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Multi-frequency encoding for rapid color flow and quadroplex imaging

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Abstract— **Ultrasonic color flow maps are made by estimating the velocities line by line over the region of interest. For each velocity estimate, multiple repetitions are needed. This sets a limit on the frame rate, which becomes increasingly severe when imaging deeper lying structures or when simultaneously acquiring spectrogram data for triplex imaging.**

This paper proposes a method for decreasing the data acquisition time by simultaneously sampling multiple lines at different spatial positions for the color flow map using narrow band signals with disjoint spectral support. The signals are separated in the receiver by filters matched to the emitted waveforms and the autocorrelation estimator is applied. Alternatively, one spectral band can be used for creating a color flow map, while data for a number of spectrograms are acquired simultaneously. Using three disjoint spectral bands, this will result in a multi-frequency quadroplex imaging mode featuring a color flow map and two spectrograms at the same frame rate as a normal color flow map. The method is presented, various side-effects are considered, and the method is tested on data from a re-circulating flow phantom where a constant parabolic flow with a peak of 0.1 m/s is generated with a flow angle of 60 degrees. A commercial linear array transducer is used and data are sampled using our RASMUS multi-channel sampling system. An *in-vivo* multi-frequency quadroplex movie of the common carotid artery of a healthy male volunteer was created.

The flow phantom measurements gave a mean standard deviation across the flow profile of 3.1%, 2.5%, and 2.1% of the peak velocity for bands at 5 MHz, 7 MHz, and 9 MHz, respectively. The *in-vivo* multi-frequency quadroplex movie showed the color flow map, and the two independent spectrograms at different spatial positions. This enables studying the flow over an arterial stenosis by simultaneously acquiring spectrograms on both sides of the stenosis, while maintaining the color flow map. A frame rate of 21.4 frames per second was achieved in this *in-vivo* experiment.

I. INTRODUCTION

Since the introduction of the autocorrelation estimator for ultrasonic blood flow imaging by Kasai et. al. [1] in the mid-eighties, it has been possible to create full color flow maps (CFM) in real time. The method is robust and fairly simple to implement. The velocities are estimated line by line over the region of interest, and for each line, repeated transmissions are needed. The performance of the estimator is closely linked to the number of repetitions used to form the estimate [2]. An inherent trade-off between frame rate and performance therefore exist. When imaging deep structures like the heart, this might severely affect either the performance or the frame rate. Also, in the triplex imaging mode, where both CFM data and a spectrogram are shown, the time for data acquisition must be divided between multiple sets of data, increasing the significance of this problem even further.

The autocorrelation estimator essentially assumes a narrow-band signal, and the variance of the velocity estimates decreases as the bandwidth is decreased [2]. Therefore a long sinusoidal narrow-band pulse is used for CFM. Nevertheless, ultrasound transducers are often designed fairly wide-band to ensure a good resolution in B-mode images. Using only a narrow frequency band for collecting CFM data does not exploit the available bandwidth efficiently.

This paper proposes a method for significantly decreasing the time spent on data collection for CFM by simultaneously sampling multiple lines using different frequency bands. The signals are then separated in the receiver by a simple filtering operation and the autocorrelation estimator is applied. The number of frequency bands M depends on the available transducer bandwidth and the intensity limits set by the Food and Drug Administration [3]. The total time spent on CFM data acquisition will decrease by a factor of M, potentially increasing the frame rate with a factor of M. Alternatively, the proposed method can be used for simultaneously acquiring CFM data and spectrogram data for triplex imaging. This paper will demonstrate how data for an extension of triplex imaging, namely Multi-Frequency Quadroplex imaging (MFQ) featuring two independent spectrograms and a CFM, can be acquired during the time normally spent on acquiring a CFM.

Parallel receive beamforming has previously been proposed for decreasing the time spent on data acquisition for B-mode images [4] and for blood flow estimation [5]. Here a broadly focused transmit beam is emitted, and multiple receive beams are generated simultaneously by steering the beams in slightly different directions. While the parallel receive beamforming presented in [4] is limited to sample closely spaced lines within the transmitted beam, the proposed method provides a larger flexibility in where data is sampled, for instance enabling MFQ imaging. Ultimately, the two methods could be combined, introducing the possibility of making real-time CFMs of 3D volumes without ECG-gating.

II. THEORY

A. Autocorrelation estimator

The autocorrelation method for ultrasonic blood flow imaging was first introduced by Kasai et. al. [1]. Narrow-band pulses are repeatedly emitted along the same direction, and the received rf data are Hilbert transformed to give the in-phase and quadrature components. This results in a complex matrix $y(l, i)$, where i is the transmission number and l is the sample

index along the rf line corresponding to depth $d_l = \frac{c}{2f_s}l$. The axial velocity at a given depth can be estimated as [1]

$$v_z = -\frac{c f_{prf}}{4\pi f_0} \arctan\left(\frac{\Im\{R(1)\}}{\Re\{R(1)\}}\right), \tag{1}$$

where c is the speed of sound, f_{prf} is the pulse repetition frequency, f_0 is the center frequency of the emitted narrow-band signal, $\Re\{\cdot\}$ and $\Im\{\cdot\}$ denotes the real and imaginary parts, and $R(1)$ is the complex auto-correlation function of $y(l,i)$ at lag 1, evaluated at a certain depth corresponding to $l = L_d$. This autocorrelation function can be estimated by

$$\hat{R}(1) = \frac{1}{(N-1)N_l} \sum_{l=0}^{N_l-1} \sum_{i=0}^{N-2} y(l+L_d,i)y^*(l+L_d,i+1), \tag{2}$$

which includes an averaging over N_l rf samples. This has been shown to lower the variance of the estimated autocorrelation function, and thereby increase the accuracy of the velocity estimate [6].

By demanding that a whole period of the signal must be observed to distinguish the flow signal from that of a stationary structure, the minimum detectable velocity of an autocorrelation estimator is given by [2]

$$v_{min} = \frac{c}{2}\frac{f_{prf}}{N f_0}. \tag{3}$$

This is a quite conservative demand, and v_{min} is not considered a rigid limit. According to the Nyquist sampling theorem, the maximum detectable velocity of an autocorrelation estimator is [2]

$$v_{max} = \frac{c}{2}\frac{f_{prf}}{2f_0 + B}, \tag{4}$$

where B is the bandwidth of the emitted signal. The variance of the velocity estimate assuming a constant velocity, and under the assumption that no noise is present, can be approximated by [2]

$$\sigma_v^2 = \frac{c}{4\pi^2 f_0^2}\frac{f_{prf}}{T}|v_z|, \tag{5}$$

where T is the duration of the emitted pulse. Both the velocity range and the performance of the estimator are therefore dependent on the center frequency f_0.

B. Spectral velocity estimation

A spectrogram displays variation of the spectral content of the slow-time signal over time, hereby yielding a direct measure of the axial velocity of the moving blood. It can be estimated from the complex signal matrix $y(l,i)$ when the number of observations is sufficiently high. At a certain time instance $t = k/f_{prf}$, the power spectrum over pulse repetitions is estimated from N_s pulse repetitions and averaged over a number of rf samples N_l, which is known as the range gate. The segment size N_s is chosen low enough to capture the frequency variations over time and high enough to give an acceptable spectral resolution. A window $w(i)$ is often applied
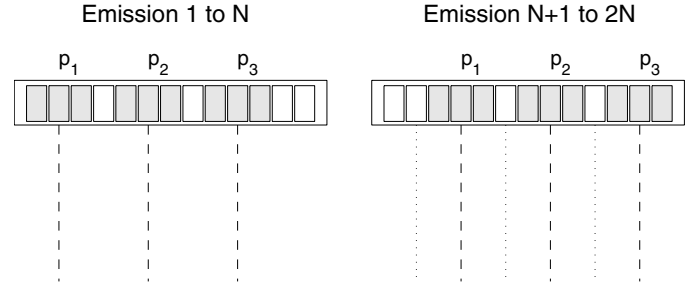

Fig. 1. The basic principle of the proposed method. Three different pulses $p_1(t)$, $p_2(t)$, and $p_3(t)$ are emitted N times and the received signals are separated in the receiver to simultaneously create three lines of the color flow map. The position of the transmitting sub-apertures are shifted and new lines are created.

to the data prior to the Fourier transform. The estimated power spectral density is given by

$$\hat{P}_y(f_p,k) = \tag{6}$$
$$\frac{1}{N_l}\frac{1}{N_s} \sum_{l=0}^{N_l-1} \left| \sum_{i=0}^{N_s-1} w(i)y(l+L_d,i+k)e^{-j2\pi f_p i/f_{prf}} \right|^2.$$

For a certain depth, corresponding to rf index L_d, $\hat{P}_y(f_p,k)$ is a function of frequency f_p and time given by pulse repetition number k. Using a Hamming window given by

$$w(i) = 0.54 - 0.46\cos\left(\frac{2\pi i}{N-1}\right), \quad i = 0,1,...,N-1 \tag{7}$$

will lower the spectral side-lobes, coming from the limited observation time, to approximately -40 dB.

C. Spatial encoding using frequency division

The proposed method aims at increasing the frame rate of color flow mapping by simultaneously sampling data for multiple lines in the CFM. Fig. 1 (left) shows a linear array transducer simultaneously emitting $M = 3$ different pulses $p_1(t)$, $p_2(t)$, and $p_3(t)$ using different sub-apertures. Assuming the signals can be separated in the receiver, three different lines can be beamformed after each emission. The emission is repeated N times at a given pulse repetition frequency f_{prf} and three lines in the CFM are created using the autocorrelation estimator [1]. The transmitting sub-apertures are slid across the aperture and the emissions are repeated until data for the entire CFM is collected. The time used for collecting data will in this example only be a third of that normally used.

Alternatively, the CFM data can be acquired as is usually done, using one narrow band signal $p_1(t)$. Simultaneously, data for two spectrograms can be acquired using the signals $p_2(t)$ and $p_3(t)$. These signals are repeatedly emitted from the same sub-apertures, continuously sampling along the same lines. Hereby a CFM and two spectrograms are acquired simultaneously yielding an expansion of triplex imaging named Multi-Frequency Quadroplex imaging (MFQ). The data acquisition time will equal that normally spent on acquiring a CFM. At some point, a transducer element will be required to emit both
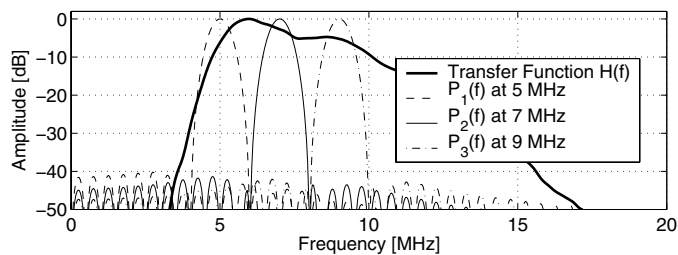
Fig. 2. The two-way system transfer function and the amplitude spectra of the three emitted pulses.

Fig. 3. Physical setup of the phantom experiment. The transducer is mounted at height $z = z_0$ over the tube center at an angle of $\theta = 60°$. The tube has radius R. The figure also shows the three regions of the CFM made at different frequency bands. Each region is sampled left to right and all regions are sampled simultaneously. The position of the three velocity profiles correspond to the profiles shown in Fig. 4.

a delayed version of $p_1(t)$ and for instance $p_2(t)$. To use the entire amplitude range while emitting each signal, $p_2(t)$ and $p_3(t)$ are delayed so that they are emitted later than $p_1(t)$.

The emitted signals $p_m(t)$ are designed as simple narrow-band pulses at different frequencies. They are given by

$$p_m(t) = w(t)\sin(2\pi f_m t), \qquad 0 < t < T \qquad (8)$$

for $m = [1, 2, 3]$, where f_m is the center frequency of the mth signal, and T is the pulse duration. $w(t)$ is a window designed to reduce the spectral leakages into the other bands. A Hamming window, which has spectral side-lobes below -40 dB, is used. The signals are then separated in the receiver by filters matched to the emitted signals, $h_m(t) = p_m(T-t), 0 < t < T$. The number of pulses M, which can be emitted, is determined by the transducer bandwidth compared to the bandwidth of the signals $p_m(t)$. Fig. 2 shows the transfer function of a commercial linear array transducer (thick line), and the amplitude spectra of three signals in (8) using $f_m = [5, 7, 9]$ MHz and $T = 2$ μs (thin lines). The spectral leakage is below -40 dB for all bands.

The minimum detectable velocity is now limited by the lowest frequency band centered at f_1 through (3), and the maximum detectable velocity is limited by the highest frequency band f_M through (4). The increased frame rate, thus, comes at the expense of a slightly decreased velocity range compared to a single narrow-band pulse. Furthermore, the method requires M parallel beamformers, which is incorporated in most commercial high-end scanners at present, and a matched filtration of the channel RF-data.

III. SIMULTANEOUS SAMPLING OF MULTIPLE CFM LINES

The method was tested in a re-circulating flow rig, where a blood mimicking fluid was pumped at a constant velocity through a rubber tube submerged in a water tank. The physical setup of the phantom experiment is shown in Fig. 3, which also show the definition of the tube depth $z_0 = 33.5$ mm, the tube radius $R = 6$ mm, and the beam-to-flow angle $\theta = 60°$. The peak velocity was $v_0 = 0.1$ m/s.

The data acquisition was performed using a 128 element linear array transducer and the RASMUS multi-channel sampling system [7]. This system can emit arbitrary waveforms on 128 elements, and sample 64 transducer elements in real-time. To acquire data from all 128 elements, the emission is
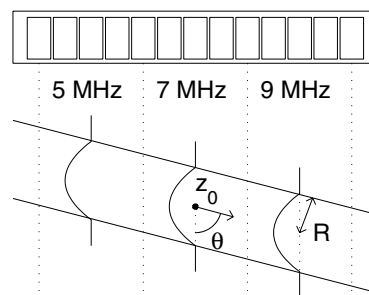
repeated and multiplexing is used. This effectively halves the frame rate. The three signals shown in Fig. 2 were used and the parameters for the experiment are given in Table I.

TABLE I

PARAMETERS USED FOR THE PHANTOM EXPERIMENT.

Parameter	Value
Transducer type	Linear array
Number of transducer elements	128
Transducer element pitch	0.26 mm
Transducer element height	4 mm
Elevation focus	15 mm
Transfer function	See Fig. 2
Transmit focus depth	28.7 mm
Number of transmit bands	3
Center frequencies, f_1, f_2, and f_3	5, 7, and 9 MHz
Pulse duration, T	2 μs
Amplitude tapering	Hamming window
Number of emit. elements / band	22
Transmit apodization	Tukey window
Number of receiving elements	128 (2x64 through multiplexing)
Receive apodization	Hanning window (over 64 elem. cnt. around the current image line)
Sampling frequency, f_s	40 MHz
Pulse repetition frequency, f_{prf}	1.65 kHz (3.3 kHz incl. multiplex)
Number of shots per estimate, N	32 (64 incl. multiplexing)
Clutter filtering	Subtracting mean of N signals
Number of CFM lines created	33
Inter-line spacing	0.52 mm

The acquired data was processed as described in Section II. 17 repetitions of the entire sequence was processed, yielding 17 full color flow maps. The velocity profiles at the center line of each of the three regions (see Fig. 3) were extracted and the mean and standard deviation over the repetitions were found. These are shown in Fig. 4 along with the expected parabolic velocity profile.

Fig. 4 reveals a slightly higher standard deviation in the lower frequency bands. The standard deviations of the velocity estimates are averaged over the tube diameter yielding 3.1, 2.5, and 2.2 % of the theoretical peak velocity $v_0 = 0.1$ m/s for the 5 MHz, 7 MHz, and 9 MHz bands, respectively. The increase in standard deviation, although present, is not as large as predicted by (5). For this phantom experiment performed
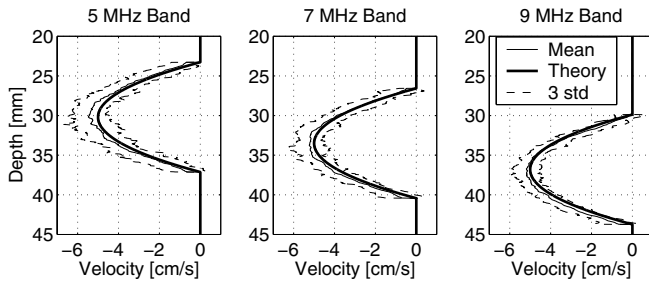
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Fig. 4. Three resulting velocity profiles made at different frequency bands. The plot shows the expected velocity profile (thick line), the average of 17 profiles (thin line), and three times the standard deviation of the 17 profiles (dashed line).
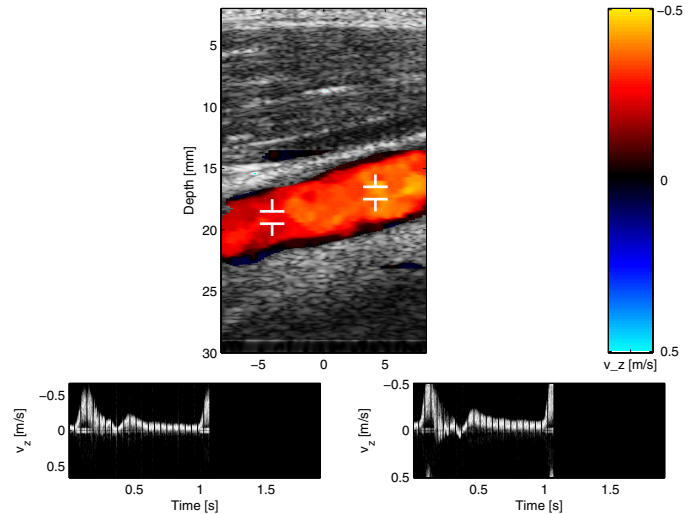

Fig. 5. Multi-Frequency Quadroplex image of the common carotid artery during systole. The upper image shows the color flow map using the 5 MHz band, the left spectral Doppler shows the 7 MHz band using the leftmost range gate, and the right spectral Doppler shows the 9 MHz band using the rightmost range gate.

in water a quite good signal-to-noise ratio is attained, and no frequency dependent attenuation is present. The increased standard deviation at lower frequency bands is therefore qualitatively in accordance with (5), which is derived assuming no noise.

IV. MULTI-FREQUENCY QUADROPLEX IMAGING

At present, triplex imaging is only possible at very low frame-rates due to the switching between acquisition of CFM data and spectrogram data. The proposed method can be used to solve this problem, and further to do MFQ imaging as described in Section II-C. The MFQ imaging was tested *in vivo* using a pulse repetition frequency of $f_{prf} = 12$ kHz (24 kHz including multiplexing) and a transmit focus depth of 15 mm. The remaining parameters were set as described in Table I. An examples from the common carotid artery of a healthy male volunteer is shown in Fig. 5. The CFM is made using the 5 MHz band and is composed of 33 lines, each sampled 16 times (32 including multiplexing), and the B-mode image is made from 65 emissions. The leftmost spectrogram was made using the 7 MHz band and the rightmost using the 9 MHz band. Each spectrogram estimate was made from 64 lines using the method described in II-B. The frame rate was 21.4 Hz. While the B-mode data was acquired (which happens 21.4 times each second), no spectrogram data are available. This is a result of the B-mode emissions using the entire transducer bandwidth. In Fig. 5, this is seen as gaps in the spectrogram. These gaps are not inherent for the proposed method, but are dependent on how the emissions are ordered, and there are multiple ways of avoiding these gaps [8], [2], [9].

V. CONCLUSION

A data acquisition method for simultaneously sampling multiple lines using different narrow frequency bands has been presented. The method can be used for increasing the frame rate of CFMs or for MFQ imaging, featuring a CFM and two independent spectrograms. The MFQ imaging mode yields the possibility of studying the change in flow over an arterial stenosis by placing a range gate at both sides of the stenosis and monitoring the flow over time, while still maintaining the CFM.

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