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Review paper

Which transducer array is best?

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Abstract

A review of the different transducer array technologies is given, with emphasis on their ability to meet the most important clinical requirements. Spatial resolution, imaging artifacts, and sensitivity are discussed, both for tissue imaging and color flow imaging. The qualities of each array type, for example small footprint, narrow slice thickness, high resolution etc., are analyzed in relation to clinical applications.

Keywords: Ultrasound; Transducer; Array; Resolution; Image quality

1. Introduction

Developments in transducer arrays and electronic beam forming have been responsible for much of the improvement in the quality of ultrasound images over the last decade (Macovski 1979; Wells 1992). The major challenges in beam forming have been to obtain narrow beams with very high sensitivity and low side-lobe levels, consistent with restraints on probe size and shape imposed by anatomical considerations. By individually delaying the signals to and from different elements, a variety of beam focusing and steering techniques are possible.

In what is commonly referred to as dynamic focusing, the receiving focus of the transducer is made to follow the source of echoes from increasing depths. By using different amplifications for the signals from the different elements, a technique called apodization, the side-lobe levels may be reduced, and hence the background noise in the image due to pick-up by the side-lobes may be reduced. The penalty for using apodization is that the width of the main lobe is increased and the lateral resolution is reduced. This can often be compensated for by increasing the width (aperture) of the transducer, although in cardiac imaging and endocavity imaging the transducer aperture is limited by anatomical features such as the inter-rib spacing or the space available in, for example, the esophagus, the vagina, the rectum or a blood vessel. The active width and apodization of the

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array are usually also varied in reception, so that a reduced aperture is used to receive the near-range echoes. This is called dynamic aperture and helps to maintain constant beam width and sidelobes when focusing at different ranges.

In transmission, it is not possible to vary the focus to follow the ultrasound pulse, because once the pulse has been transmitted to a particular focus it is beyond the control of the machine. Hence, in order to reduce the effective transmission beam width it is necessary to transmit several pulses, each set for a different position of focus along the same scan line, and only use echoes from a limited range around the focus of each transmission to build the image. This is known as multiple zone transmission focusing. An extension of this, sometimes referred to as confocal imaging, is to transmit high frequency ultrasound pulses for imaging at the nearer ranges, and lower frequency pulses for imaging the greater depths. Besides optimizing the ultrasonic frequency for the different depths, this technique allows a higher frame rate than is possible with standard multiple zone focusing. This is because, at the high frequencies used for the shorter ranges, the signal attenuates faster and it is not necessary to wait for echoes to return from the deeper ranges before transmitting the following lower frequency, and more deeply focused, pulse. A necessary prerequisite for confocal imaging has been the development of wide bandwidth transducers, since the high frequency near-range pulses and the low frequency far-range pulses must be transmitted by the same transducer.

For percutaneous abdominal and vascular imaging there are few practical limitations on the aperture, but heterogeneity in the tissue, especially fatty layers between muscular tissue, introduces aberrations in the phase fronts (Fig. 1) which impair the focusing of the beam. By adding individual delays to the signal channels from each element of an array, it is possible, at least in part, to correct for such aberrations. Hence, one expects to see even further improvements in the image quality from future developments in array beam forming.

For cardiac applications, a high frame rate is essential and, if the benefits of improved lateral resolution achievable by multiple pulse transmission are to be gained, it will be necessary to transmit

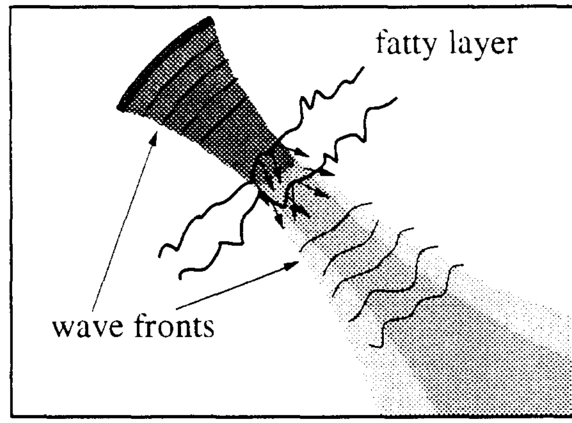


Fig. 1. Aberrations of the wavefronts as they pass through irregular layers of fat, producing a defocusing of the beam.

and receive beams in different directions simultaneously.

2. Commonly used array types

2.1. Switched linear arrays

Switched linear arrays (Bom et al. 1973) are composed of a large number of elements in a line, as illustrated in Fig. 2a. The beam is stepped along by selecting a sub-group of the elements for each beam position, and steering and focusing are obtained by adding a variable delay to each element channel. The linear array has a flat front producing a rectangular image format which is used for vascular imaging and some abdominal imaging. For blood velocity imaging and Doppler measurements it is desirable that the beam can be steered to an angle, and this can be achieved by varying the delays of the signals to and from each element (Fig. 2b).

A weak convex curvature array produces a useful sector imaging format for deep abdominal imaging, with a wide image field at large depths, while having a moderate scan head size, as illustrated in Fig. 3a.

An array with tight convex curvature (Fig. 3b), produces a sector image format with a small footprint, so that imaging between the ribs is possible. At the outer edges of the array there are problems with reflections from the ribs, so the array is not particularly well suited to cardiac imaging.

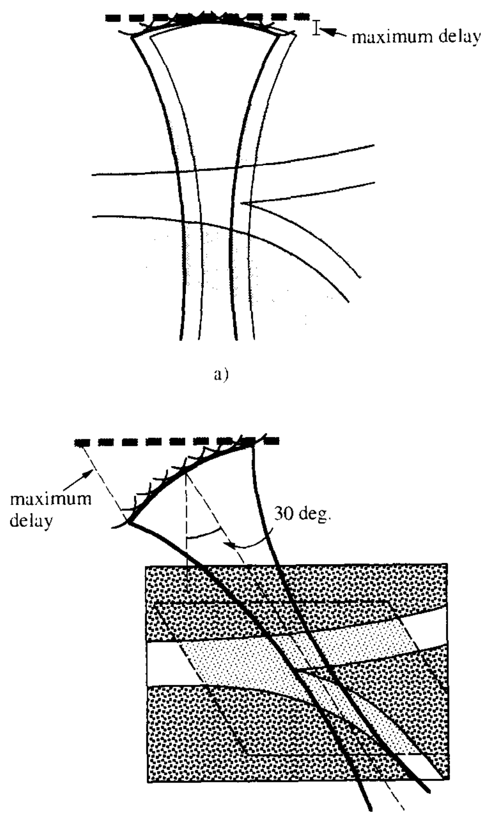


Fig. 2. Switched linear arrays. (a) Illustrates how a lateral sweep of the beam is achieved by advancing an active group of elements by one element at a time, while electronically variable focusing is achieved by transmitting from the outer elements in advance of the central elements. (b) Illustrates how the beam direction can be deflected by progressively delaying the transmissions from elements across the array. This allows the beam to be aligned better with blood flow in order to give a greater Doppler shift.

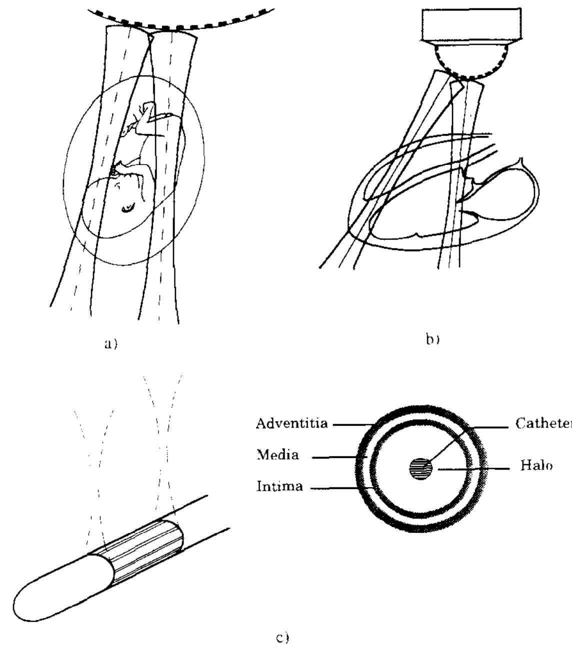


Fig. 3. Three forms of sector scanners using switched arrays. (a) A weakly curved array (45° sector). (b) A tightly curved array (90° sector). (c) A cylindrical catheter tip array (360° array) and a representation of a scan from within a blood vessel.

A cylindrical ring array has been mounted at the tip of a catheter or a gastroscope to produce 360° cross-sectional imaging from within vessels or the esophageal-intestinal tract, (Fig. 3c). With a concave curvature of the array, it is possible to make all beams cross each other in a sector format at a defined depth (1–2 cm) below the skin, for example, in the space between the ribs (Fig. 4). This gives the possibility of improved access to the heart between closely spaced ribs, but use of this type of array has been limited due to ultrasound reverberations between the array and the skin.

The main advantage of the switched linear and curved arrays is the wide variety of probes and image forms which may be offered with a fairly small amount of electronics. Their principal limitations arise from difficulties in achieving adequate skin contact in some applications, e.g. the heart.

2.2. Phased arrays

Phased arrays produce a beam that is steered in a sector format by adding delays to each element channel (Somer 1968a,b; 1993), as illustrated in Fig. 5. The small footprint of the array makes it well suited for trans-thoracic imaging between the ribs, and endocavity imaging from, for example, the esophagus, the gastro-intestinal tract, the rectum, the vagina, etc. Three-dimensional imaging, for example of the heart from the esophagus, may

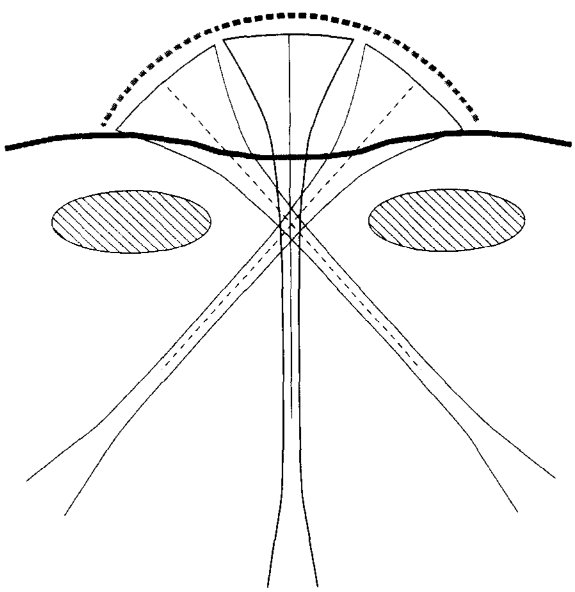


Fig. 4. Sector scanning with a concave switched array. All beams pass through the same point, in this example between ribs.

be obtained by mechanical rotation of an array at the tip of a gastroscope (Pandian et al. 1992).

2.3. Annular arrays

Annular arrays are composed of a set of concentric circular elements, allowing the focus to be varied by individually delaying the signal to and from each element, as illustrated in Fig. 6a (Burckhardt et al. 1975). The beam may be steered to make a sector image by mechanically pivoting the array in a dome filled with fluid. In some cases this gives reduced access between the ribs compared to the phased array, while in other situations the dome helps the access. The array may also be scanned laterally (Green 1977). The sector scanning probe has a small footprint, making it well suited to the same anatomical sites as the phased array.

Scanning systems that produce three-dimensional scanning of the beam in a sector format as illustrated in Fig. 6b, have also been developed (Angelsen et al. 1993a). Using a mirror for scanning the beam, one can also obtain 360° cross-section scanning (Fig. 6c). This method has applications in the esophageal-intestinal tract as well as in blood vessels.

A unique advantage of an annular array is that a symmetrical beam is obtained, producing a thinner scan slice thickness than the other types of arrays, as illustrated in Fig. 7. An advantage over sector scanners employing phased arrays is that the annular array has fewer elements (typically 4–8) with larger surface area per element. This makes it easier to make low-loss transducers, thereby improving the sensitivity for Doppler measurements of blood velocity and colour flow imaging (CFM). The fewer elements also make it easier to avoid cross-talk between the transmitting and receiving transducers in continuous wave (CW) Doppler measurements of blood velocity. The annular array has therefore some advantages compared to the phased array for Doppler measurements of blood velocity in the heart (Angelsen et al. 1993). The larger elements also

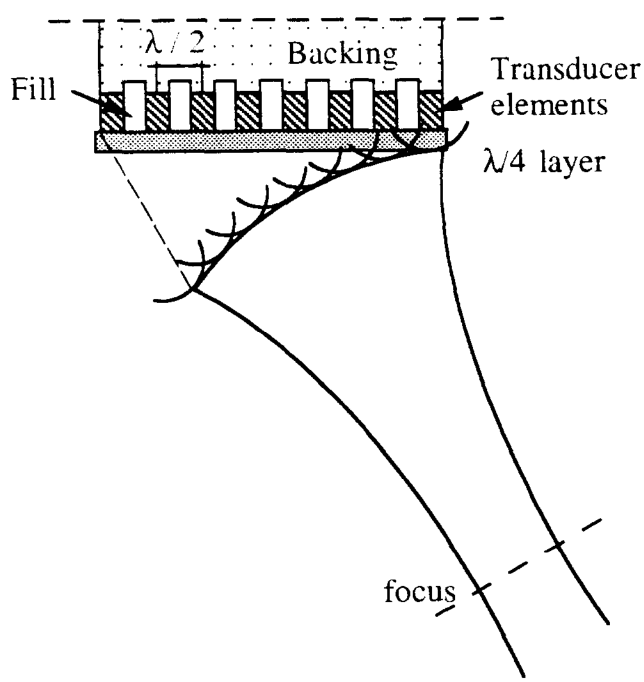
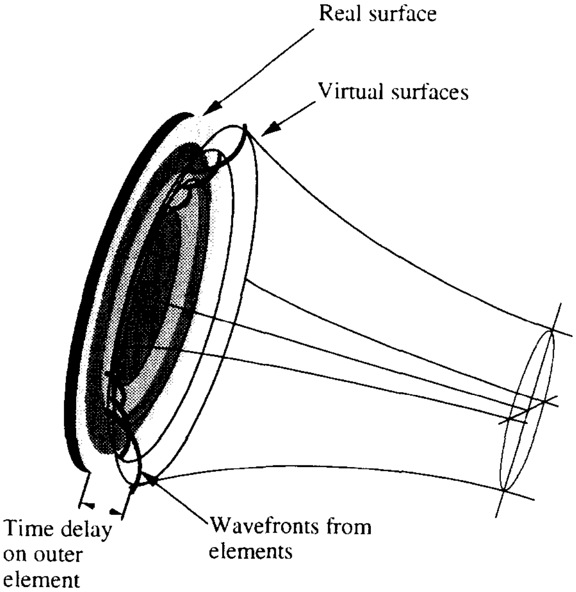


Fig. 5. The closely spaced elements of a phased array may permit variable focusing and beam deflection over a sector of 90° or more, without substantial side-lobes.



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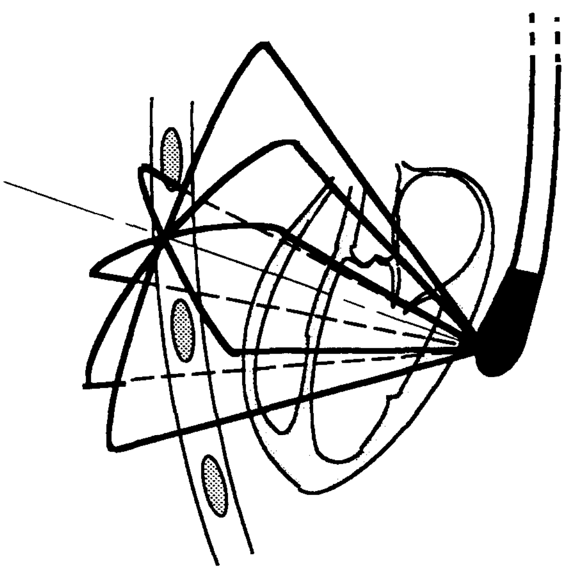

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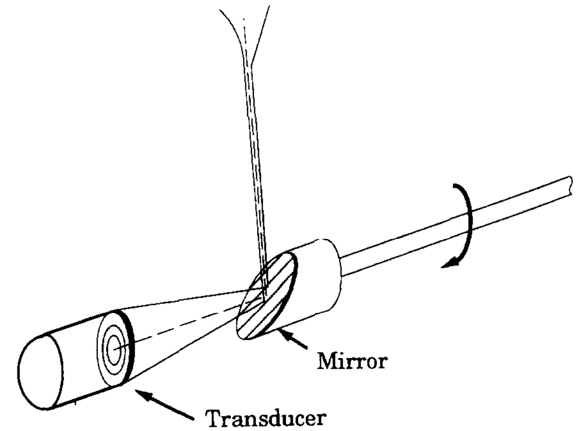

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Fig. 6. (a) Generation of a symmetrically focused beam from an annular array transducer. Delaying transmissions from the outer elements creates a concave wavefront that may be thought of as a virtual radiating surface. The beam is scanned by mechanically rotating or moving the annular transducer array. (b) Some probes allow rotation of the sector scan plane, permitting 3D scanning and rapid changing of a 2D scan plane in endocavity probes. (c) An endoprobe incorporating a rotating mirror permits a 360° scan.

make it easier to produce high frequency transducers with up to 20 MHz centre frequency. These features of the annular array are shared to a lesser extent by switched linear and curvilinear arrays, which also have fairly large elements.

3. Considerations for high resolution and sensitivity

3.1. Focusing in the scan plane

Range (depth) resolution is obtained by transforming the time delay t between transmission of a short pulse, and arrival of the echo from a target at range r as:

$$r = ct/2 \qquad (1)$$

where c is the speed of sound in the tissue. The image range resolution $\Delta r = c\Delta T_P/2$ is determined by the pulse length ΔT_P measured at −6 dB points relative to the peak of the pulse envelope. The pulse length is inversely proportional to the bandwidth B_{TR} of the transducer, which may be thought of as the band of frequencies where the transducer is most efficient. The shortest pulse a transducer can produce is therefore $\Delta T_P = 1/B_{TR}$. In practice, the bandwidth is proportional to the centre frequency f_0 of the transducer, i.e.

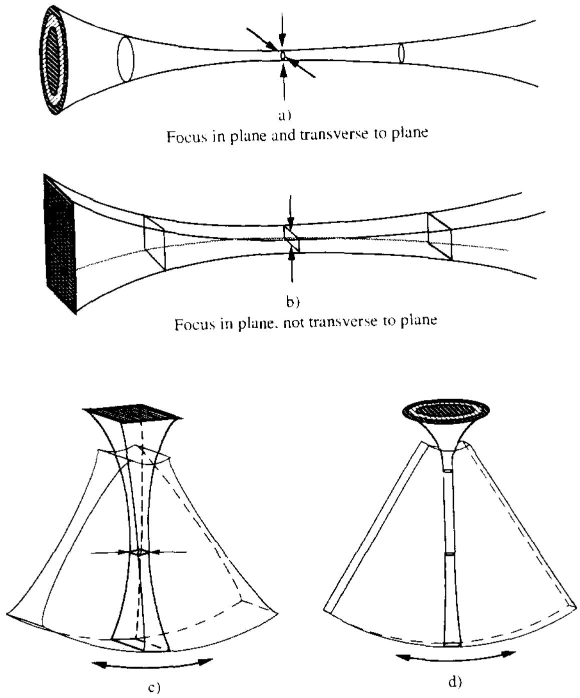

Fig. 7. Dynamic focusing produces a beam with an extended narrow region. For an annular array (a) the beam is symmetrical, whereas for a switched or phased array the benefit applies only to the scan plane (b). Hence the slice thickness of an annular array scanner is narrower than for other scanners (c) and (d).

$B_{TR} = b_{TR}f_o$, where the constant b_{TR} is typically 0.5–0.7 for the new composite transducer materials (Takeuchi et al. 1989). The range resolution is therefore

$$\Delta r = c/2B_{TR} = \lambda/2b_{TR} \tag{2}$$

where the wavelength $\lambda = c/f_o$. Thus Δr is typically $\sim 1\ \lambda$. Frequency dependent attenuation in tissue will reduce the bandwidth, and thus the range resolution obtained will decrease with increasing depth.

A two-dimensional (B-mode) image is obtained by laterally scanning the beam while transmitting pulses to collect range data. The lateral resolution is hence determined by the beam width, which is minimized by focusing the beam. For a circular transducer with diameter D, and focus set at a distance F from the transducer, the diameter of the beam at the focus is

$$d_F = 2F\lambda/D = 2\lambda N \tag{3}$$

where $N = F/D$ is known as the f-number of the transducer. Typical f-numbers are in the range 1.5–4.0 so that d_F is typically in the range 3–8 λ. In this formula, the beam width d_F is measured between the points where the intensity has dropped to −12 dB (or 6.3%) below the intensity on the beam axis. This is not an absolute definition of the beam width, since side-lobes introduce energy outside this diameter, but it is an adequate definition for most purposes.

For a fixed focus, the narrowest part of the beam only extends over the length of the focal zone. The length of a focal zone is known as the depth of focus, and is given by

$$L_F = 2d_FN = 4\lambda N^2 \tag{4}$$

In order to get a thin beam over the full depth of the image field one must use several focal settings, end to end. With dynamic focusing the foci should be so close that their focal zones overlap.

Thus, both lateral resolution and range resolution are proportional to λ, i.e. inversely proportional to the frequency. Hence, to get the best possible range as well as lateral resolution, one should use as high an ultrasound frequency (short wavelength) as possible. However, attenuation in tissue increases with frequency, and one must therefore keep the frequency low enough to get adequate penetration. The variation of frequency with depth introduced by the confocal imaging technique thus optimizes the resolution/penetration balance for all depths.

3.2. Focusing perpendicular to the scan plane

A valuable feature of the annular array is that the beam is symmetrically focused, i.e. the focusing normal to the scan plane is the same as that in the scan plane (Fig. 7a). For linear arrays, electronic control of focusing is only achieved in the scan plane (Fig. 7b), i.e. beam width improvements are restricted to the so-called azimuth direction. Normal to the scan plane (the elevation

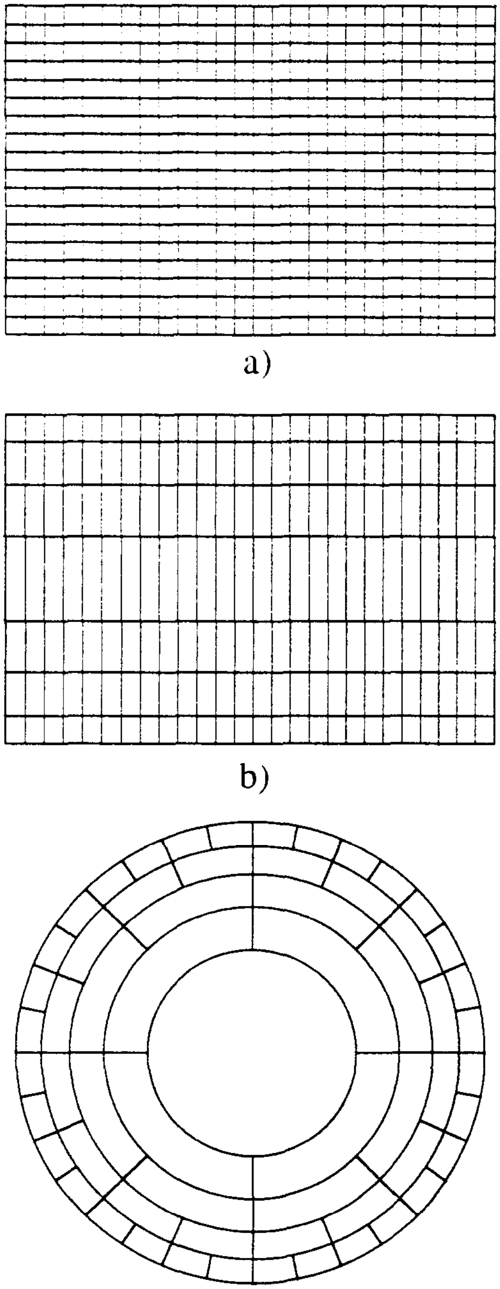

Fig. 8. (a) A 2D array of small elements could steer a beam in 3D and provide focusing in elevation, but the number of elements poses enormous difficulties. (b) A '1.5D' array permits focusing in elevation with a manageable number of elements. Correction of phase aberrations may also be achieved with (a) and (b) but an r-θ annular array (c) could do this with fewer elements.

direction) the focus is fixed. Hence, beams have a fixed focal length F_E in the elevation direction, and an electrically variable focal length F_A in the azimuth direction. The slice thickness of an annular array scanner is therefore less than that of a switched or phased array.

For a rectangular transducer, Eq. (3) can also be used to define the beam width, with slight modifications. For a constant amplitude weight across the array, the formula defines the width of the focus to the first zero of the intensity, while with apodization the formula defines the width where the beam has fallen to a certain intensity, as in the case of the circular aperture. The focal widths d_E and d_A in the elevation and azimuth directions, respectively, are then

$$d_E = 2F\lambda/D_E = 2\lambda N_E \quad (5a)$$

$$d_A = 2F\lambda/D_A = 2\lambda N_A \quad (5b)$$

where D_E is the elevation width of the array normal to the scan plane, and D_A is the width of the active part of the array in the scan plane. The depth of the focus in the two directions is given by Eq. (4), substituting the respective f-numbers N_E and N_A. To achieve adequate depth of focus for the fixed focus in the elevation direction, one must use a high f-number (typically $N_E = 5$), i.e. a low elevation aperture, which means there is a fairly wide beam in the elevation direction. The linear array elements can also be divided in the elevation direction to form a two-dimensional array so that one can obtain focusing and possible beam steering perpendicular to the scan plane, as illustrated in Fig. 8a. Focusing in the elevation direction requires only a coarse division of the elements in this direction (e.g. into just 7 elements), as illustrated in Fig. 8b. This is often called a 1.5D array, in contrast to a full 2D array which also allows beam steering in the elevation direction. Steering of the beam requires a finer division, into say 48–128 elements in the elevation direction.

Focusing requires equal delays for the elements that are symmetrically placed around the scan plane. These elements can therefore be connected together directly at the transducer, and hence only 4 delay channels are required to steer the 7 rows of elements. A 1.5D linear array can therefore be obtained with between $4 \times 64 = 256$ and $4 \times 128 = 512$ individual delay channels. This is within reach with present technology, and it is expected that such instruments will appear on the market in the future.

3.3. Grating lobes and high frequency imaging

Arrays of regularly spaced elements produce repetitions of the main lobe (grating lobes) in other azimuth directions if the distance between the centres of the elements is greater than $\lambda/2$. This is particularly relevant to phased arrays since their elements are small enough to radiate energy over a wide range of angles. The smallest element spacing that can be achieved with present day transducer technology is around 0.1 mm, limiting the largest wavelength to 0.2 mm. Phased array probes are thus currently limited to maximum frequencies of around 5–7 MHz. In switched linear arrays, the element spacing exceeds $\lambda/2$, but the elements are directional themselves, and transmit reduced energy in the direction of the grating lobes. Typical distances between the centres of the elements for the switched linear and curvilinear arrays are λ–2λ. This makes it possible to make switched linear array transducers with frequencies up to 20 MHz. Steering of the beam at an angle for Doppler measurements, as illustrated in Fig. 2b, requires smaller elements. This problem is generally solved by limiting the angle of beam deflection and using a lower frequency for the steered Doppler beams than for the unsteered imaging beams.

Most designs of annular array do not suffer from grating lobes and for this reason may have even larger element widths than the switched linear arrays. This permits the use of even higher frequencies.

3.4. Sidelobes, reverberations, phase aberrations and acoustic noise

Sidelobes are skirts of ultrasound energy around the main lobe. These induce the transducer to signals from targets that are outside the main lobe, and these spurious signals appear as a background noise in the image. A similar type of noise is caused by reverberations, which are ultrasound pulses that bounce back and forth between strong targets and the transducer, or simply between strong targets.

The ultrasound velocity varies between different types of tissue, and the difference is largest between muscular tissue and fat. The oblique and sometimes irregular shape of the boundaries between different types of tissue causes the wave front to be distorted by refraction, producing the phase aberrations mentioned in the introduction. Phase aberrations impair the focusing of the beam and increase the side-lobe level. This type of beam distortion, together with reverberations, is responsible for the increase in background noise which occurs in difficult obese patients. The noise is produced by the transmitted ultrasonic pulse and its echoes and is therefore called acoustic noise in contrast to electronic noise that arises in the receiver amplifier. As the acoustic noise is proportional to the transmitted power, the signal to noise ratio cannot be improved by increasing the transmit power, as is the case for the electronic noise.

Acoustic noise limits the ability to detect strong and weak targets close to each other, i.e. the contrast resolution of the imaging system. Reverberations have been reduced by improvements in transducer technology reducing reflections at the transducer face, but they still limit the contrast resolution of scanners today. There is a potential for further reduction of reverberation noise by processing the received ultrasound RF signal. The phase aberrations from smooth interfaces between tissues can be compensated for by using a two-dimensional array with additional corrective delays in each individual element channel (O'Donnell and Flax 1988; Nock et al. 1989; Gambetti and Foster 1993; Liu and Waag 1993). However, methods for estimating the correction delays are computer intensive, with limited applicability today.

3.5. Requirements for Doppler blood velocity measurements and colour flow imaging

In deep Doppler measurements, for example in the heart and the large thoracic and abdominal vessels, the major challenge is to obtain enough sensitivity to register the weak signals from the blood. The sensitivity to weak signals of the annular array has proved to be superior to that of other arrays for practical reasons. The elements are larger and fewer, which makes it easier to avoid cross-talk between the elements that otherwise limits the sensitivity of the CW Doppler. With the larger elements one can use lighter backings for mechanical support, which reduces losses to the transducer backing.

Two-dimensional imaging of blood velocities, known among others as colour flow mapping (CFM), is obtained by measuring the blood velocity profile along the beam with a pulsed wave (PW) multiple range-gate Doppler unit, and sweeping the beam direction laterally. For each range, the signal is similar to a conventional PW Doppler signal, so that the sensitivity of a blood velocity imaging system can be made comparable to the single gate PW and CW Doppler. The resolution Δf in the frequency determination is inversely proportional to the time T of observation, i.e.

$$\Delta f = 1/T \tag{6}$$

This introduces a time conflict for CFM compared to B-mode tissue imaging and stationary beam Doppler measurement. For tissue imaging it is sufficient to transmit only one pulse in each beam direction, while for CFM one must transmit several pulses (typically 4–16) in each direction to obtain adequate frequency resolution. This reduces the frame rate of CFM compared to B-mode tissue imaging. For CW and single gate PW Doppler measurements the beam direction is constant, so that there is much more time to make the frequency measurement.

In order to maintain a high enough frame rate with Doppler CFM, there is only sufficient time for a limited number of pulses in each direction. This reduces the ability to image low blood velocities, compared to what can be achieved with CW or single gate PW Doppler. In veins and some other low velocity flows, it is a challenge to discriminate very low blood velocities (1–2 cm/s) from movements in the tissue and the scan-head.

To measure low blood velocities, one must be able to discriminate between Doppler signals from slowly moving tissue, such as a wall of a heart chamber or a blood vessel, and the faster moving blood by means of a high-pass filter. This requires a small Δf which, from Eq. (6), requires a longer measurement time T for each beam direction than is required, for example, to measure the higher velocities through the cardiac valves.

For shallow ranges, one can increase the observation time in each direction by interleaving transmission pulses along different beam positions. It takes a short time to collect the data for shallow ranges (55 μs for 4 cm depth), and therefore there is sufficient time between transmitting successive pulses at one beam position to transmit pulses at other beam positions. This permits a long observation time at each beam position, without reducing the frame rate.

4. Requirements for different clinical applications

Ultrasound has developed as a local region imaging modality, where the probe is placed as close to the object as possible. Small probes that can be inserted into body cavities, blood vessels and surgical wounds, have made possible many different

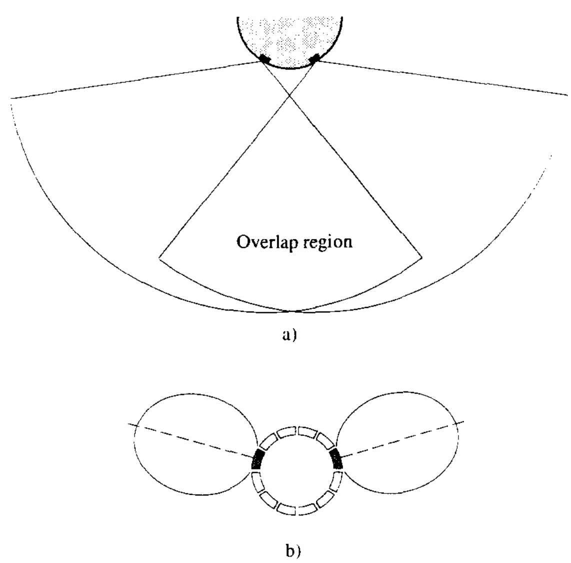

Fig. 9. The limited overlap of beams in tightly curved (a) and cylindrical (b) switched arrays limits the aperture that can be used for beam forming.

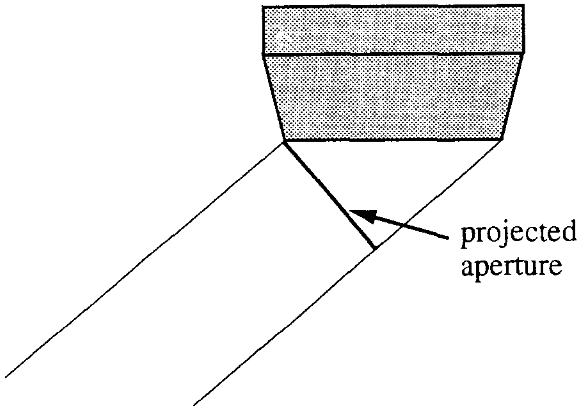

Fig. 10. The projected aperture reduces and hence the beam width increases as beam deflection increases with a phased array.

single 'best' solution. One can consider the different array types according to the scan features that are important for different applications.

4.1. Small footprint

Important for applications including: Inter-rib imaging of the heart and the liver. Endocavity imaging: e.g. trans-esophageal imaging of the heart and the aorta, intra-vascular imaging, gastrointestinal imaging, trans-rectal imaging, trans-vaginal and intrauterine imaging, imaging of the urinary tract, etc.

The small footprints of phased arrays and annular arrays are best suited for these types of applications. Tightly curved switched arrays are also interesting for these applications, but are less efficient since the directional elements transmit their energy in different directions due to the curvature of the array. The usable aperture is limited, therefore, to those elements whose beams have a good overlap, as illustrated in Figs. 9a and 9b. Phased arrays also have sensitivity and resolution problems at the outer edges of the image sector, since the effective aperture of the array is the projection of the array surface in the beam direction, as illustrated in Fig. 10. Both the phased and annular arrays can be rotated to produce 3D imaging from the esophagus and other cavities, as illustrated in Fig. 6b.

4.2. Wide near image field

Important for applications including: Regions where there is wide access to the target through soft tissue, e.g. abdominal imaging, vascular imaging, thyroid imaging, imaging from within the surgical wound, etc.

Switched linear and slightly curved linear arrays are well suited for these applications. Linear arrays also have interesting applications in endocavity imaging if they are mounted in the long axis direction at the tip of an endoscope. Similarly,

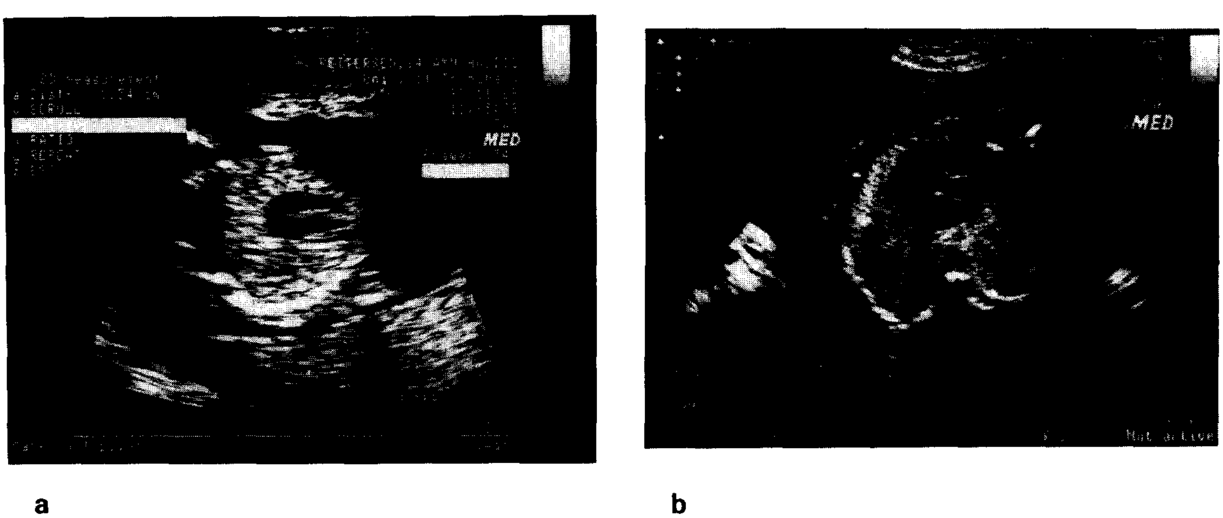

Fig. 11. Images of fetal hearts obtained with a 5 MHz annular array. (a) 14 week gestation, 15 mm overall length. (b) 25 week gestation, four chamber view.

they have been used for guidance during laparoscopic surgery, by inserting them through a trocar at the tip of a rod.

4.3. Narrow slice thickness

Important for applications including: Three-dimensional imaging and imaging of small objects like small cysts, the fetal heart, etc.

One might think that the narrow slice thickness achieved by symmetrical focusing (Fig. 7d) would be important in all applications, but in many diagnostic situations, the object has very slow variation normal to the scan plane. Typical examples are short-axis and long-axis imaging of the heart, imaging of the fetal trunk and head, short-axis imaging of a vessel, etc. In these situations the crispness of the image is principally determined by the focus in the scan plane, and focusing normal to the scan plane has less effect. However, it is important to have narrow slice thickness for 3D imaging and also for 2D imaging of objects that vary rapidly normal to the scan plane. A typical example is imaging of a small fetal heart which is shown in Fig. 11.

4.4. 3D imaging

Important for applications including: Complex structures like congenital heart lesions, regional deficiencies in myocardial contraction, movements of heart valves before surgery or repair, fetal imaging, imaging of the vascular tree around tumours at surgery, etc.

3D imaging has been proposed from the earliest days of ultrasound, but has fundamental problems in the visualization of the 3D data. Modern computer technology has provided the opportunity for interesting visualization at acceptable cost, leading to the introduction of experimental systems to the market.

Another problem with 3D imaging is that it takes time to collect the data. For measurements down to 15 cm depth it takes 200 μs per beam, and for static objects one can then collect data within a 90° × 90° sector in 2–3 s for these depths. The data collection time is increased proportionally with depth. With moving objects, like the heart, time is a fourth dimension which makes the data set 4-dimensional. To collect data from such an object, one has to rely on rhythmic movements so that the spatial data can be collected synchronously over several cycles. With the heart one can then collect the 4D data in a few minutes, depending on how much the heart rate varies.

The full 2D array is the ultimate technology for 3D and 4D imaging, but for the time being it is practical to collect 3D data by moving a 2D scan plane, for example by rotation, tilting, or linear translation. It is highly desirable to have a thin symmetrically focused beam, and the annular array is at present the most practical solution, but one should expect that 1.5D arrays will become available for symmetrical focusing.

4.5. High frequency (7–30 MHz)

Important for applications including: Peripheral vascular imaging, surgical imaging, neonatal imaging, intra-vascular imaging, eye imaging, etc.

It is difficult to make phased arrays at these frequencies owing to the problem of grating lobes discussed earlier. Linear arrays are suitable for high frequency imaging where there is sufficient access. Curvilinear arrays suffer from less efficient apertures due to the limited overlap of the beams from individual elements, discussed in 4.1. and illustrated in Fig. 9a. Annular arrays are very attractive for intra-cavity and surgical imaging, since they provide a large aperture within the access space available and provide a very thin beam throughout the image field, as illustrated in Fig. 6c.

4.6. Highly sensitive and steerable CW Doppler

Important for applications including: Cardiac imaging and Doppler measurements in large vessels.

Phased arrays and annular arrays are best suited for applications requiring beam steering. The annular array has produced the best sensitivity for steerable CW Doppler measurements in the heart, because it has few and large elements.

4.7. Ability to measure low blood velocities

Important for applications including: Peripheral vascular and venous imaging.

Switched linear and curvilinear arrays combine reasonable sensitivity with the ability to interleave transmissions at more than one beam position simultaneously, thus overcoming the frame rate penalty otherwise associated with measuring low velocities, as discussed in section 3.5.

5. Trends

5.1. Electronics and beam forming

The evolution towards high speed and high resolution analogue to digital converters (ADCs) is opening up new possibilities for digital beam forming. Experiments with digital beam forming started more than 10 years ago, and some commercial instruments are using such methods. However, due to the limited number of bits available in cost effective ADCs, the dynamic range of these beam formers has been limited. This has produced weak sensitivity in deep and CW Doppler measurements of blood velocity. New ADCs are in development which will change this picture and allow the use of precise and reproducible digital technology in wide dynamic range beam formers. Digital beam formers also facilitate the use of multiple beams with linear arrays, allowing higher frame rates and sensitivity to be achieved without compromising image resolution and ability to measure low velocity. They also facilitate optimization of frequency and focusing for different applications.

Digital storage of the RF echoes from a single element gives interesting opportunities for correcting phase front aberrations. Digital processing already at the RF level also introduces opportunities for reduction of the acoustic noise, which still limits the applicability of ultrasound imaging in difficult patients.

5.2. Arrays and scanning strategies

The two-dimensional array, as illustrated in Fig. 8a, is the futuristic ideal for beam forming developments. With this array one can steer the beam in all directions of a 3D segment, and the beam can be focused symmetrically at all depths (Smith et al. 1991). As the present phased array scanners typically use 64–128 elements, the 2D array should use from 64 × 64 to 128 × 128, which is 4096 to 16 384 elements and delay channels. A significant problem with so many channels is the connection of cables between each transducer element and the electronics, and for such arrays a large amount of electronics must be integrated on the transducer array. The high electrical impedance of the individual elements is another problem. This means that we will not see an operational 2D array in the near future.

Electronic steering of the beam in three dimensions has been achieved with two separate 1D phased arrays mounted in a 90° cross (Thurnbull and Foster 1991). The net beam is the product of the transmitted and the received beams, and since transmitted and received beams are rotated 90° relative to each other, the two-way transmit and receive side-lobes are reduced. A problem with this array is that both the transmitted and received beams spread out widely in their respective elevation directions, thus reducing the sensitivity and increasing acoustic noise due to higher side-lobe level.

The 1.5 D array for symmetric focusing of the beam is the most obvious extension of present array technology, as discussed above. Scanning in the elevation direction must be done by a mirror or by moving the array.

For focusing with the 1.5D array, one can make use of symmetry around the scan plane to reduce the number of delay channels. For phase aberration correction one needs to delay the signals from each side of the azimuth plane individually, which means a 7-layer 1.5D array, which needs in total 7 × 64 (= 448) to 7 × 128 (= 896) individual delay channels. This requires a large number of cables with complex connections to each element, presenting technical challenges that have not yet been solved.

The two-dimensional modification of the annular array, the r-θ array shown in Fig. 8c, then gives the possibility of phase aberration correction with much fewer delay channels. With 8 radial elements, and a number of angular elements per radial element of 1, 4, 8, 12, 16, 24, 32 and 48, there are a total of 145 individual delay channels. Cable connection to this number of elements is within reach of today's transducer technology, but it will be difficult to move the array for scanning. Beam steering is in this case best done with a mirror. Given the development of sufficiently efficient phase aberration correction algorithms, one can see such an array applied to wide aperture abdominal imaging.

6. Conclusion

From the above discussion it is clear that the different types of arrays have advantages and drawbacks, depending on the application. Hence there is no 'best' array. With new electronic developments one can make beam formers that can handle all types of arrays, so that the type of array optimal for the particular application can be used.

For abdominal and fetal imaging, a switched curvilinear array offers a wide field of view, but it would be a clear advantage to be able to switch to an annular array with its symmetrical beam to image small objects like the fetal heart. Three-dimensional imaging also benefits from the narrow symmetrical beam of the annular array.

In cardiology, while phased arrays with multiple beams can provide very high frame rates, annular arrays provide the greater sensitivity that is important for Doppler measurements. Annular arrays also offer the greatest efficiency at the higher frequencies necessary for imaging of the neonatal heart.

Summarizing the differences between the types of arrays, annular arrays produce symmetrical focusing giving thinner 2D image slices at all depths and high lateral resolution in all directions for 3D imaging. The dome covering the annular array can sometimes facilitate acoustic contact when the beam is steered in different directions within body cavities as, for example, in the esophagus. They also have advantages at the higher frequencies (> 7.5 MHz), and will therefore have special applications in neonatal medicine, endocavity imaging, and for guidance of surgical procedures. r-θ annular arrays have the capability of doing phase front aberration correction with a manageable number of elements and delay channels, although the larger number of leads will probably restrict beam scanning flexibility.

Linear and curvilinear arrays have the advantage of very rapidly switching the beam position, which minimizes the reduction in frame rate otherwise associated with multiple zone focusing or confocal imaging, and with colour flow imaging of very low blood velocities at shallow ranges.

Phased arrays also allow rapid beam switching and have a small footprint which may or may not permit better access between the ribs than does the dome of an annular array, depending on the patient's build and the target. In addition, the flat face of the phased linear array produces contact problems when positioned at an angle to the skin.

Linear 1.5D arrays will produce narrower slice thicknesses with full electronic control of the focus in both elevation and azimuth. This will give the greatest flexibility in beam focusing and interleaving of beams and pulses that one can see in the future.

References

Angelsen BAJ, Hoem J, Dørum S, Brubakk AO, Skjaerpe T, Torp HG, Olstad B, Maehle J. Annular array multiplanar TEE probe allowing 3D reconstruction of cineloops of the heart. In Hanrath P, Uebis R, Krebs W, eds. Cardiovascular Imaging by Ultrasound: Kluwer Ac Publ 1993a; 287–296.

Angelsen BAJ, Torp H et al. Color Flow Mapping with an Annular Array Mechanical Scanner. Cardiac Ultrasound. Edinburgh, Churchill Livingstone, 1993b; 53–62.

Bom N, Lancée CT et al. Multiscan Echocardiography I. Technical description. Circulation 1973; 68: 1066–1074.

Burckhardt CB, Grandchamp P-A et al. Focussing ultrasound over a large depth with an annular transducer — An alternative method. IEEE Trans Sonics Ultrasonics 1975; 22(1): 11–15.

Gambetti C, Foster FS. Correction of phase abberations for sectored annular array ultrasound transducers. Proc IEEE Ultrasonics Symp 1993; 106.

Green Ph S. Real-time, high resolution, ultrasonic arteriography system. Proceedings of Symposium on Non-invasive Diagnostic Techniques in Vascular Disease, San Diego, March 14–18, 1977.

Liu D-L, Waag RC. Wavefront back-propagation and focus correction. Proc IEEE Ultrasonics Symp 1993; 106.

Macovski A. Ultrasonic imaging using arrays. Proc IEEE 1979; 67; 194–195.

Nock L, Trahey GE et al. Phase aberration correction in medical ultrasound using speckle brightness as a quality factor. J Acoust Soc Am 1989; 85 (5): 1819–1833.

O'Donnell M, Flax SW. Phase-aberration correction using signals from point reflectors and diffuse scatterers: Measurements. IEEE Transact Ultrason, Ferroelect, Freq Contr 1988; 35(6): 768–773.

Pandian NG, Hsu T-L, Schwartz SL, Weitraub A, Cao Q-L, Schneider AT, Gordon G, England M, Simoeti J. Multiplane transesophogeal echocardiography. Echocardiography 1992; 6: 9.

Smith SW, Pavy Jr HG, von Ramm OT. High speed ultrasound volumetric imaging system — Part 1: Transducer design and beam steering. IEEE Trans UFFC 1991; 38; 100.

Somer JC. Electronic sector scanning for ultrasonic diagnosis. Ultrasonics 1968a; 6: 153.

Somer JC. Instantaneous and continuous pictures obtained by a new two-dimensional scan technique with a stationary transducer. Proceedings in echoencephalography. Berlin, Springer Verlag, 1968b; 234.

Somer JC. Principles of phased-array imaging. Cardiac Ultrasound. Edinburgh, Churchill Livingstone, 1993.

Takeuchi H, Mazuzawa H, Nakaya C, Ito Y. Medical ultrasonic probe using electrostrictive ceramic/polymer composite. Proc IEEE Ultrasonics Symp 1989; 705–708.

Thurnbull DH, Foster FS. Beam steering with pulsed two-dimensional transducer arrays. IEEE Trans UFFC 1991; 38; 320–333.

Wells PNT. Ultrasonics: a window into biomedical science. Ultrasonics 1992; 30; 3–7.