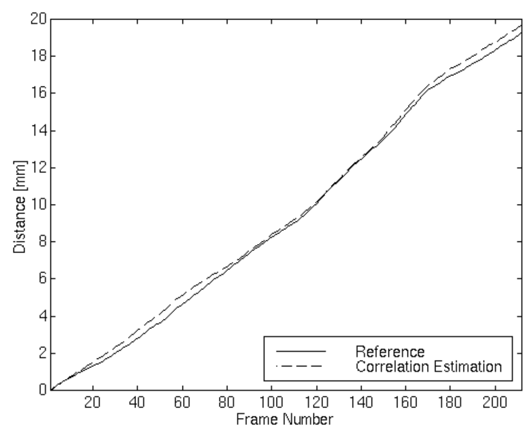

FIGURE 2. Calculated and reference values for the total distance traveled by the transducer in a breast scan.

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