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Transverse flow imaging using synthetic aperture directional beamforming

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Abstract - Current ultrasound scanners only determine the velocity along the ultrasound beam, since data is only focused along the emitted beam. Synthetic aperture ultrasound systems have the capability of focusing simultaneously in all directions. This is used here to focus along the flow direction and then cross-correlate these measurements to obtain the correct velocity magnitude.

The approach was investigated in a flow system with a laminar flow. The flow profile was measured with a B-K Medical 8804 7.5 MHz linear array transducer. A plastic tube with an entrance length of 1 m and a diameter of 17 mm was used with an EcoWatt 1 pump generating a laminar, stationary flow. The velocity profile was measured for flow angles of 90 and 60 degrees. The RASMUS research scanner was used for acquiring RF data from 128 elements of the array using 8 emissions with 11 elements in each emission. A 20μs chirp was used during emission. The RF data were subsequently beamformed off-line and stationary echo canceling was performed.

The 60 degrees flow was determined using 16 groups of 8 emissions and the relative standard deviation was 0.36 % (0.65 mm/s). Using the same setup for the purely transverse flow gave a std. of 1.2 % (2.1 mm/s).

An *in-vivo* image of the carotid artery and jugular vein of a healthy 29 years old volunteer. A full color flow image using only 128 emissions could be made with a high velocity precision.

I INTRODUCTION

Current ultrasound systems measure the blood flow by emitting pulses in one direction a number of times. The velocity is then found by segmenting the signals and correlating these segments to find either the phase shift or displacement and thereby the velocity. This technique has several problems: Foremost the frame rate is directly dependent on the number of directions the flow is found in. Normally 8 to 16 emissions have to be made per direction and 50 to 100 directions have to be made. For a pulse repetition frequency of 5 kHz, the frame rate is then 3 images/sec in the worst case. This is too low to adequately sample the pulsatility of the flow. The second problem is that only few lines are used in the estimation. This makes the standard deviation quite high, and it is often 10 to 20 %. The third problem is that only the velocity component projected onto the ultrasound beam direction is found. No velocity or a disturbed velocity can often only be found, since many vessel run along the skin surface.

The coupling between the frame rate and image size can be solved by using the aperture flow method suggested in [1], where continuous data can be acquired simultaneously in all directions of the image. This also makes it possible to average over more measurements to improved the standard deviation of the estimates. The third problem can be solved by using directional focusing along the flow direction as suggested in [4, 2]. Here the received ultrasound signal is focused along the direction of the flow, and these signals are correlated and the displacement found. From this the correct velocity magnitude can be found, and a flow transverse to the ultrasound beam can be detected. These two approaches are combined in this paper to make a color flow system that in 128 emissions can yield a full image of a transverse flow with an accuracy of 1.2 %. The theory of the approach is described in Section II, which details how image acquisition and focusing is done. The approach is investigated using an experimental ultrasound scanner and a flow rig in Section IV, and full color flow images are shown in Section V along with an *in-vivo* example in Section VI.

II DIRECTIONAL SYNTHETIC APERTURE FLOW ESTIMATION

A synthetic aperture image is made by emitting spherical waves from the ultrasound transducer and then receive the signal on the individual elements. Since the position of emission is known, the precise time from emission to reception can be calculated and used in the focusing of signals. The distance from emission to reception is given by:

$$d(\vec{r}_p) = |\vec{r}_p - \vec{r}_e| + |\vec{r}_p - \vec{r}_r| \qquad (1)$$
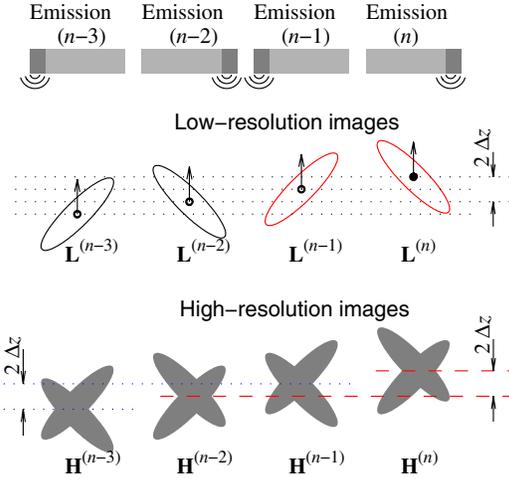
Figure 1: Beamformation of high resolution images from the low resolution images using two emissions (from [1]).

Figure 2: Creation of the cross-correlation function and its averaging over several emissions (from [1]).

where $\vec{r}_p$ is the point in the image, $\vec{r}_e$ is the location of the transmission, and $\vec{r}_r$ is the position of the receiving element. Focusing for a single transmission is then done by adding the received signals $g(t, \vec{r}_r(i))$ from all elements as

$$y(\vec{r}_p) = \sum_{i=1}^{N_e} g\left(\frac{d(\vec{r}_p)}{c}, \vec{r}_r(i)\right) \quad (2)$$

where c is the speed of sound, i is the element number, and N_e is the number of elements. To make a fully focused image, the beamformed signal from each of the individual emissions are summed to give a high resolution image. The advantage of this approach is that the image is fully focused in both transmit and receive at all positions in the image.

Since the image is constructed over a number of emissions, the signal from a moving object will not be at the same position for the individual emissions. This has been used as an argument against the use of synthetic aperture for flow estimation, which is not correct. For velocity estimation it is only necessary to compare two measurements, and then find the positional shift between the measurements. The only demand is that the two measurements are obtained in exactly the same way, so the difference is only the movement between measurements. This was used in [1] to devise a SA method, that can be used for flow estimation.

The approach is illustrated in Fig. 1. The high resolution images are here created from two emissions and the individual low resolution images are seen in the middle row. High resolution images can be made by combining data from emission $[(n-3), (n-2)]$, $[(n-2), (n-1)]$ and $[(n-1), (n)]$. It can be seen that the distortion from using $[(n-3), (n-2)]$ and $[(n-1), (n)]$ is exactly the same, and when correlated the only change will be the motion of the scatterers. This motion can then be found by cross-correlating signals from the two high
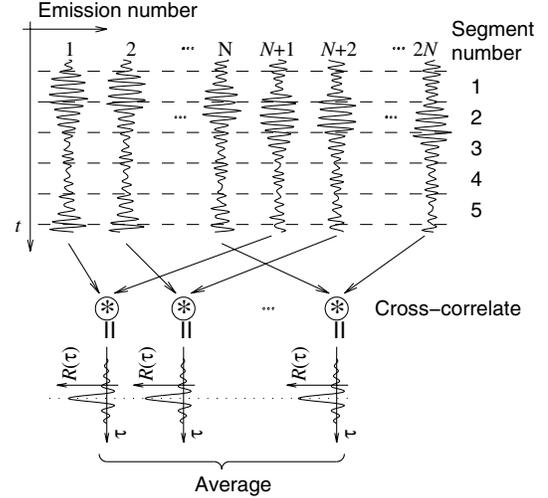
resolution images and then find the shift in position. Dividing by the time between the two images gives the velocity. It should be noted that the image created from $[(n-2), (n-1)]$ can be used and correlated with $[(n), (n+1)]$, so that a new usable high resolution image is created after every pulse emission. These images can be created using recursive ultrasound imaging [3].

The cross-correlation function can be averaged over all the high-resolution pairs of images, since the shift between the pairs is the same. This is shown in Fig. 2. The cross-correlation is performed between high resolution data N emissions apart, where N is the number of emissions for creating one high resolution image.

It should be noted that data for the whole image region is continuously available for all places in the image, which makes stationary echo canceling easier, and the velocity estimates can be averaged over many more emissions than in a traditional system.

III DIRECTIONAL VELOCITY ESTIMATION

In a traditional ultrasound scanner the received signal is focused along the direction of the emitted beam. This is not necessary in a SA system, since data is available for all directions of the image simultaneously. The beams can therefore be formed in any order and direction, and it is possible to beamform signals along the direction of the flow as suggested in [4] and [2]. The receive focus points are placed along lines parallel to the flow direction. These lines are denoted receive focus lines.

The receive focus lines obtained at subsequent time instants (with intervals of T_{prf}) are cross-correlated and the peak of the correlation signal is found. The position of the

peak is at

$$d_s = |\vec{v}| T_{prf}, \tag{3}$$

where $|\vec{v}|$ is the velocity magnitude and T_{prf} is the time between the directional signals. Cross-correlating the signals and finding the position of the maximum in the function then yields the velocity as:

$$\hat{v} = \frac{\hat{d}_s}{T_{prf}} = \hat{n}_s \frac{dx}{T_{prf}}, \tag{4}$$

where dx is the distance between receive focus points in a receive focus line and $\hat{n}_s$ is the lag.

IV FLOW PHANTOM MEASUREMENTS

An experimental flow rig has been used for revealing the performance of the approach. A re-circulation flow rig consisting of a pump (Smedegaard EcoWatt 1), reduction valve, and a tube was used for generating the flow. The flow entered a heath shrinking tube, with a diameter of 17 mm after going through 1 m of 20 mm diameter steel tube to ensure a non-turbulent parabolic velocity profile. The heath shrinking tube was submerged in water and a transducer fixture ensured a precise alignment of the transducer with respect to angle and distance to the tube. A 7 MHz B-K Medical 8804 linear array transducer was used in the experiment along with the RAS-MUS experimental ultrasound scanner [5]. The scanner can emit arbitrary signals in 128 individual channels and it can simultaneously measure 64 receive channels at 40 MHz and 12 bits precision. A 2-to-1 multiplexing makes it possible to cover all 128 transducer elements in 2 emissions. The data is stored in the systems 16 GBytes of memory in real-time, and the data is the transferred to a Linux computer cluster with 32 CPUs for later processing.

The measurements have been performed by using 8 emissions equally spread over the 128 element aperture, with each emission using 11 defocused elements to emulate a spherical wave emission [6]. A frequency encoded chirp is used in transmission to increase the signal-to-noise ratio [7]. The chirp has a duration of 20 μs and a bandwidth of 7 MHz centered around the transducer center frequency. The receiving elements are multiplexed to be closets to the emitting center element and all 128 receiving elements are hereby sampled. The transmission is then repeated with $f_{prf} = 3$ kHz and 3,000 individual emissions have been acquired by repeating the sequence 375 times.

The received RF signals are first matched filtered to focus the signals in the axial direction and increase the signal-to-noise ratio. Then the low resolution signals are focused in direction of the flow. The high resolution images are made by adding the latest 8 low resolution images. Stationary echo canceling is done by finding the mean value of the focused lines and then subtract this from the signals. The focused directional lines from the same image formation are
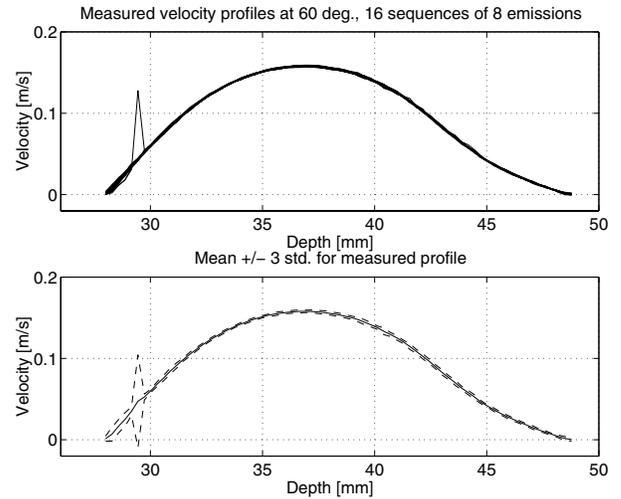

Figure 3: Estimated profiles from the flow rig at a 60 degree flow angle. The top graph shows the 20 independent profiles estimated and the bottom graph shows the mean profile (solid line) $\pm$ 3 standard deviations (dashed lines).

subsequently cross-correlated and added to the other cross-correlation functions, and the velocity is found from the combined cross-correlation function.

Two experiments at flow angles of 60 degrees and 90 degrees (transverse velocity) has been performed. The velocity was found from 16 sequences of 8 emissions, each corresponding to a total of 128 emissions. This is the same number of emissions that is used in normal spectral velocity imaging [8] over which the flow normally can be considered quasi-stationary in the human body.

The resulting velocity profiles are shown in Fig. 3 for a flow angle of 60 degrees. The 20 individual profiles are shown on the top and the mean of the profiles $\pm$ 3 standard deviations are shown on the bottom. It can be seen that the tube is not exactly round and a single false velocity peak is detected [8]. The relative standard deviation averaged over the profile is 0.36 % (0.65 mm/s) compared to the peak velocity in the vessel, when the single false peak is neglected.

The same information for 90 degrees (transverse flow) is shown in Fig. 4. Here the relative standard deviation is 2.8 % (3.6 mm/s) compared to the peak velocity in the vessel. The accuracy is lower, but it should be remembered that traditional scanners cannot detect a transverse flow. Larger errors are seen at the edge of the vessel, and this is due to the low signal after stationary echo canceling here. Taking only the standard deviation beyond the edges gives a value of 1.2 % (2.1 mm/s).
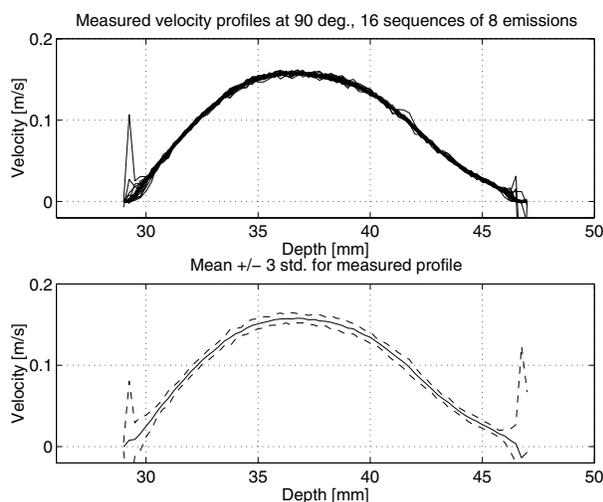
Figure 4: Estimated profiles from the flow rig at a 90 degree flow angle. The top graph shows the 20 independent profiles estimated and the bottom graph shows the mean profile (solid line) $\pm$ 3 standard deviations (dashed lines).

V   LINEAR ARRAY COLOR FLOW IMAGES

The profiles displayed in the previous section were made using 128 emissions, which is considerably more than the traditional 8 to 16 emissions. The difference is, however, that a full color flow image can be made from the same data, since the received signal can be focused at all places in the image. An example of this for the 60 degrees flow data is shown in Fig. 5. The color scale indicates the velocity along the flow direction, where red hues indicate forward flow and blue reverse flow. The intensity of the color indicates the velocity magnitude.

The received data has been beamformed to make what corresponds to a linear array scan. The profiles are found with the same accuracy as the one for Fig. 3. Estimates outside the vessel are set to zero by considering the signal energy after the stationary echo canceling compared to the noise level in the measurement system. If the signal is below the noise level, the velocity estimate is set to zero. It should also be noted that no image processing has been employed on the displayed image and only the actual estimates are shown.

The second example for the image for the 90 degrees flow data is shown in Fig. 6. The full image is again made from only 128 emissions, and it can be seen that the fully transverse flow can be accurately detected. A ghost flow image is seen beyond 50 mm. This is due to re-reflections between the vessel wall and the transducer. It can possible be eliminated by a more advanced tissue-flow discriminator.
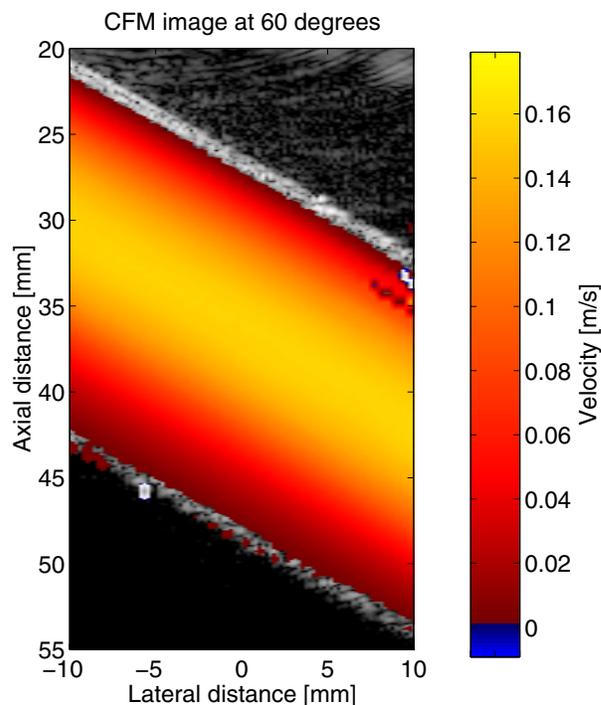

Figure 5: Synthetic aperture color flow map image of flow rig data at a 60 degree flow angle obtained using 128 emissions.
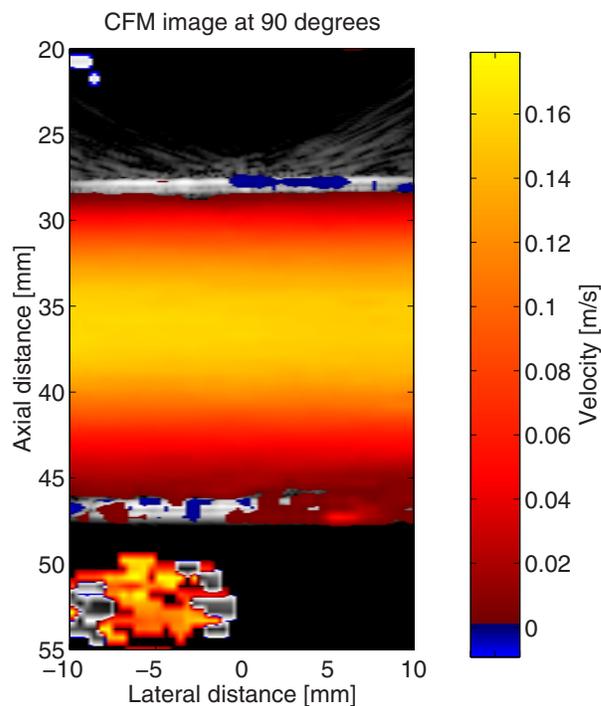

Figure 6: Synthetic aperture color flow map image of flow rig data at a 90 degree flow angle obtained using 128 emissions.
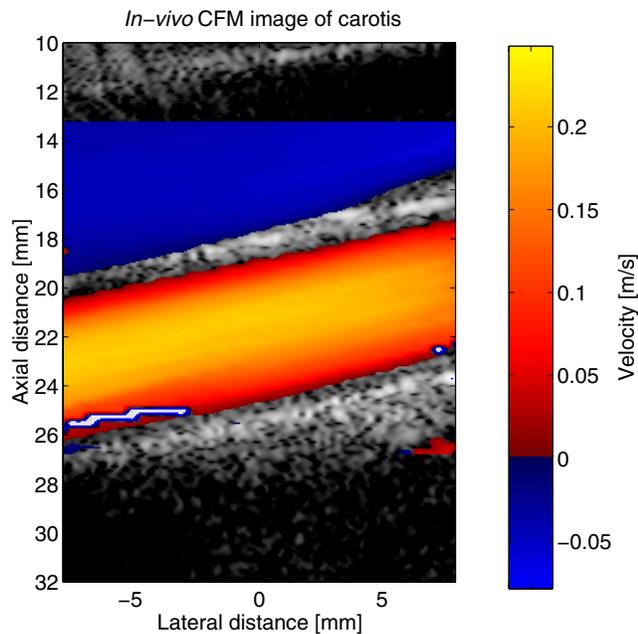
In–vivo CFM image of carotis

Figure 7: *In-vivo* color flow map image at a 77 degree flow angle for the jugular vein and carotid artery. The color scale indicates the velocity along the flow direction, where red hues indicate forward flow and blue reverse flow.

VI   *In-vivo* MEASUREMENTS

The setup as used for the flow rig images have been used to make the first *in-vivo* image of the jugular vein and the carotid artery. The result is shown in Fig. 7, where the color scale indicates the velocity along the flow direction. A red hue indicate forward flow and blue reverse flow. The 20μs chirp employed has a length corresponding to 15 mm, and the velocity can therefore not be shown above 15 mm. The pulse repetion frequency was 3 kHz and 128 emissions (16 groups of 8 emissions) was used for finding the velocity. The discrimination between showing the color flow image and the B-mode image was done solely on the magnitude of the velocity. It can be seen that the flow follows the vessel boundaries, and that the velocity vectors have opposite directions in the two vessels. A larger velocity is seen at the center of the vessels than at the edges, which is consistent with what would be expected. A few wrong peaks can be seen at the lower edge of the carotid artery, where the signal-to-noise ratio is low, due to the stationary echo canceling.

The image is measured before the systolic phase of the cardiac cycle. At systoli there is a rapid acceleration of the flow and the method encounters problems, since the averaging of the cross-correlation functions are not valid here. We are currently investigating how to solve this problem. It should be noted that f_{prf} is quite low for this measurement and that it for this depth can be increased to nearly 19 kHz, which would give a much higher correlation between the individual measurements. This would possible solve the problem.

VII   CONCLUSION

A method for making synthetic aperture flow imaging using directional focusing has been presented. The method can find the correct velocity magnitude of the flow and it can using few emissions generate a full color flow image with a standard deviation on the order of 1 %. It can also estimate flow transverse to the ultrasound beam.

ACKNOWLEDGEMENT

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