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Performance of a vector velocity estimator

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Abstract

It is a well-known limitation of all commercially available scanners that only the velocity component along the propagation direction of the emitted pulse is measured, when evaluating blood velocities with ultrasound. Proposals for solving this limitation using several transducers or speckle tracking can be found in the literature, but no method with a satisfactory performance has been found that can be used in a commercial implementation.

A method for estimation of the velocity vector is presented. Here an oscillation transverse to the ultrasound beam is generated, so that a transverse motion yields a change in the received signals. The method uses two ultrasound beams for sampling the in-phase and quadrature component of the lateral field, and a set of samples (in-phase and quadrature in both time and space) are taken for each pulse-echo line. These four samples are then used in an autocorrelation approach that yields both the axial and the lateral velocity, and thus the velocity vector. The method has the advantage that a standard array transducer and a modified digital beamformer, like those used in modern ultrasound scanners, is sufficient to obtain the information needed. The signal processing preceding the beamforming can be implemented using standard signal processors, and it is robust since the autocorrelation method is used.

Measurements are obtained using a point scatterer and a sponge. 250 pulse-echo lines where measured for each object under investigation making it possible to obtain 22 estimates, when using 20 lines with a 50 % overlap in the transverse estimator. The movement of the scatterer was created with a translation stage with a controllable step size. The step size used for the measurement was 40 μm and 250 lines were recorded for angles of 90, 75, 60, and 50 degrees. The results show a bias in the lateral velocity estimate of -18%. The overall standard deviation averaged over all angles was 29.5%.

1 Introduction

Medical ultrasound is extensively used for studying flow dynamics in the human body. This is done using both spectral estimation techniques and color flow mapping. The first finds the velocity distribution over time at one given position in the vessel. The second technique displays a color image of the flow superimposed on the normal anatomic B-mode image. Both techniques measure the velocity component along the ultrasound beam direction, and a flow transverse to the beam is not displayed. This is shown in Fig. 1, where the flow in the carotid artery and the jugular vein is displayed. The image is

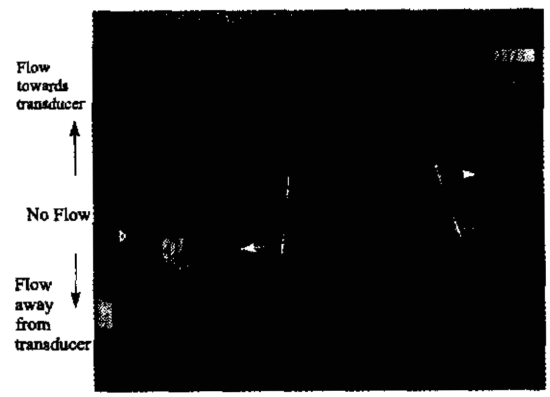

Figure 1: Color flow image of the carotid artery and the jugular vein scanned with a convex array transducer. Notice the change of the angle between the ultrasound beam and the velocity vector around the dashed line.

acquired with a convex array, and the angles between flow direction and the ultrasound beam change over the image. Notice the change of estimated flow direction around the dashed line in both vessels due to the change of angle between the flow and the ultrasound beam. This is one of the main limitations of current ultrasound flow systems, since most vessels are parallel to the skin surface, and therefore it is a problem to get a sufficiently small angle between the flow and the beam. Also the flow is often not parallel to the vessel surface, and it

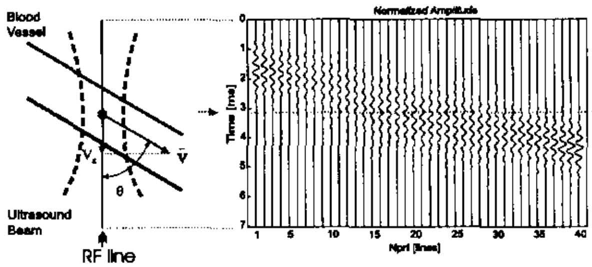

Figure 2: Received RF signals for a pulsed wave system with one scatterer slowly moving past the range gate indicated by the dashed line.

is therefore difficult, if not impossible, to estimate the correct angle and compensate for it [1].

Several authors have attempted to remedy this artifact. Fox [2] suggested using two beams to find the transverse component. The system works well for large transducers and investigations close to the transducer, but the variance of the transverse component increases for situations with large depths and smaller transducers as used in cardiac scanning through the ribs. Trahey and co-workers [3] have suggested using speckle tracking in which a small search region in one image is correlated or compared to a subsequent image. This approach has problems in terms of frame rate, since images are compared, and the resolution of the velocity estimates can be low. Newhouse et al. [4] developed a method in which the total bandwidth of the received signal is affected by the transverse velocity. It is, however, often difficult to find this bandwidth due to the inherent noise in the signal.

In this paper we will present experimental results for the approach described previously [5], which allows the estimation of the flow vector. This new approach introduces a transversely oscillating field through receive beamforming. The oscillation in the transverse direction makes it possible to track the movement orthogonal to the beam direction and thereby find the velocity vector in the two-dimensional scan plane.

2 Traditional velocity estimation

In traditional ultrasound systems for blood velocity estimation a number of consecutive ultrasound pulses are emitted in the same direction in order to track the movement of the blood particles. The pulsed field interacts with the scatterers, and the signal is received by the transducer. The scatterers will have moved a distance proportional to the blood velocity, when the next ultrasound pulse impinges on the scatterers, and the received ultrasound signal will be time shifted compared to the first received response. This is demonstrated in Fig. 2 for a single scatterer. The measurement situation is shown on the left in the figure and the corresponding RF signals are displayed on the right for a number of pulse emissions. It can be seen how the scatterer slowly moves away from the transducer. A signal for detecting this movement can be measured at the horizontal line superimposed on the RF signals. Taking out one sample at a specific depth for each line gives a sampled signal with a frequency proportional to the scatterer's velocity. The time shift of the RF signal from pulse to pulse is

$$t_s = \frac{2v_z}{c} T_{prf}, \tag{1}$$

where v_z is the velocity in direction of the ultrasound beam, c is the speed of sound, and T_{prf} is the time between pulse emissions. A sinusoidal pulse sampled is this way gives a received signal of [6]

$$r(i) = \sin(2\pi \frac{2v_z}{c} f_0 i T_{prf} + \theta), \tag{2}$$

where f_0 is the emitted frequency, i is the pulse-echo number, and θ is a phase factor accounting for the propagation delay. The frequency of the received signal is, thus, proportional to the blood velocity, due to the sinusoidal oscillations in the emitted signal and the sampling of the slow movement of the scatterer past the measurement point. The sampling operation scales the frequency of the emitted signal with a factor of $2v_z/c$, and the spectrum of the received signal is a replica of the emitted spectrum [6], where the frequency axis is scaled by $2v_z/c$.

The sign of the velocity cannot be detected since the spectrum of the received real signal is two-sided. This can be remedied by using a one sided spectrum, which is obtained by either doing a complex demodulation or a Hilbert transform of the received signal to obtain the in-phase and quadrature signals and get a one-sided spectrum. The sign of the received frequency and thereby velocity can then be detected.

3 Transverse velocity estimation

Based on the understanding of the 1D measurement situation, it is now possible to suggest a method for a 2D measurement system. The feature that makes it possible to estimate the axial velocity is the sinusoidal pulsed field. Introducing an oscillation transverse to the beam direction, thus, makes it possible to detect transverse velocities. An oscillation in the lateral direction must, thus, be created and two parallel channels are needed to create a quadrature signal in the lateral direction.

3.1 Acoustic field

We have previously shown [5] through simulations that it is possible to generate two sampling fields, which are displaced laterally to each other. This is done using different delay profiles during reception for the two parallel channels and a sinc

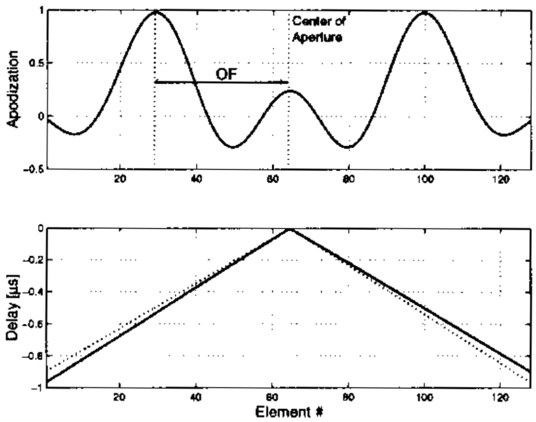

Figure 3: Apodization of array (top graph) and delay profile (bottom graph) for the in-phase (...) and quadrature receive beamformer (—).

apodization across the elements. The apodization and delay values are shown in Fig. 3. The displacement of the acoustic fields is set according to the lateral frequency f_x. The lateral oscillation period ($d_x = 1/f_x$) is given by

$$d_x = \lambda \frac{ZF}{OF}, \qquad (3)$$

where ZF is the depth of interest, OF is defined in Fig. 3, and λ is the axial wavelength. This delay and apodization setup has been used to beamform the measured data.

To validate the beamforming approach, an image of the field at a certain depth was measured. A ruby point reflector was moved horizontally parallel to the transducer surface in steps of 40 μm at a distance of 38.5 mm from transducer surface. 128 elements of a linear array transducer with a center frequency of 6 MHz was used for measuring the response. The emitted beam was focused at a depth of 38.5 mm and a Hanning apodization was applied on the elements. For each position of the point reflector, 128 RF-lines were recorded (one for each active element). The recorded data were beamformed off-line for each position of the point reflector, and the 2-D response from one depth was created. The envelope of the beamformed signal was also calculated using a Hilbert transform, and the two responses, RF and envelope, are shown in Fig. 4.

This result shows that it is possible to generate a laterally oscillating acoustic sensitivity at a controlled lateral frequency. The two fields are displaced one quarter of the lateral oscillation period relative to each other, yielding signals in the two parallel channels which have a quadrature relation. Thereby the spectrum of the transverse signal is one-sided, and the sign of the transverse velocity can be determined.

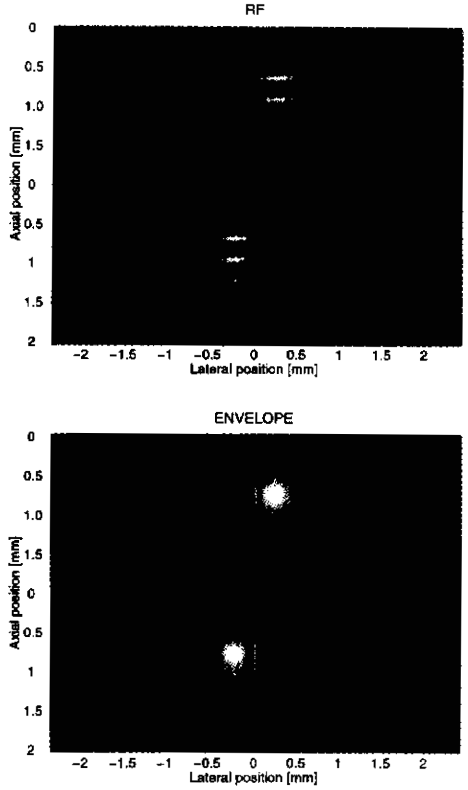

Figure 4: Map of pulse-echo sensitivity measured from a ruby in gelatine. The top graph shows the voltage as a function of position and the bottom graph the envelope of the voltage signal. The zero on the axial position corresponds to a depth of 38.5 mm from the transducer surface.

3.2 Tracking approach

In order to understand the signals from this acoustic system, a simple model is used to interpret the interaction with a reflector. The result of a single scatterer moving through a 2-D oscillating continuous wave (CW) field is shown in Fig. 5. A pure lateral (no axial velocity component) movement results in a oscillation controlled by the lateral oscillation of the acoustic field. A pure axial movement (no lateral velocity component) results in a oscillation controlled by the axial oscillation of the acoustic field. An angulated movement results in an oscillating signal that is a product of both the lateral and the axial oscillating signals. A tracking scheme is used to remove the influence of the axial oscillation on the lateral signal.

The effect of the method is demonstrated in Fig. 5. The signals are generated by a scatterer moving through a CW field with constant frequency and envelope in both the axial and lateral direction. The RF fields are shown on the left side and the signals, when transversing the fields at the black lines, are shown on the right. The top right trace shows the double

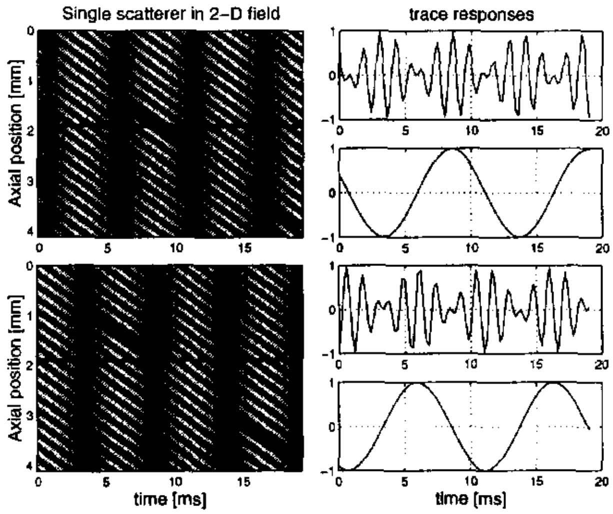

Figure 5: Tracking through RF data to compensate for the axial velocity. The left part of the figure is the RF signals for the in-phase (top) and quadrature (bottom) channels. The right part is the responses of the two channels following the black traces marked in the RF data.

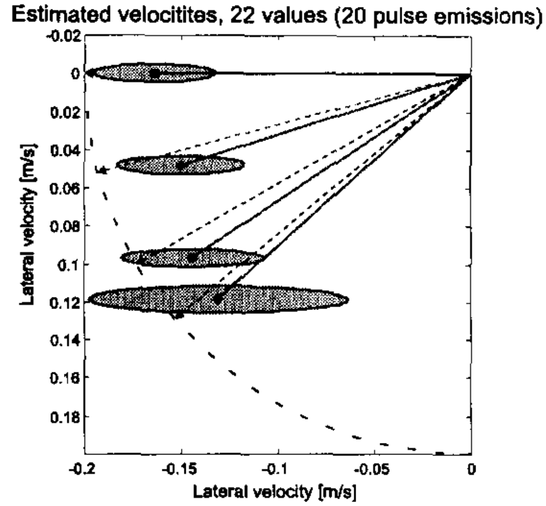

Figure 6: Performance of the new 2D approach using measured data. The ellipses indicate one standard deviation of the estimate and the arrows indicate the correct velocity. The dots in the ellipses are the mean of the estimate.

modulation from both the axial and transverse oscillation and the lower graphs show the lateral signals, when compensating for axial motion. The tracked data ideally only have one frequency component and the autocorrelation approach [7] can, thus, be used to determine the lateral velocity component.

The tracking trace is determined by the estimated axial velocity. Ideally the sampled signal will have the same axial phase, i.e 'iso-phase' tracking. Any error in the estimation of the axial component will affect the estimation of the lateral component. The axial velocity estimate is therefore calculated for both parallel channels and averaged to obtain a better estimate.

4 Performance of approach

The measured data for evaluating the method was generated using a very fine grain sponge to obtained a signal with speckle characteristics. The transducer was angulated and moved relative to the sponge. For each position of the sponge, 128 RF-lines were recorded (one for each active element). The recorded data are beamformed off-line for each position of the sponge reflector. The step size for the measurement was 40 μm and 250 lines were recorded for each angle of interest. The angles used are 90, 75, 60, and 50 degrees.

The results obtained are shown in Fig. 6. Twenty beamformed RF-lines was used for each estimate and using a 50% overlap of the data gave 22 estimates for each direction. The mean value and the standard deviations were calculated and are shown in Fig. 6. The mean value is indicated by the dots and the ellipses show the calculated standard deviations. The true velocities are indicated by the arrows. Biased estimates are obtained due to a misalignment in the generated in-phase and quadrature signals.

The bias on the estimate of the lateral velocity, which was not present in the simulated data, was approximately -18 %. The overall standard deviation, averaged over all angles, is 29.5 %. Individual results are listed in Table 1

5 Summary

An approach for estimating the 2-D velocity vector has been presented. The performance is slightly degraded compared to the results obtained by the simulations due to the use of fewer pulse-echo lines and a more focused fields. A bias in the estimate is seen, but the standard deviation is comparable to the results obtained in previous simulations. The velocity vector can thus be estimated for flow parallel to the transducer and an improved display of blood velocity can be obtained.

6 Acknowledgement

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θ	90	75	60	50
v_z	-0.003 (0.000)	0.048 (0.052)	0.097 (0.100)	0.118 (0.129)
v_x	-0.164 (-0.200) -18 %	-0.150 (-0193) -22 %	-0.144 (-0.173) -17%	-0.131 (-0.153) -14%

θ	90	75	60	50
σ_z	0.0048	0.0047	0.0046	0.0071
σ_z/v_z	-	0.098	0.047	0.060
σ_x	0.0317	0.0330	0.0371	0.0672
σ_x/v_x	-0.19	-0.22	-0.26	-0.51

Table 1: Results for the performance of the new 2D approach using measured data. The top table shows the estimated mean axial (v_z) and transverse (v_x) velocities. The values in parenthesis show the true values and below is given the bias in percent. The bottom table shows the corresponding standard deviations.

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