Vector blood velocity estimation in medical ultrasound

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Two methods for making vector velocity estimation in medical ultrasound are presented. All of the techniques can find both the axial and transverse velocity in the image and can be used for displaying both the correct velocity magnitude and direction. The first method uses a transverse oscillation in the ultrasound field to find the transverse velocity. In-vivo examples from the carotid artery are shown, where complex turbulent flow is found in certain parts of the cardiac cycle. The second approach uses directional beam forming along the flow direction to estimate the velocity magnitude. Using a correlation search can also yield the direction, and the full velocity vector is thereby found. An examples from a flow rig is shown.

Keywords: Medical ultrasound, vector velocity estimation, directional beam forming.

1 INTRODUCTION TO VECTOR VELOCITY ESTIMATION

Imaging of the blood velocity has been made for the last 20 years. The original paper by Kasai et al. [1] demonstrated that the velocity could be estimated in the direction of the ultrasound beam. This often gives problems, since the vessel run along the skin surface and the ultrasound beam is, thus, perpendicular to the flow direction. This can be corrected by tilting the beam, but often the geometry of the vessels is complicated and a single correction is not sufficient. There is therefore a real need in medical ultrasound for vector flow techniques. This paper will present two techniques for vector flow imaging developed in our laboratory.

Many authors have developed techniques for vector velocity estimation. Fox [1,2] developed the crossing beams technique, where the velocity is found along two beam directions. The approach is, however, difficult to use for imaging, as it is difficult to get crossing beams simultaneously over the full image. Another approach is speckle tracking [3], where two images acquired at different times are correlated to find the vector motion. A number of other techniques have also been introduced, but none have so far made it into commercial products.

2 TRADITIONAL VELOCITY ESTIMATION

Traditional velocity estimation is performed by emitting a pulsed field a number of times in the same direction as shown in Fig. 1. Here the left graph shows the received signal from a single scatterer moving away from the transducer. The consecutive signals are shown with an increasing *y*-axis offset. The signal is sampled at the depth of interest resulting in the signal on the right. The frequency of this signal is proportional to the velocity as [4]

$$f_p = \frac{2\,|v|\cos\Theta}{c} f_0 = \frac{2v_z}{c} f_0 \tag{1}$$

where Θ is the flow angle between the ultrasound beam and the velocity vector, $|v|$ is the velocity magnitude, v_z is the axial blood velocity, c is the speed of sound, and f_0 is the emitted ultrasound center frequency. A larger velocity will, thus, give a larger shift between signals and a higher frequency. The generating mechanism is the oscillation in the ultrasound signal and the shift in position between pulse emissions.

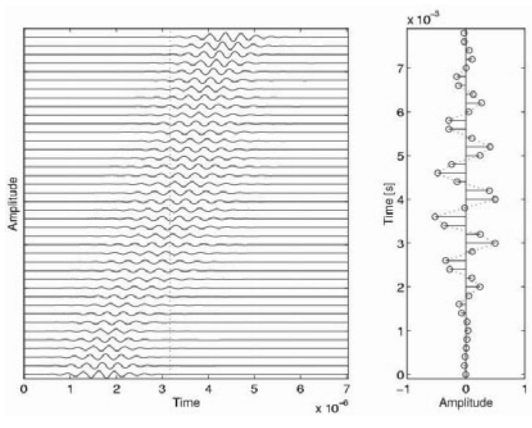

Figure 1: Received RF signal for a single scatterer moving away from the transducer. The left graph shows the individual traces received after each pulse emission and the right graph shows the sampled signal obtained at the dashed line in the left graph at 3.2 μs (from [4]).

Figure 2 shows the major problem with this approach. The velocity has here been determined at a number of places in the image from the frequency given by (1). A blue color shows velocity towards the transducer and a red color away from the transducer. The jugular vein and carotid artery in the neck has been scanned. A change in color is seen around the green line. This is due to the cos Θ factor in (1) as the flow angle changes along the green line. This is easily detected in this image, but very difficult to compensate for in more complicated flow patterns. A major problem is also that the projected velocity is shown. This will always make the velocity lower than the true velocity, and this makes quantitative velocity imaging difficult.

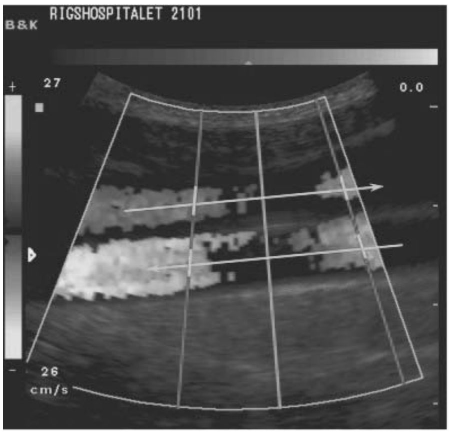

Figure 2: Velocity image of the jugular vein (top) and carotid artery (bottom). The blue color indicates velocity towards the transducer and red away from it. The change in color along the blue center line is due to the change in angle between the ultrasound beam and the velocity direction (from [16]).

3 TRANSVERSE OSCILLATION APPROACH

The important feature for estimating the velocity in traditional systems is the sinusoidal pulse in the axial directions. This generates the received frequency and makes it possible to determine the velocity. The transverse velocity cannot be estimated, since there is no corresponding transverse oscillation, as shown in Fig. 3. Here the point spread function (psf) is depicted with a black value corresponding to a negative pressure and a white value is positive pressure. A scatterer going through the psf in the axial direction will give an oscillation as described in the previous Section with a frequency given by (1). A corresponding lateral movement will give the signal shown in the right graph in Fig. 3. Here no oscillation is seen and the mean frequency of this signal will be zero.

The basic idea in the transverse oscillation approach [5, 13] is to introduce a lateral oscillation in the field as shown in Fig. 4. A lateral motion will then give the signal shown in the right graph in Fig. 4, with an oscillation frequency proportional to the lateral velocity as

$$f_l = \frac{v_x}{\lambda_x} \qquad (2)$$

where v_x is the lateral oscillation and λ_x is the lateral oscillation period in the signal. Such a transverse oscillation can be introduced by splitting the transducer aperture in two and emitting waves from each part as shown in Fig. 5. The two fields will then interfere and generate a transverse oscillation.

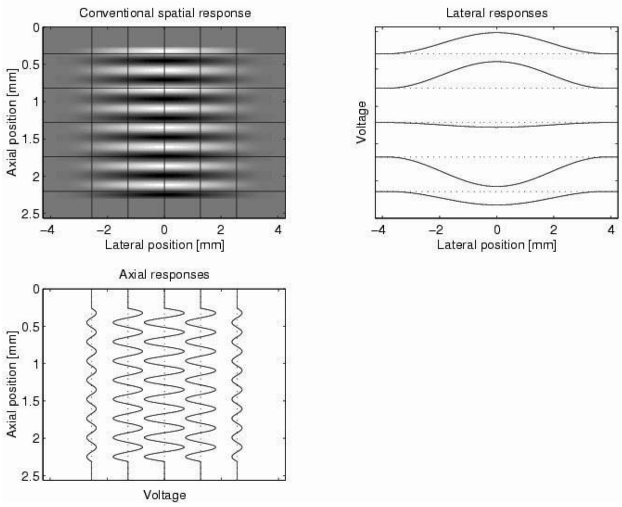

Figure 3: Point spread function (psf) for ordinary ultrasound velocity estimation. The top left graph shows the psf and the lower graph shows the response from a scatterer transversing the field in the axial direction. The right graph shows the signal for the scatterer transversing the field in the lateral direction (from [5]).

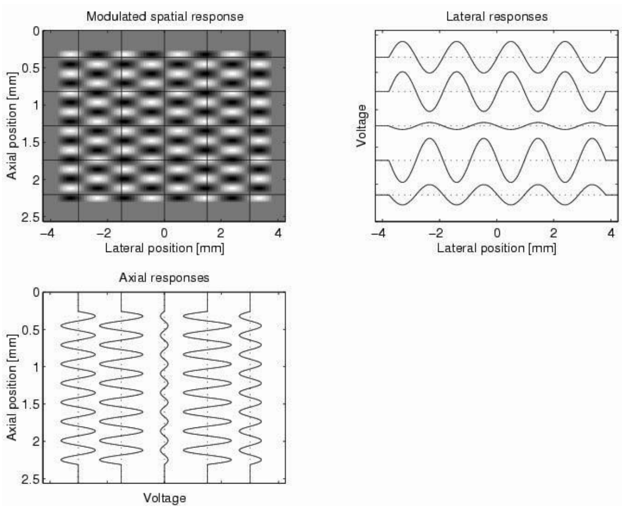

Figure 4: Point spread function (psf) for transvers oscillation method. The top left graph shows the psf and the lower graph shows the response from a scatterer transversing the field in the axial direction. The right graph shows the signal for the scatterer transversing the field in the lateral direction (from [5]).

The received signal can then be processed to yield both the axial and lateral velocities independently as described in [6].

An example of such an image is shown in Fig. 6, which has been measured on the carotid artery of a human volunteer. A 128 element, 7 MHz linear array probe was used connected to the RASMUS experimental ultrasound scanner [14]. The data was stored in the scanner and then processed in a Linux cluster to estimate both the axial and lateral velocities. The image shows the velocity magnitude as a red color with the magnitude corresponding to color intensity. The arrows on the image indicate both direction and magnitude. The flow angle is nearly 90 degrees and an estimate of the velocity is obtained. For this simple case the ultrasound beam could just be titled to give an estimate with the conventional method, but this is not possible for more complex patterns.

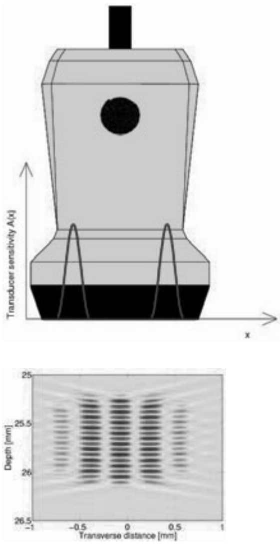

Figure 5: Transducer aperture used for transmission and the corresponding field as simulated by the Field II program [12].

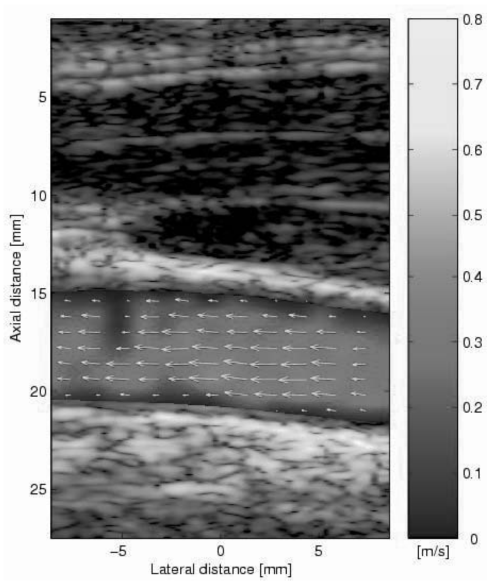

Figure 6: Vector velocity image of the carotid artery. The color indicates velocity magnitude and arrows indicate both velocity direction and magnitude (from [7]).

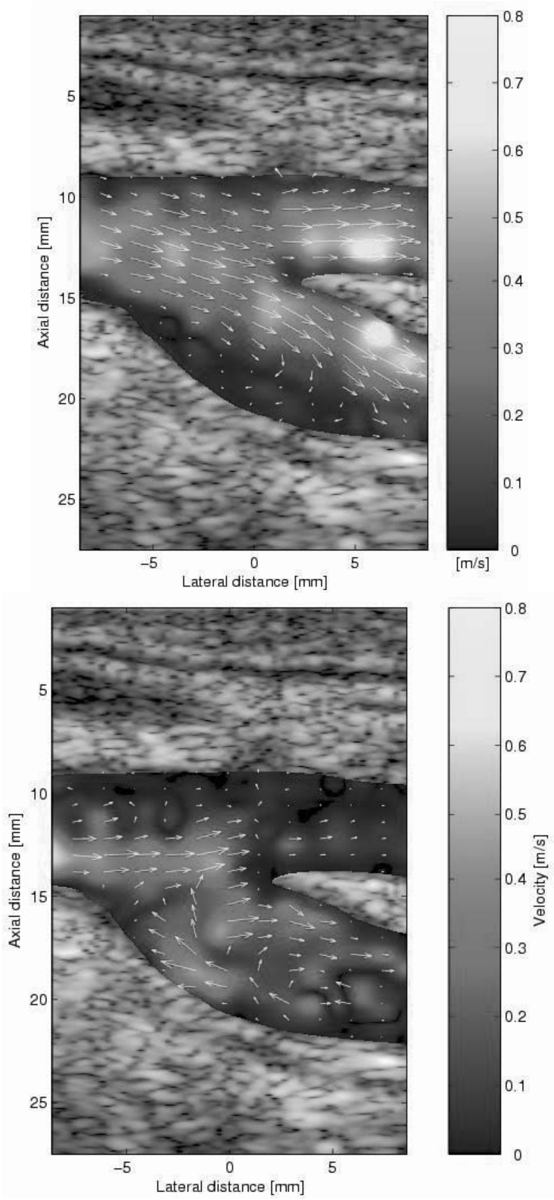

Figure 7: Vector velocity image of the carotid artery bifurcation at peak systole (top) and 0.1 s after peak systole. (from [7]).

This is demonstrated in Fig. 7, where the flow patterns in the carotid bifurcation are shown around the maximum contraction of the heart. The flow at peak systole is fairly laminar with a lower velocity at the lowest part of the vessel, whereas there is a very disturbed, rotating flow right after peak systole. The velocity arrows here rotate full circle within 100-200 ms demonstrating the importance of showing the full velocity vector to capture and display the very complex flow patterns in the human body.

4 DIRECTIONAL BEAMFORMING APPROACH

There are also other methods for finding the velocity vector. The directional beam forming approach uses focusing along the flow direction to find the velocity vector [8,9]. Here a normal focused ultrasound field is emitted and the scattered signal is sampled from all the receiving elements. The direction of the flow is determined from the B-mode image and signals are then focused along this direction as shown in Fig. 8. The process is repeated and a number of signals are found and the signals are cross-correlated to yield the shift in position between the signals. The spatial shift will be equal to the velocity magnitude times the time between the signals. Finding the shift and dividing by the time will then yield the velocity. This is the principle in the directional beam forming method.

An example of such an image is shown in Fig. 9. It was made using a circulating flow rig with a mass flow meter as reference. A 128 element, 7 MHz linear array probe was used connected to the RASMUS experimental ultrasound scanner and 20 pulse echo lines were acquired for each of the 65 imaging lines. The velocity was then found by focusing lines perpendicular to the ultrasound beam, cross-correlating them, and then finding the velocity.

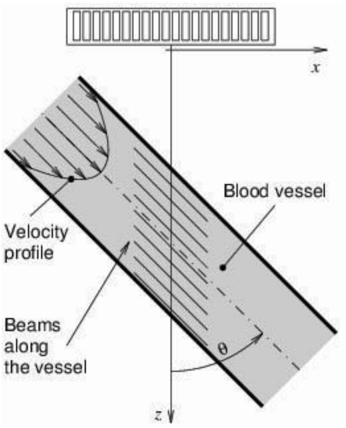

Figure 8: Directional focusing along the flow direction (from [10]) .

The standard deviation across the parabolic profiles is 4.3% relative to the peak velocity in the vessel, which is comparable to the accuracy obtained in traditional scanners, that here would show no velocity at 90 degrees.

The angle for the flow can also be determined for this approach. This is done by beam forming in a number of directions and then find the velocity and normalized peak cross-correlation for the different angles. The highest correlation coefficient then denotes the correct angle, since the signals will de-correlate for others angles as the velocity is different in different parts of the directional signal. Such an approach has been investigated in [15].

5 SUMMARY

Two different methods for vector velocity estimation have been presented. It has been shown that conventional techniques can lead to cross errors in determining the correct velocity and that vector velocity techniques are needed in order to display the full complexity of flow in the human body.

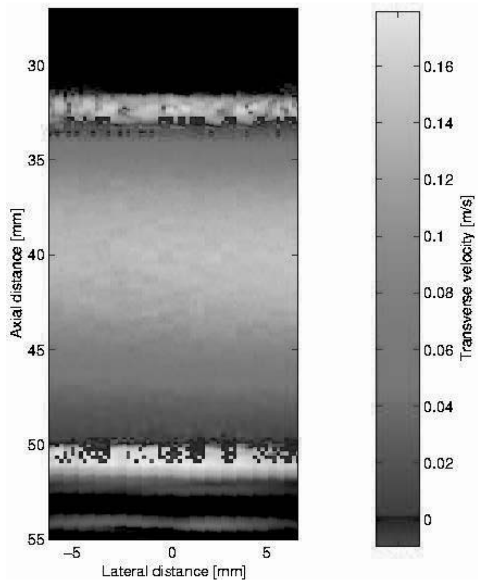

Figure 9: Velocity image at 90 degrees in a flow rig (from [9]).

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