Navigator Echo Based In-Slice Motion Artifact Suppression in Ultrasound Images Based on Synthetic Aperture

Abd El-Monem El-Sharkawy, Abou-Bakr M. Youssef, and Yasser M. Kadah
Biomedical Engineering Department, Cairo University, Giza, Egypt.

ABSTRACT

We propose two techniques for robust motion artifact suppression in ultrasound images based on what is called "navigator echoes". The navigator echo approach has been introduced and widely used in the area of magnetic resonance imaging where motion is detected along one of the image directions based on simple registration of one-dimensional projection of the image assuming rigid body motion. In ultrasound imaging, the same concept can be applied to correct for simple shifts (e.g., with linear array probes) or rotations (e.g., with convex array probes) along the width and depth direction. The first technique assumes a rigid body motion model. Hence, simple one-dimensional template matching can be used to obtain the motion parameters as in magnetic resonance imaging. Alternatively, the second technique considers a more general spatially-variant motion model. The motion parameters of this model can be obtained through localized optimization of an information theoretic criterion. The motion model parameters are then employed in the reconstruction thus providing images with substantially reduced artifacts. The proposed method was implemented on an experimental ultrasound system in which each line is obtained from 2-4 acquisitions from interleaved frames. The motion models are compared and their practical implementation for clinical systems is evaluated.

Keywords: Synthetic Aperture, Motion artifact, Navigator echo, information theoretic image enhancement.

1. INTRODUCTION

Ultrasound imaging is among the most versatile imaging technologies available today. Utilizing non-ionizing radiation and a variety of applications, its use has become a routine in medicine for diagnosis and follow-up of patients. Even though the basic theory of B-mode imaging has been established a long time ago, advances in its implementation to provide higher resolution artifact-free images are still a classical part of medical imaging research until now. A major part of this research is directed towards ultrasound imaging using electronic arrays. As a general rule, the spatial resolution of ultrasound imaging scales directly with the aperture size in such systems. That is, the number of array elements used to collect the ultrasound echoes increases for any element size. This is a direct consequence for the fact that the beam can be focussed better and hence the lateral resolution becomes superior when the aperture size becomes larger. Phased array systems represent the optimal scenario where all array elements are used in signal transmission and reception to achieve high image quality.

In low-end ultrasound imaging systems, however, only a small number of independent channels is utilized to collect ultrasound echoes in order to reduce the system complexity and consequently its cost. This has the effect of degrading the image resolution and limiting the diagnostic value of the output images. In order to solve this problem, a technique called synthetic aperture was developed to enhance the spatial resolution employing the same number of channels. The basic idea of this technique is to acquire the desired aperture through multiple acquisitions where such acquisitions are combined to form an effective aperture size that is larger than the original. The basic idea of this technique can be further generalized to include interlaced acquisition and averaging. They all share the common feature of reconstructing the image from multiple acquisitions.

Since multiple acquisitions of the same structures must be done at different times, it is likely that the field of view undergoes some changes in between acquisitions. This results in motion artifacts because the image was acquired from acquisitions that do not correspond to the same field of view. Such motion artifacts can be rather severe and limit the diagnostic value of the reconstructed images in most cases. Hence, a technique that can reconstruct artifact free images using multiple acquisitions would be valuable to enhance the clinical use of such approaches.

In this work, we address the problem of reconstructing ultrasound images from multiple acquisitions. This problem applies to synthetic aperture, interlaced, and averaged acquisition modes. We propose two techniques that can be used to solve this problem. The first is based on the a similar idea to that of navigator echo motion correction in magnetic resonance imaging. In this technique, a one-dimensional projection of the acquired raw ultrasound lines before image reconstruction. The one-dimensional functions from consecutive acquisitions are registered together to obtain an estimate of the field of view motion in between acquisitions. Hence, this technique assumed rigid-body motion model for the field of view. On the other hand, the second technique relies on optimizing a local information theoretic objective function during the

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reconstruction as a function of the motion parameter. This enables a more general spatially variant motion model to be considered. The two techniques were implemented and their performance is evaluated and compared.

2. ULTRASOUND IMAGING USING ELECTRONIC ARRAYS

In electronic array imaging, a fixed number of elements is used to construct window. This window is subsequently shifted in small increments through the rest of the array transducer elements. Each window is used for acquiring an ultrasound line in space with a specific aperture size. The acquired line direction is determined by the preset delay values used. In most low cost scanners these values are fixed for each frame and can not be dynamically changed when the window is shifted over the array elements. This is because the beamforming process is mostly hardwired and the software capabilities within this part are usually limited. In particular, software changes to the acquisition beamformer may not be possible except between frames in low-end ultrasound systems. An example of such problem arises with those techniques using interlaced acquisition to increase the frame rate while maintaining the spatial resolution. An illustration of such techniques is shown in Figure 1. These techniques reconstruct the image frame from multiple interlaced field acquisitions.

Degradations in the reconstructed image quality may result due to motion between acquisitions of the imaged object, tissue motion, and/or motion of the scanning probe. In order to compensate for motion artifacts in general a spatially variant global motion model must be considered. Such motion models compensation are computationally expensive, and may not be applicable for on line processing. Simpler motion models are considered for faster image reconstruction. In interlaced imaging the motion of the probe is the most dominant as it causes a total spatial reorientation of the scanning beam and are most likely to occur by the sonographer. Organs motion does not have such effect especially if the frame rate is still high enough.

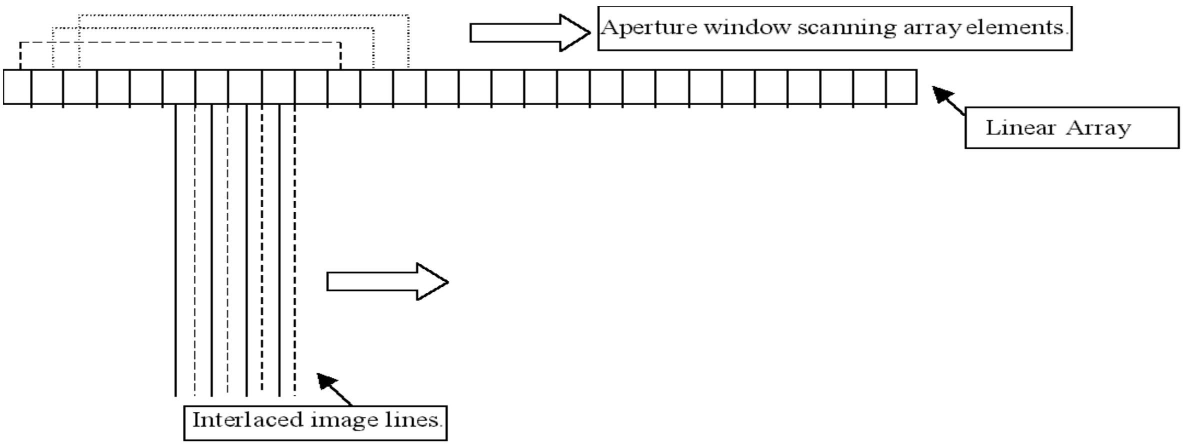

Figure 1: Illustration of electronic array imaging in case of interlaced acquisition.

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3. NAVIGATOR ECHO MOTION CORRECTION

In magnetic resonance imaging (MRI), the process for acquiring one image can last for a minute or more. As a result, motion artifacts are commonly encountered during long acquisition sequences as a result of motion of the imaged subject during the data collection period. In order to reduce this artifact, a technique called *navigator echo* was developed whereby a one-dimensional projection of the image is collected with each acquisition. These navigator echoes are registered together to estimate the translational motion in between acquisitions. Such estimates are subsequently utilized during the reconstruction process to compensate for this motion and yield images with much less motion artifacts.

In ultrasound imaging, a similar technique to navigator echo motion estimation can be devised. Even though it is not possible to obtain this projection information as in MRI, the acquired raw ultrasound lines can be used to obtain this projection information. That is, the correction will be *frame-based* and not *line-based* like in magnetic resonance imaging. Hence, the problem formulation in ultrasound imaging has several unique features as compared to the one in MRI. First, the navigator echo in ultrasound imaging is virtual unlike with MRI where real echo that may be different from the image data is collected. Second, the motion in ultrasound imaging almost exclusively happens in the direction of the probe elements. That is, simple lateral shift in linear array imaging and angular shift in convex array imaging. This reduces the complexity of the problem unlike with MRI where the simple translational motion happens in both image directions in addition to the possible rotational motion within the image plane. Therefore, even though the classical navigator echo approach is limited to simple shift in one dimension, this does not pose any limitations on its use in ultrasound imaging where the actual motion follows this one-dimensional motion model closely.

An example of this approach can be seen in Figure 2. This figure shows a simulated image where the acquisition probe rotates. Under the assumption of stationary field of view in between acquisitions, an object in the middle of the field of view shows a rotational motion in between frames. Given that this object and the field of view structures are assumed rigid between frames, this motion appears as a simple shift in the computed navigator echo in Figure 3. Here the horizontal axis of this projection has units of angle. An illustration of the effect of probe motion is shown in Figure 4.

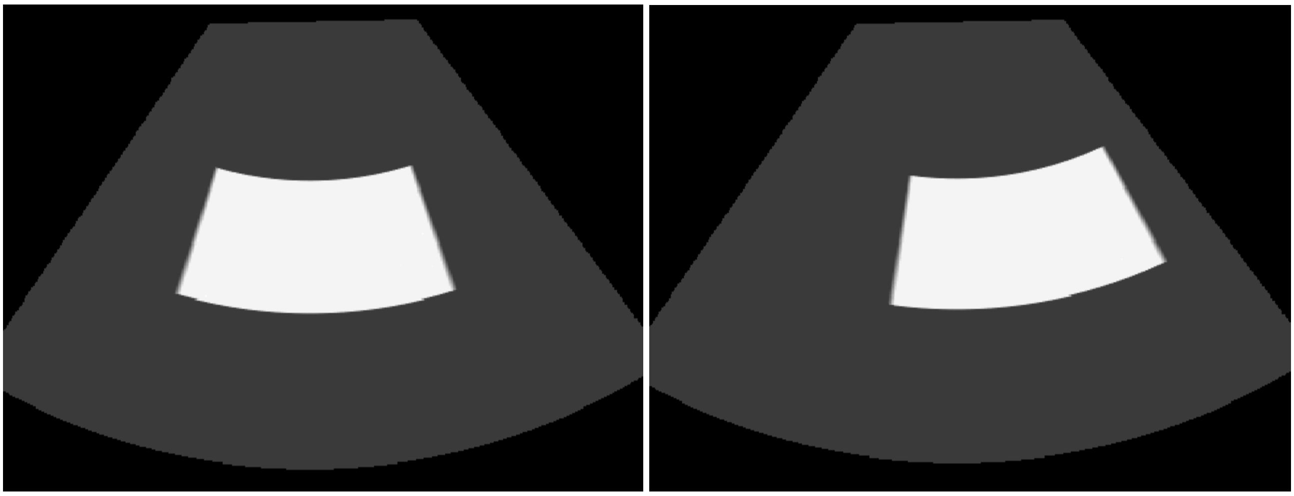

Figure 2: Rigid body motion in the angular direction.

Figure 3: The angular motion in Figure 2 corresponds to a simple linear shift in the navigator echo.

It is clear from this illustration that the total beam power in either the radial or the angular directions can be used as a guide for gross motion detection. The following is a mathematical formulation for the motion detection and correction process.

Consider $F(r,\phi) = F_1(r,\phi) + F_2(r,\phi + \Delta\phi)$ (1)

Where F is the reconstructed image, F_1 is the image whose lines have zero initial shift in space, F_2 is the image with angular shift in space (i.e. interlaced lines), and r and φ are the polar coordinated. The projections in the angular and radial directions can be calculated from the following equations,

$$P_1(\phi) = \sum_{r=0}^{Depth} F_1(r,\phi) \qquad P_2(\phi) = \sum_{r=0}^{Depth} F_2(r,\phi)$$
$$P_1(r) = \sum_{\phi=0}^{Angle} F_1(r,\phi) \qquad P_2(r) = \sum_{\phi=0}^{Angle} F_2(r,\phi) \qquad (2)$$

A correlation vector between the angular and radial power vectors can be computed as,

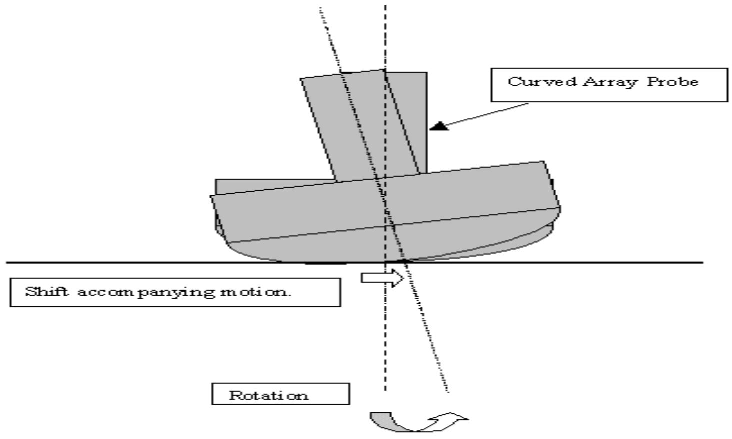

Figure 4: Illustration of probe movement that can lead to motion artifacts

$$Corr_{\phi} = \sum_{i=-Angle}^{Angle-1}\sum P_1(\phi)P_2(\phi - i + 1)$$
$$Corr_{r} = \sum_{i=-Depth}^{Depth-1}\sum P_1(r)P_2(r - i + 1) \tag{3}$$

The angular and radial motion estimates are computed by finding the index of the maximum value in both correlation vectors. The final corrected image can subsequently be reconstructed given the relative motion between the two acquisitions using simple interpolation.

To summarize, this technique can be used to estimate motion in both the angular and radial directions even though the motion in the radial direction is not usually of concern. It has the advantage of being simple and easy to compute for real-time reconstruction. On the other hand, the rigid-body motion model may be limiting in some cases where motion is severe.

4. INFORMATION THEORETIC CORRECTION

Another class of correction technique solves the problem using optimization techniques. The basic idea is to formulate an objective function that relates directly to the motion parameters and optimize this measure (i.e., maximize or minimize depending on the formulation) to obtain the motion parameters that are most likely to have caused the present motion. Hence, the performance of this class of techniques is highly dependent on the choice of the objective function.

Among the most commonly used objective function formulations for motion suppression is the one based on information theoretic criteria. A good example of this class is the entropy focusing criterion given as,

$$E = -\sum_{x,y \in ROI} \frac{I(x,y)}{I_{\max}} \cdot \log\left[\frac{I(x,y)}{I_{\max}}\right], \tag{4}$$

where $I(x,y)$ represents the reconstructed image intensity, and I_{max} is the maximum intensity within the region of interest (ROI). The reconstructed image intensity varies according to the estimate of the motion. In particular, we start with an initial estimate for the motion and perform the reconstruction and repeat the procedure with different estimates until the optimal shift value is obtained. The property of the entropy measure is its preference of single strong peaks over the same peak energy distributed over a large area. Since the motion artifact itself can be modeled as blurring, this measure is very sensitive to motion artifacts and deteriorates rapidly with slight motion. Hence, minimization of the above criterion should provide an accurate estimate for the motion parameter and consequently the sharpest reconstruction for an image.

So, this technique is apparently more general and provides corrected images with a sense of optimality in the procedure. It can be applied to correct different parts of the image independently and hence seem to be more suited in cases of severe within-frame motion. Nevertheless, the technique requires repeated reconstruction of the image regions to perform its iteration. This can be prohibitively slow in many cases. The method used to implement this technique is as follows:

1. A mask of certain number of lines (e.g., 20 lines) is selected. The angular motion is assumed to be within a certain range (e.g., from –5 to 2 degrees) using a pre-selected step (e.g., 0.5 degrees).
2. This mask scans the image sequentially is steps of half the mask size and the entropy focusing criterion is calculated for each shift estimate within the area of the mask.
3. Estimates of minimum entropy are selected as the model of motion of the scanned line.

5. EXPERIMENTAL RESULTS

Data were collected from a New Sonics experimental ultrasound system (International Electronics – Biomedical Division, Egypt) using a 96-element convex array transducer of 3.5 MHz center frequency. The scanning method involved interlacing two frames of different apertures to form a higher resolution reconstruction. The line acquisition time was 350 μsec and the number of lines in each frame was 80. The two available apertures were swapped each frame to collect whole frames with each. A standard B-mode imaging phantom was scanned using this system by rotating the probe. The sequence of frames is collected and analyzed.

Figure 5 illustrates a B-mode scan using a stationary probe. The image is free of artifacts. On the other hand, with a simple movement of the probe, artifacts such as the image in Figure 6 appear. Figure 7 shows the corrected images using the navigator echo approach, while the results from the entropy focusing appear in Figure 8. As can be seen in both images, the blurring inside the image has been reduced significantly. The result from entropy focusing is slightly better especially on the sides of the image. This is a direct consequence of the spatially variant motion model used. The estimated angular shifts from both techniques appear in Figure 9. As can be observed, the entropy focusing estimate vary with each image line. The estimate of the navigator echo approach appears as a constant line because it can only work on global shifts. An illustration of the navigator echoes is shown in Figure 10. The method used to estimate the shift is based on measuring the slope of the phase curve of the result of dividing the Fourier transform of both echoes shown in Figure 11. This approach may be of more computational burden but it allows the shift to be estimated in a rather continuous fashion. This is unlike the simple correlation, which has a mandatory step that is equal to the angular spacing between the acquired lines.

Comparing the two techniques, we can see that the computational efficiency of the navigator echo approach is much higher than entropy focusing. In the mean time, the quality of the results is comparable from both techniques. The navigator echo technique using simple correlation can be implemented with minimal extra computational effort for real-time image reconstruction.

6. CONCLUSIONS

Two techniques for motion artifact suppression have been implemented for ultrasound image reconstruction fom multiple acquisitions. The technique based on navigator echo was shown to suitable for real-time reconstruction while providing comparable results to the entropy focusing method. Further work is needed to implement the new techniques for different clinical applications.

ACKNOWLEDGEMENTS

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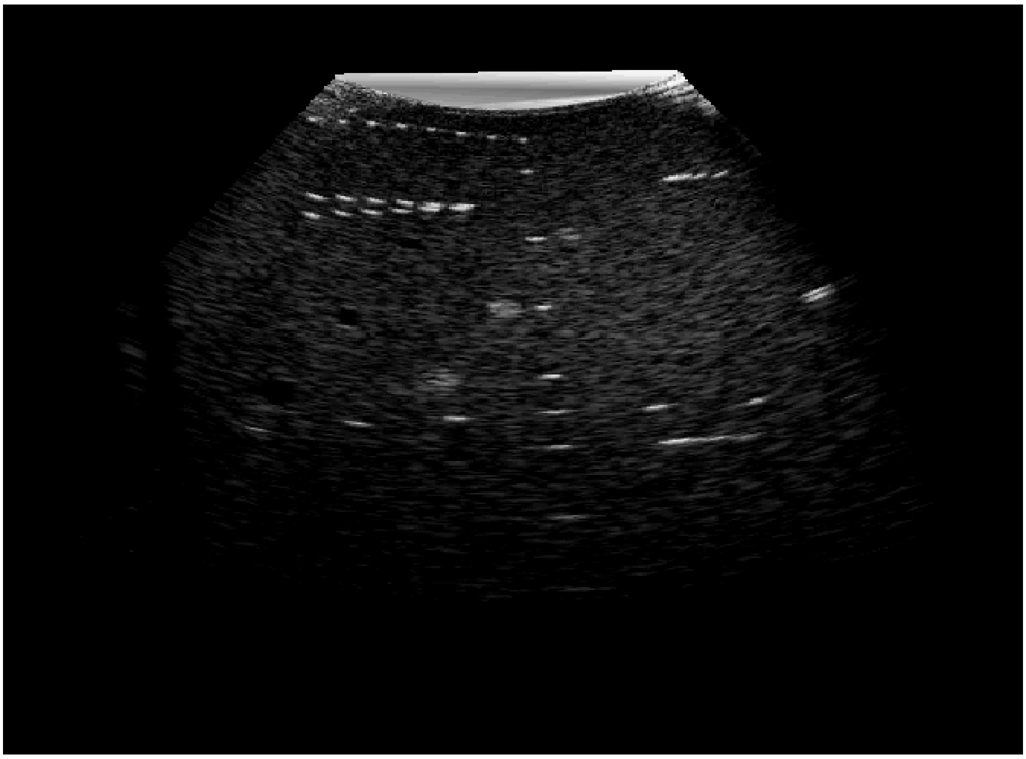

Figure 5: Original phantom image without motion artifacts.

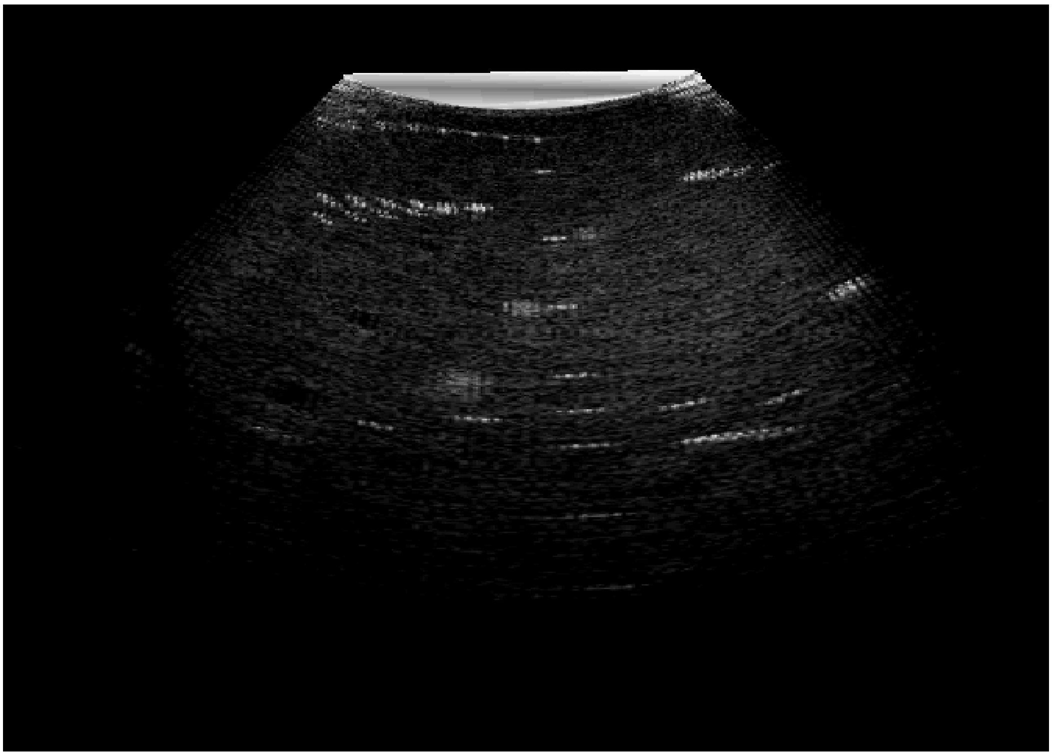

Figure 6: Image with motion artifacts present

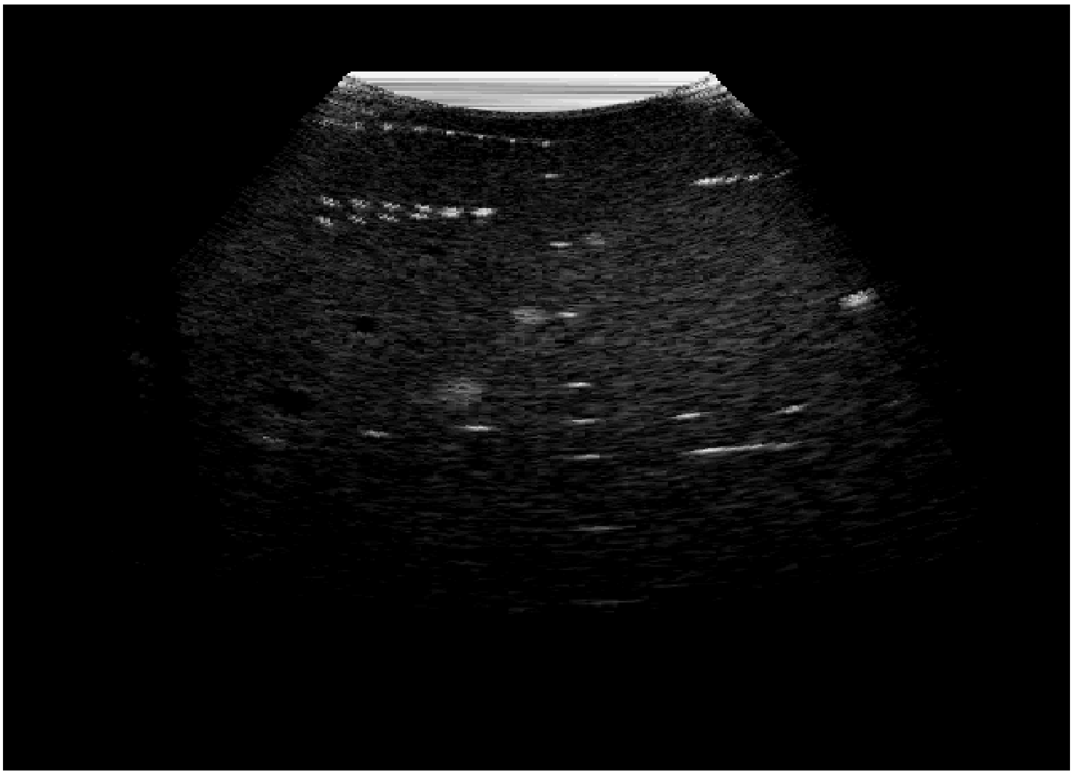

Figure 7: Image corrected with Navigator echo approach.

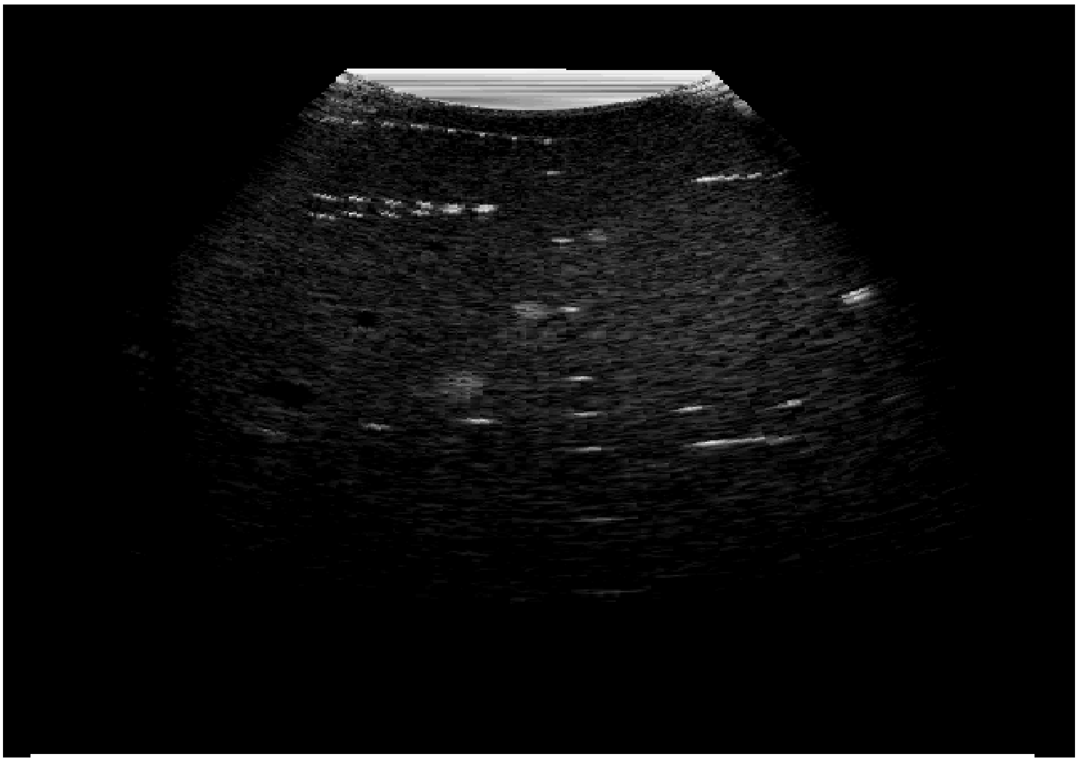

Figure 8: Image corrected using entropy focussing

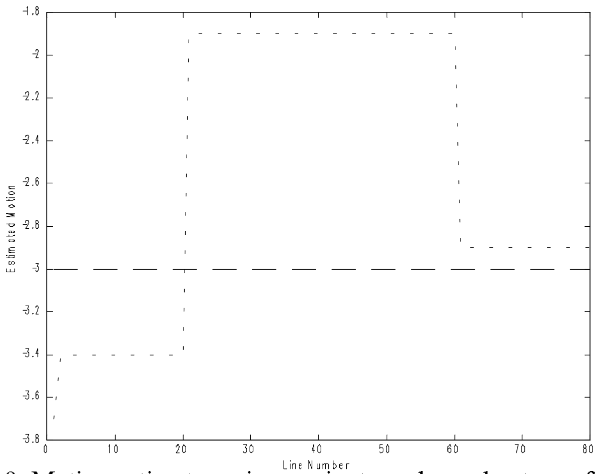

Figure 9: Motion estimates using navigator echo and entropy focussing

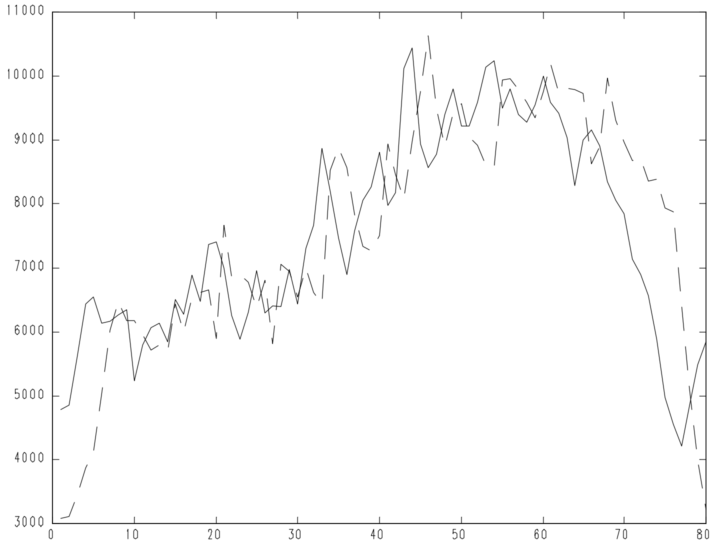

Figure 10: Illustration of navigator echoes from 2 consecutive frames.

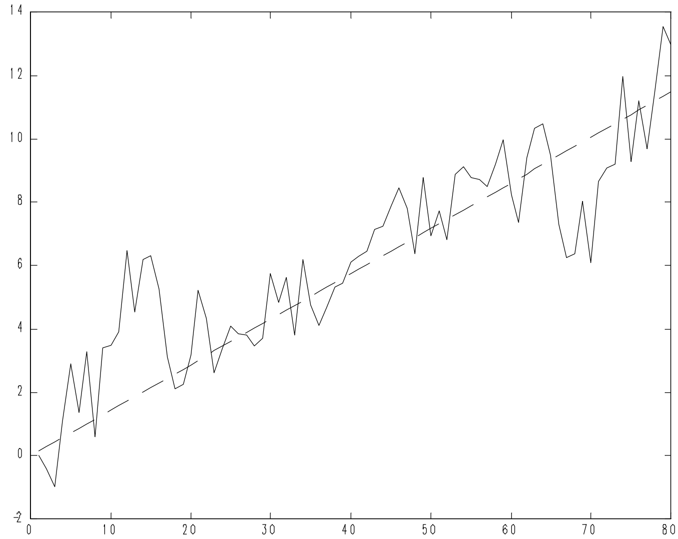

Figure 11: Navigator echo motion estimation from the Fourier-domain linear phase