Real-time 3D Medical Ultrasound — Signal Processing Challenges

Sverre Holm

Department of Informatics, University of Oslo
P. O. Box 1080, N-0316 Oslo, Norway
E-mail: Sverre.Holm@ifi.uio.no

ABSTRACT

Real-time 2D ultrasound systems are used routinely in every hospital and are a huge success both technically and commercially. This paper discusses the signal processing problems that needs to be tackled in order to move from 2D to 3D real-time ultrasound systems.

The first problem discussed is that of handling 2000—10000 elements in the transducer. Sparse array methods is a way to reduce the number of elements and cost without compromising quality. Examples of performance with sparse arrays are presented.

The second important problem is that of frame-rate. In 3D the frame-rate will be so low that real-time acquisition will be impossible unless some form of parallelism is exploited. Various ways of doing that such as multiple receive beams, coded transmit excitation and limited diffraction beams are discussed.

1. INTRODUCTION

Real-time 2D medical ultrasound systems are in common use in every hospital today. The images are of so high quality that in many medical specialities ultrasound is to some degree replacing other imaging modalities. This is also due to the mobility of the scanner and the simple procedures involved in perfoming an ultrasound examination compared to e.g. computer tomography or magnetic resonance imaging.

Despite this success, new ultrasound imaging modalities are under development. One of the interesting ones, both from a clinical and a technical perspective, is 3D ultrasound. Experimental systems have been underway for a while and have demonstrated the benefits. To take the field of cardiology as an example, the advantages are improved surgical planning due to better diagnosis of complex anatomy like heart valves and septal defects, unrestricted 'any-plane' 2D imaging, and improved volume quantification.

In most of the demonstrated 3D systems the data acquisition has been based on mechanical scanning in at least one of the dimensions. One of the main problems of 3D ultrasound is the limited frame rate achievable due to the slow data acquisition, but 2D arrays with electronic scanning in both dimensions have the greatest potential for acceptable frame rates. This is due to the greater beam agility and the possibility for parallel beams.

The two areas where work remains to be done in order to achieve the goal of 3D real-time imaging are 2D array technology and beamforming and signal processing methods and hardware realization of parallel beam formation.

2. 2D ARRAYS

In medical imaging, 1D arrays with from 48–192 elements are used to do 2D scans [1]. In order to achieve 3D imaging in near real-time it is necessary to use 2D arrays with close to the squared number of elements, typically 2000–10000. Since each element needs a cable and an electronic front-end that includes preamplification, A/D-converter and digital programmable delay, it is desirable to reduce the element count as much as possible [2]. For this reason sparse 2D arrays, where elements are removed by thinning, are considered to be necessary.

The starting point are arrays that are regularly sampled with sample distance equal to half the wavelength. This is the spatial equivalent of the Nyquist criterion. It is assumed that the sampling is regular, ie. on a square grid. The thinning may be random or it may be found from some sort of optimal algorithm. The trivial thinning of just keeping the full central part of the aperture is avoided. In this way the aperture is maintained and thus the resolution. The remaining elements may be weighted or they may be unweighted.

Steinberg [3] has given a comprehensive theory for the unweighted randomly thinned array. The main results for the far-field continuous wave (CW) beampattern is that the sidelobe level can be described in a statistical sense. Some distance away from the main lobe, the ratio of the mean sidelobe power to the main lobe peak power is $1/M$ where M is the number of remaining elements. This result is independent of the statistical distribution of the elements.

2.1. Sparse array optimization

There is a long history in the radar litterature for analysis of beampatterns for sparse arrays for the far-field single-frequency case (analysis of the one-way beampattern). In ultrasound imaging, this was the approach used in [4] where it was partially confirmed that Steinberg's results for average one-way sidelobe levels can be squared to estimate the levels for the two-way beampattern for pulsed 2-D arrays. In high-resolution sonar imaging there has also been a recent interest in sparse arrays [5, 6]. We have done work to find the properties of random-like thinning patterns. It is based on optimization of the one-way response by either changing the element weights or the element positions or both [7]. Due to the properties of 2D array elements in ultrasound (high impedance, low sensitivity) it is often undesirable to weight the elements. The goal of the work was therefore not primarily to propose practical weighting functions, but rather the optimization methods are used to find properties of the beampattern of such arrays. Of special interest is the determination of the minimum peak sidelobe level and comparison with the predictions from random theory. A method was also described for optimizing the element positions of a random-like sparse array. This optimization gives results that are more directly useful in an array design.

The optimization criterion is usually a minimization of the maximum sidelobe. This is a criterion which is related to imaging of a strong reflecting point target in a non-reflecting background containing other point targets. An alternative criterion is to minimize the integrated sidelobe energy. In a medical imaging system, this is related to imaging of a non-reflecting area like a cyst or a ventricle in a background of reflecting tissue. Some results on weight optimization for 1D arrays using this criterion and quadratic optimization have been reported in [8].

2.2. The beam pattern of a planar array

The far-field continuous wave (CW) beampattern of an array with N omnidirectional elements is given as [9]:

$$W(\vec{k}) = \sum_{n=1}^{N} w_n e^{j\vec{k}\cdot\vec{x}_n} \tag{1}$$

where the array element locations are $\vec{x}_n \in \mathbb{R}^3$ with the corresponding weights $w_n \in \mathbb{R}$. The wavenumber vector $\vec{k} \in \mathbb{R}^3$ has amplitude $|\vec{k}| = 2\pi/\lambda$ where λ is the wavelength.

Let the unit direction vector be $\vec{s}_{\phi,\theta} = (\sin\phi\cos\theta, \sin\phi\sin\theta, \cos\phi)$ in rectangular coordinates, see Fig. 1. Then the wavenumber vector is $\vec{k} = 2\pi\vec{s}_{\phi,\theta}/\lambda$.

The elements of a 2D planar array are located in the xy-plane with element n at $\vec{x}_n = (x_n, y_n, 0)$.

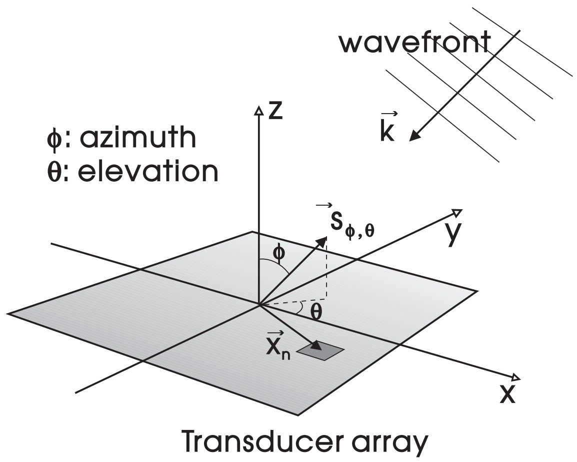

Figure 1. A 2D planar array with coordinate system.

The beampattern is:

$$W(\phi,\theta) = \sum_{n=1}^{N} w_n \exp\left(j\frac{2\pi}{\lambda}\sin\phi(x_n\cos\theta + y_n\sin\theta)\right) \tag{2}$$

This is the array response to a monochromatic wave from direction (ϕ,θ). It has the following properties:

- For real weights, the beampattern is conjugate symmetric, i.e. $W(k_x, k_y) = W^*(-k_x, -k_y)$.
- Symmetric arrays with symmetric weights give a real beampattern.

2.3. Optimization of beampattern

Two optimization problems may be formulated as linear programming problems. The first is a minimization of the maximum sidelobe level by varying element weights. The second problem gives rise to a mixed integer linear programming problem which is considerably harder to solve. It is a minimization of the number of active elements and an optimization of the weights in order to achieve a specific maximum sidelobe level. Due to the properties of the linear programming algorithms it is required that the beampattern is real, i.e. that the array is symmetric. The formulation of these problems may be found in [7].

2.4. Example of optimized 2D sparse arrays

A 2D array for 3.5 MHz with 12 by 12 elements with half wavelength spacing in both dimensions was considered. The array is inscribed in a circle giving 112 elements. Random thinning to 64 elements (57%) and optimization of the weighting gives a beampattern with a sidelobe level of -12 to -15 dB. The procedure for finding the optimal thinning and weighting was then used with a sidelobe target of -19.5 dB. The optimized layout was then input with varying start-angles in the

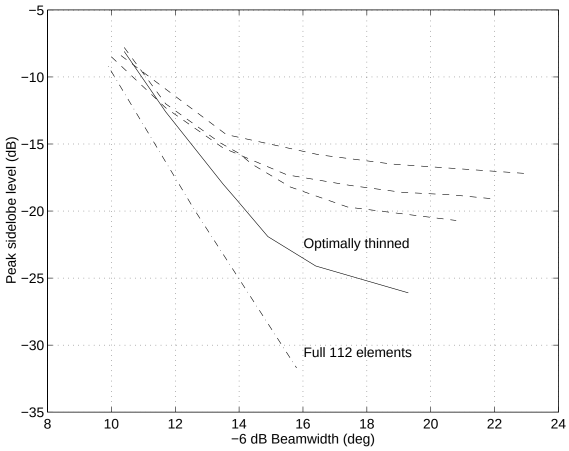

Figure 2. Sidelobe level as a function of beamwidth for several weight-optimized uniform sidelobe cases: 112 element full array (dash-dot line) and three realizations of random thinning to 64 elements (dashed lines). The best result is obtained for a layout-optimized 62 element thinning (solid line).

weight optimization algorithm. The peak sidelobe level can now be reduced down to -20 to -22 dB (Fig. 2). Each curve is the result of between 5 and 8 optimizations with different start values for the azimuth angles. The sidelobe value should be compared to the value predicted for mean sidelobe level of $1/64 = -18.1$ dB, and shows that there is a potential of getting a peak value which is 3 dB lower than that predicted for the mean if optimized thinning patterns can be found. This is about the largest array size where optimized element layouts can be found with reasonable use of computer resources using the linear programming methods. The four element layouts are shown in Fig. 3. An example of a beampattern is shown in Fig. 4.

Further examples may be found in [7]. Algorithms that are suitable for optimizing more realistically sized arrays with thousands of elements must be based on heuristic methods that do not guarantee global convergence [10, 6]. Present research is focused on faster optimization methods and on understanding better the relationship between image quality and the thinning. Alternative ways based on periodic thinning have also been proposed [11].

3. FRAME RATE IN 3D IMAGING

The problem of getting high enough framerates may be illustrated by an example using a typical cardiac transducer array. Assume a frequency of $f = 3.5$ MHz and a velocity of sound of $c = 1540$ m/s. The wavelength is $\lambda = c/f = 0.44$ mm. The framerate in 1D, 2D, and 3D imaging will now be found.

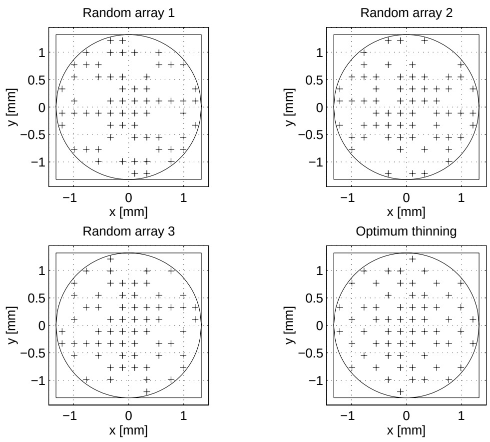

Figure 3. Element layouts for 112 element full array thinned to three different random 64 element layouts and a 62 element optimized layout. The random arrays are sorted according to the peak sidelobe level in Fig. 2 with Random 1 having the highest peak sidelobe level for large beamwidths.

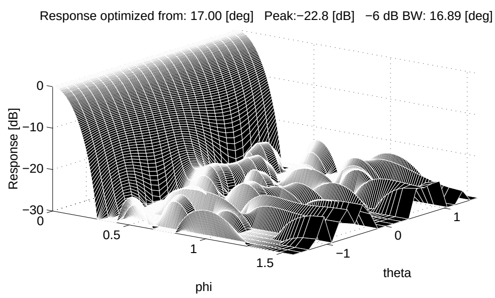

Figure 4. Beampattern for weighted, optimally thinned layout (62 elements out of 112)

3.1. 1D Framerate

The Pulse Repetition Frequency (PRF) is the 1D framerate, or the framerate in M-mode (Motion mode — a mode where a beam is stationary in space and samples a line in time). It it determined by the maximum required depth.

$$PRF = c/2d_{max} \tag{3}$$

Since ultrasound attenuation increases with frequency, the maximum depth will decrease with frequency. For our example, a typical depth will be $d_{max} = 15.4$ cm. This gives $PRF = 5000$ Hz.

3.2. 2D Framerate

The 2D framerate must be found from the beamwidth in the azimuth direction:

$$\theta_{az,-3dB} = k_{-3dB} \cdot \lambda / D_{az} \tag{4}$$

where $k_{-3dB} = 0.89$ for an unweighted transmitter. According to the sampling theorem there must be at least two beams per beamwidth, thus the distance between beams is $\theta_{az} = 0.5 \cdot \theta_{az,-3dB}$. A typical azimuth aperture of $D_{az} = 19$ mm gives a beamspacing of $\theta_{az} = 0.6°$. For a sector size, Θ_{AZ}, the number of transmit beams is:

$$N_{az} = \Theta_{AZ}/\theta_{az} \tag{5}$$

The 2D framerate is determined by the PRF and the number of transmit beams per frame:

$$FR_{2D} = PRF/N_{az} \tag{6}$$

An azimuth sector of $\Theta_{AZ} = 90°$ in our example will require $N_{az} = 152$ beams and the resulting frame-rate will be $FR_{2D} = 32.8$ frames/second which is an acceptable frame-rate. It may also be increased by for instance restricting the sector size (zooming).

3.3. 3D Framerate

There are several ways that 3D acquisition can take place, but one of the simplest ones is tilting acquisition. It can be implemented both with mechanically moving 1D arrays or with a 2D array and electronic scanning. A sampling in the elevation (short-axis) direction similar to the one in the azimuth (long-axis) direction takes place. Thus the angular spacing in the elevation direction is:

$$\theta_{el} = 0.5 \cdot k_{-3dB} \cdot \lambda / D_{el} \tag{7}$$

The number of scans in the elevation direction is:

$$N_{el} = \Theta_{EL}/\theta_{el} \tag{8}$$

Maximum 3D framerate is:

$$FR_{3D} = FR_{2D}/N_{el} \tag{9}$$

Assume an aperture in the elevation direction of $D_{el} = 13$ mm. The elevation beam-spacing is then $\theta_{el} = 0.86°$. For a sector size of $\Theta_{EL} = 60°$, the 3D frame-rate will be $FR_{3D} = 0.47$ volumes/second.

4. INCREASING FRAME-RATE

The result for 3D framerate is much lower than desired. In fact, one would have liked to have the 2D frame-rate in 3D. The question is: How can one achieve an increase of 50—100 in 3D frame-rate? Several ways to achieve this have been proposed.

4.1. Parallel receive beams

This mode is used routinely in sonar where often the whole sector, Θ_{AZ} is illuminated so that 2D frame-rate is equal to the PRF. It cannot be used in the same way in ultrasound due to the increased reverberation level in the medium and the intensity limitations due to the risk for biomedical hazard. An adaptation of the method was first proposed for ultrasound in [12]. If the transmit beam is made a little wider than usual, a receive beamformer with several parallel beams can be used to acquire several beams at slightly offset angles. Today 2—4 parallel beams are used for 2D imaging, giving an increase in frame-rate by the same factor in (6). In 3D, this number may be squared giving a factor of 4—16.

For 4 parallel beams in each dimension in the example, the 2D frame rate will increase to $FR_{2D} = 131.2$ frames/sec, and the volume frame rate will increase to $FR_{3D} = 7.5$. A decrease of the azimuth angle to $\Theta_{AZ} = 60°$ will give an additional increase to $FR_{2D} = 196.8$ and $FR_{3D} = 11.3$. For cardiac imaging, where the heart moves with about a beat per second in a normal subject, the volume frame-rate is still a little low, but for more stationary organs it is satisfactory. Even higher rates are achievable if the angular sampling theorem is slightly violated. The parallel beam approach requires hardware in the form of parallel receivers.

4.2. Coded transmit pulses

Instead of sending a wide transmit beam, in this approach a transmitted signal consisting of many coded pulses, each one individually beamformed for a unique direction is sent [13]. It is proposed to send one in each of the desired azimuth directions. On reception, each direction is recovered by a pseudoinverse operator implemented in the form of a transversal filter bank. The pseudoinverse operator is the key to the performance because it eliminates correlation artifacts that are due to the non-zero correlation between the different coded sequences used for transmission. Standard processing like a matched filter is not able to do that.

This system would allow for 2D data acquisition at the same rate as 1D data acqusition, eg. $FR_{2D} = PRF = 5000$ Hz in the example. When used in a 3D system, with parallelism only in the azimuth direction, a 3D frame rate of $FR_{3D} = 0.47$ volumes/second would increase by the number of azimuth beams $N_{az} = 152$ to $FR_{3D} = 71.5$ volumes/second which is an acceptable figure.

The cost would be a new transmitter design with the ability to send coded pulses in many directions simultaneously and a new receiver design, both of them with greatly increased complexity. The method needs to be validated for proper operation in a medium which is aberrating and has attenuation, and in a situation where peak and average intensitites are limited due to the risk for biomedical hazard.

4.3. Limited diffraction beams

In [14] it is proposed that different limited-diffraction beams may be summed to give array beams. Such beams may be used in conjunction with the transmission of plane waves, where the array-beams are used for reception. It is claimed that the image may be recovered using the Fast Fourier transform and thus simplify beamformer hardware. The frame-rate increase is achieved since only a single transmission is required to illuminate the whole volume. Thus 3D framerates equal to the PRF is achieved. One limitation is that since a single plane wave is used, steering is not possible, limiting the application to those of linear arrays. The limitations mentioned for the coded transmission method may also be applicable.

5. CONCLUSION

The main signal processing challenges in order to achieve real-time 3D medical ultrasound have been outlined. The problem of reducing the number of elements in order to save cost in the front-end of the medical imaging instrument was discussed and examples of optimized ways to thin an array were shown.

Since this paper has focused on signal processing, the problem of making the 2D array transducer has not been addressed. This gives challenges in material technology, acoustics and connection technology.

The problem of getting high enough update rate for the scanned volume was then discussed and an example given that illustrates the problem. The limited velocity of sound is really the fundamental problem. Possible solutions are found in exploiting parallelism by acquiring several beams in the image simultaneously. This can be done by parallel receive lines, transmission of coded pulses, and by Fourier reconstruction of the imaged object using limited diffraction beams.

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