

Ultrasound System Considerations and their Impact on Front-End Components

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Introduction

Medical ultrasound machines are some of the most sophisticated signal processing machines in use today. As in any machine there are many trade-offs in implementations due to performance requirements, physics, and cost. This paper will try to show the trade-offs for ultrasound front-end circuits by starting from a high-level system overview followed by a more detailed description of how ultrasound systems work. Some system level understanding is necessary to fully appreciate the desired front-end integrated circuit (IC) functions and performance, especially for: the Low Noise Amplifier (LNA); Time Gain Control (TGC); and A/D Converters (ADC). “Front-end” is defined in this paper as all circuitry including the beamformers, even though the primary focus is on the analog signal processing components up to and including the ADCs.

The main motivation for the author to write this paper was that most books and publications focus on the system level aspects of ultrasound since they are written by system designers, but those don’t explain what the effects of parameters in the front-end circuit components are on diagnostic performance. For a semiconductor company to provide optimal components though, it is essential for the IC designer to know what specifications are of particular importance and what effect they will have. Additionally there are trade-offs in terms of integration and semiconductor process technology that will force certain choices; these in turn are important for the ultrasound system designer to know so that he can achieve the most advantageous system partitioning.

System Introduction

In ultrasound front-ends, as in many other sophisticated electronic systems, the analog signal processing components are key in determining the overall system performance. The front-end components define the *bottleneck* to system performance; once noise and distortion have been introduced it is essentially impossible to remove them. This is, of course, a general problem in any receive signal processing chain, be it ultrasound or wireless. It is interesting to note that ***ultrasound is really very similar to a radar or sonar system*** - radar works in the GHz range, sonar in the kHz range, and ultrasound in the MHz range - but the system principals are essentially the same. Actually, an advanced ultrasound system is practically identical to a Synthetic Array Radar (SAR). Originally the ‘phased array’ idea of steerable beams has been conceived by radar designers, however, ultrasound designers expanded on the principle and today those systems are some of the most sophisticated signal processing equipment around.

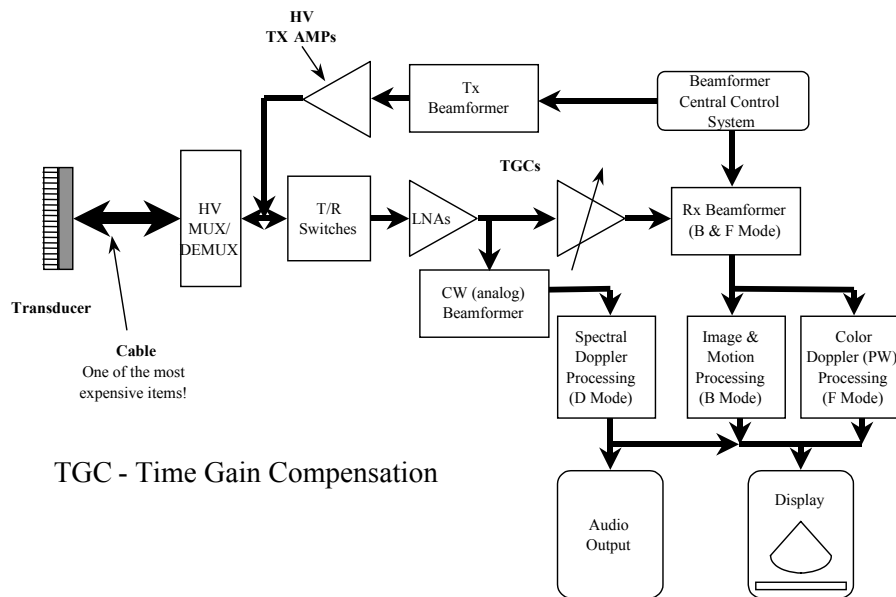


Figure 1. Ultrasound System Block Diagram

Figure 1 shows a simplified diagram of an ultrasound system. In all systems there is a transducer at the end of a relatively long cable (ca. 2m). This cable has from a minimum of 48 up to 256 micro-coaxial cables and is one of the most expensive parts of the system; today there is ongoing research into 3-D ultrasound for which even 512 cables and higher are being investigated. In practically all systems the transducer elements directly drive the cable which can result in significant signal loss due to the loading of the cable capacitance on the transducer elements; this in turn demands that the receiver noise figure (NF) is lower by the amount of the cable loss. One can expect a loss on the order of 1-3 dB depending on transducer and operating frequency. In most systems multiple probe heads can be connected to the system, this allows the operator to select the appropriate transducer for optimal imaging. The heads are selected via High Voltage (HV) relays; these relays introduce large parasitic capacitance in addition to the cable.

A HV Mux/Demux is used in some arrays to reduce the complexity of transmit and receive hardware at the expense of flexibility. The most flexible systems are Phased Array DBF systems where ALL transducer elements can be individually phase and amplitude controlled, these tend to be the most costly systems due to the need for full electronic control of all channels. However, today's state-of-the-art front-end ICs like the AD8332 (VGA) and the AD9238 (12b ADC) are pushing the cost-per-channel down continuously such that full electronic control of all elements is being introduced even in medium to low cost systems.

On the transmit side the Tx beamformer determines the delay pattern and pulse train that set the desired transmit focal point. The outputs of the beamformer are then amplified by high voltage transmit amplifiers that drive the transducers. These amplifiers might be controlled by DACs to shape the transmit pulses for better energy delivery to the transducer elements. Typically multiple transmit focal regions (zones) are used, i.e. the field to be imaged is divided up by focusing the transmit energy at progressively deeper points in the body. The main reason for doing this is to increase the transmit energy for points that are deeper in the body because the signal gets attenuated as it travels into the body.

On the receive side there is a T/R switch, generally a diode bridge, which blocks the high Tx voltage pulses, followed by a low noise amplifier and VGA(s) which implement the TGC and sometimes also apodization (spatial "windowing" to reduce sidelobes in beam) functions. TGC is under operator control and used to maintain image uniformity. After amplification, beamforming is performed which can be implemented in analog (ABF) or digital (DBF) form, but it is mostly digital in modern

systems except for continuous wave (CW) Doppler processing whose dynamic range is mostly too large to be processed through the same channel as the image. Finally, the Rx beams are processed to show either a gray scale image, color flow overlay on the 2-D image, and/or a Doppler output.

Ultrasound System Challenges

To fully understand the challenges in ultrasound and their impact on the front-end components it is important to remember what this imaging modality is trying to achieve. First, it is supposed to give an accurate representation of the internals of a human body, and second, through Doppler signal processing, it is to determine movement in the body as represented by blood flow, for example. From this information a doctor can then make conclusions about the correct functioning of a heart valve or blood vessel.

Acquisition Modes

There are three main ultrasonic acquisition modes: B-mode (Gray Scale Imaging; 2D); F-mode (Color Flow or Doppler Imaging; blood flow); and D-mode (Spectral Doppler). B-mode creates the traditional gray scale image; F-mode is a color overlay on the B-mode display that shows blood flow; D-mode is the Doppler display that might show blood flow velocities and their frequencies.

Operating frequencies for medical ultrasound are in the 1-40 MHz range, with external imaging machines typically using frequencies of 1-15 MHz, while intravenous cardiovascular machines use frequencies as high as 40 MHz. Higher frequencies are in principal more desirable since they provide higher resolution, but tissue attenuation limits how high the frequency can be for a given penetration distance. The speed of sound in the body (~ 1500 m/s) together with desired spatial resolution determines ultrasound frequencies. The wavelength range associated with the 1-15 MHz frequency range is 1.5mm to 100 μm and corresponds to a theoretical image resolution of 750 μm and 50 μm respectively. The wavelength, and consequently the frequency, directly determine how small of a structure can be distinctly resolved [12].

However, one cannot arbitrarily increase the ultrasound frequency to get finer resolution since the signal experiences an attenuation of about **1 dB/cm/MHz**, i.e. for a 10 MHz ultrasound signal and a penetration depth of 5 cm, the signal has been attenuated by $5 \times 2 \times 10 = 100$ dB! Add to this an instantaneous dynamic range at any location of about 60 dB and the total dynamic range required is 160 dB! Such a dynamic range of course is not achievable on a per channel basis but through use of TGC, channel summation, and filtering this can be achieved. Nevertheless, because transmit power is regulated the SNR is limited, and therefore one has to trade off something, either penetration depth or image resolution (use lower ultrasound frequency). Another big challenge is that bone has a very high attenuation of about 12 dB/cm at 1MHz; this is similar for air. This makes imaging of the heart difficult through the ribs and lungs; one needs to use the open spaces between the ribs.

The largest dynamic range of the received signal, due to the need to image near-field strong echoes with minimal distortion simultaneously with deep weak echoes, presents one of the most severe challenges. Very low noise and large signal handling capability are needed simultaneously of the front-end circuitry in particular the LNA; for anyone familiar with the demands of communications systems these requirements will sound very familiar. Cable mismatch and loss, directly add to the noise figure of the system. For example, if the loss of the cable at a particular frequency is 2 dB, then the NF is degraded by 2 dB. This means that the first amplifier after the cable will have to have a noise figure that is 2 dB lower than if one would have a loss-less cable. One potential way to get around this problem is to have an amplifier in the transducer handle, but there are serious size and power constraints, plus needed protection from the high voltage transmit pulses that make such a solution difficult to implement.

Another challenge is the large acoustic impedance mismatch between the transducer elements and the body. The acoustic impedance mismatch requires matching layers (analogous to RF circuits) to efficiently transmit energy. There are normally a couple of matching layers in front of the transducer elements, followed by a lens, followed by coupling gel, followed by the body. The gel is used to insure good acoustic contact since air is a very good acoustic reflector.

A further problem is the high Q of the transducer elements; they can ring for a long time once excited by a high voltage pulse, this necessitates damping to shorten the pulse duration. However, the time duration/length of the transmit pulse determines axial resolution: the longer the pulse, the lower the resolution. A big drawback of the damping is the loss of energy, which necessitates a higher voltage pulse for a given amount of energy.

Another important issue for the receive circuitry is **fast overload recovery**. Even though there is a T/R switch that is supposed to protect the receiver from large pulses, part of these pulses leak across the switches and can be large enough to overload the front-end circuitry. Poor overload recovery will make the receiver 'blind' until it recovers, this has a direct impact on how close to the surface of the skin an image can be generated. In the image this effect can be seen as a region at maximum intensity (white typically in a gray scale image).

Transmitter Types

There are three common transmitter types:

- Pulse
- Pulse Wave Doppler (**PW Doppler**)
- Continuous Wave Doppler (**CW Doppler**)

Pulse-type, i.e. a single 'spike' that is transmitted is theoretically ideal for B-mode imaging since it gives the best axial resolution, however, since the transducers have a bandpass response anyway it doesn't make much sense to transmit an 'impulse' but rather use a transmit pulse that is ideally optimally matched to the transducer element impulse response; this may be as simple as a single cycle of the carrier for simplicity and cost reasons. Using multiple periods of the transmit carrier is done in most systems today since it increases the amount of energy transmitted into the body and it allows the sharing of the Pulse and PW Doppler transmit modes. A few cycles of a sine or square wave are transmitted at a time and normally also windowed (Hamming, etc.) to reduce the Tx spectral bandwidth [12]. The 'longer' a pulse train is, the lower the Tx voltage can be for a given amount of energy transmitted; remember that energy is the area under the curve and there are regulations from organizations like the FDA in the USA on how much energy can be transmitted into the patient. This is another constraint that generally forces the ultrasound system manufacturers to try to make the receiver as sensitive as possible such that they can back off the transmit signal energy to the lowest possible level and still achieve a desired diagnostic capability.

PW and CW Doppler

- PW Doppler
 - Measures the Doppler (shift) frequency at a particular range location along the beam
 - Drawback: Highest Doppler shift is limited by pulse repetition rate (Ex.: 12 cm depth, 1540 m/s => max. pulse rate = 6.4 kHz => max. measurable frequency = 3.2 kHz because of Nyquist criteria)
- CW Doppler
 - Half the array is used for transmit, the other for receive
 - Can measure higher Doppler shifts -- BUT -- can't tell distance to scatterer
 - Need high instantaneous dynamic range

PW Doppler allows for frequency shift measurements at a particular location in the body, its main drawback is the limitation of the measurable Doppler shift frequency by the pulse repetition rate [11]. CW Doppler, in comparison, can measure higher Doppler frequency shifts, but can't locate where along the beam a particular frequency is coming from. Also, CW Doppler requires high instantaneous dynamic range since large reflections coming from surfaces close to the skin are simultaneously present with very small signals from deep within the body. Together PW and CW Doppler can be used to find the location and the frequency of a Doppler frequency shift.

In Doppler signal processing the receiver may be deliberately overloaded to extract the small Doppler signal from the large carrier. **In the Doppler modes a perfect limiter is desired**, i.e. the phase, and therefore Doppler, information is perfectly preserved even when the receive components are in overload. The worst possible distortion in this mode is Amplitude Modulation to Phase Modulation (AM/PM) conversion, for example, duty cycle variation of the carrier due to overload. Note the strong similarity to communications systems, for example, in a constant envelope digital communications system like GSM all the information is in the phase, and AM/PM conversion (also known as phase noise in an oscillator or jitter in a clock) can be a serious problem. A great way of measuring the receiver performance under overload is to input a sine wave and slowly increase the signal level until overload occurs, if the receiver components behave as ideal limiters under overload then one should only see the fundamental and its harmonics on a spectrum analyzer but **NO 'noise skirts' next to the fundamental and harmonics**. To understand this one needs to know that Doppler information shows up as sidebands next to the fundamental. For example, in an ultrasound Doppler measurement the carrier frequency may be 3.5 MHz yet the Doppler frequency information due to movement in the body are somewhere between a few Hz (breathing) and 50 kHz (blood jets in the heart), so close-in phase noise due to AM/PM conversion because of receiver overload can easily mask the very weak Doppler information.

How is an ultrasound image generated? – B-Mode

Figure 2 shows how the different scan images are generated, in all four scans the pictures with the scan lines bounded by a rectangle are an actual representation of the image as it will be seen on the display monitor. Mechanical motion of a single transducer is shown here to facilitate understanding of the image generation, but the same images can be generated with a linear array without mechanical motion. As an example, for the Linear Scan the transducer element is moved in a horizontal direction and for every scan line (the lines shown in the 'Images') a Tx pulse is sent and then the reflected signals from different depths recorded and scan converted to be shown on a video display. How the single transducer is moved during image acquisition determines the shape of the image. This directly translates into the shape of a linear array transducer, i.e. for the Linear scan, the array would be straight while for the Arc scan, the array would be concave.

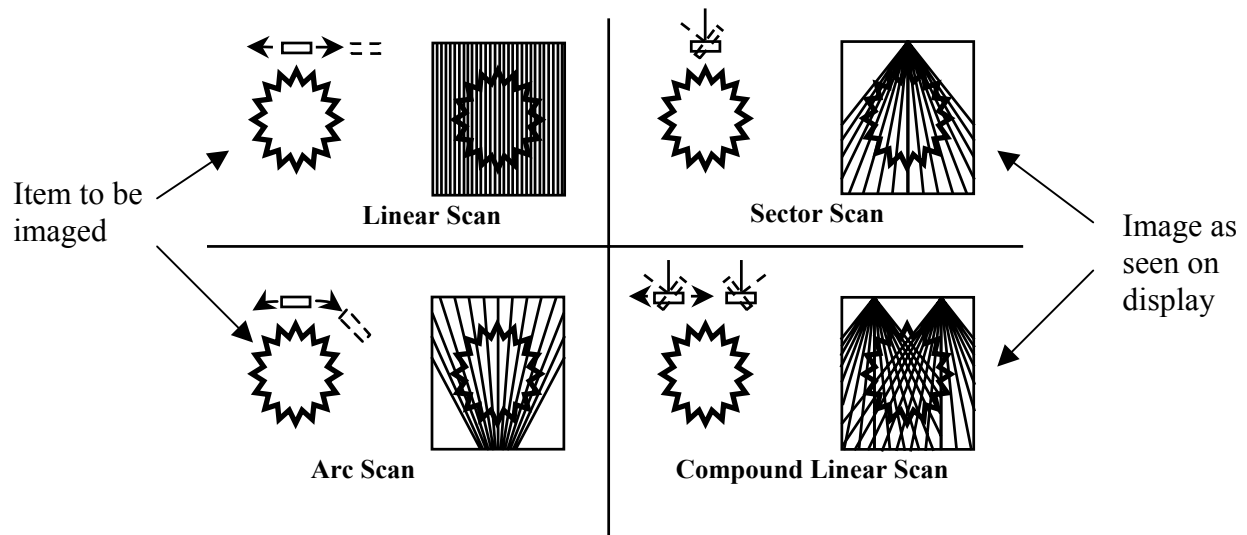


Figure 2. Single Transducer Image Generation [3]

The step that is needed to go from a mechanical single transducer system to an electronic system can also be most easily explained by examining the Linear Scan in Figure 2. If the single transducer element is divided into many small pieces, then if one excites one element at a time and records the reflections from the body, one also gets the rectangular image as shown, only now one doesn't need to move the transducer elements. From this it should also be fairly obvious that the Arc Scan can be made of a Linear Array that has a concave shape; the Sector Scan would be made of a Linear Array that has a convex shape.

Even though the example above explains the basics for B-mode ultrasound image generation, in a modern system more than one element at a time is used to generate a scan line because it allows for the aperture of the system to be changed. Changing the aperture, just like in optics, changes the location of the focal point and thereby helps in creating clearer images. Figure 3 shows how this is done for a Linear and Phased Array; the main difference is that in a Phased Array all elements are used simultaneously while in a Linear Array only a subset of the total array elements is used. The advantage in using a smaller number of elements is a savings in electronic hardware; the disadvantage is that it takes longer to image a given field of view. The same is not true in a Phased Array; because of its pie shape a very small transducer can image a large area in the far field. This is also the primary reason why Phased Array transducers are the transducers of choice in applications like cardiac imaging where one has to deal with the small spaces between the ribs through which the large heart needs to be imaged.

The linear stepped array on the left in Figure 3, will excite a group of elements, which is then stepped one element at a time, and each time one scan line (beam) is formed. In the phased array, all transducers are active at the same time. The direction of the scan line is determined by the delay profile of the pulses that are shown representatively by the 'squiggles' on the lines that lead to the array (blue). Time is as shown in Figure 3 and the darkened lines are the scan lines that are scanned for the representative pulsing patterns. If the pattern would have a linear phase taper in addition to the phase curvature, then the scan lines would be at an angle as is shown in Figure 4.

In the left part of Figure 4, three focal points are shown for three different time delay patterns on the individual elements; the 'flatter' the delay profile the farther the focal point is from the transducer element plane [5]. For simplicity reasons all delay patterns are shown over the full width of the array, however, typically the aperture is narrowed for focal points that are closer to the array plane like F(R3) for example. On the right half of Figure 4 one can see how linear phase tapers introduce beam steering; the curvature on top of the linear phase taper does the focusing along the beams.

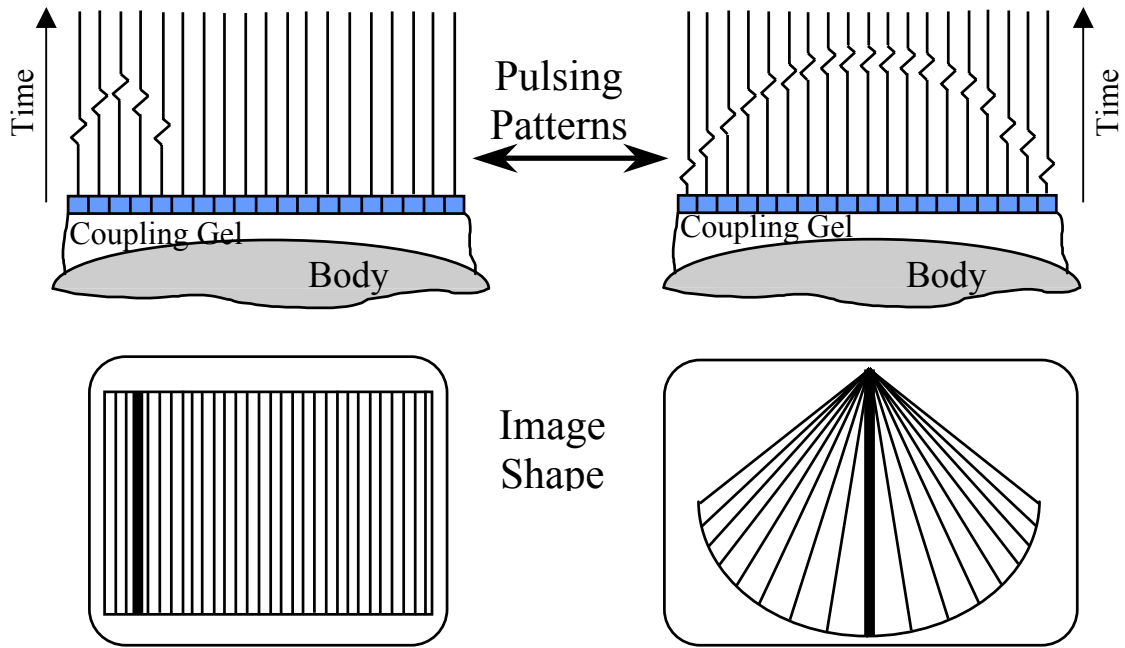


Figure 3. Linear vs. Phased Array Imaging [7]

Note that delay shapes are determined either by an analog delay line or digital storage (digital delay line). The shapes determine where the focal point will be. To increase the resolution on a scan line, one needs more taps on an analog delay line or equivalently more digital storage resolution in the delay memory. There are two ways to achieve higher resolution, the brute force way is to use higher sampling speeds on the ADCs and increase the digital storage or correspondingly the number of taps along the delay line. Too get the needed resolution though, this would require excessively high sampling speeds which result in unacceptably high power consumption if it can be done at all with today's ADC technology. Therefore, in a realistic system today, the ADC sample rate is determined by the highest frequency to be imaged, and interpolation (upsampling) or phasing of the channels is used to provide the necessary resolution for good beamforming.

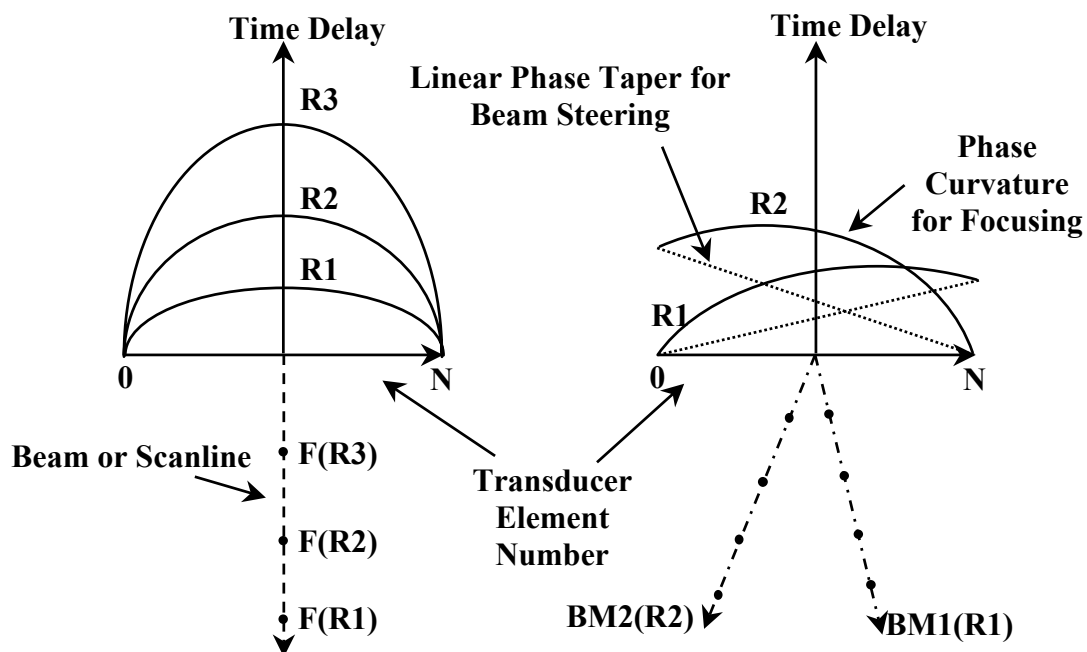


Figure 4. Phased Array Beam Steering [5]

Figures 5, 6, and 7 attempt to help visualize what has been explained above. Figure 5 shows the key terms for a focused beam. The transducer array apertures determine the resolution in the lateral and elevation planes; the elevation aperture is fixed because of element height for a given 1-D transducer while the lateral aperture can be varied dynamically. However, a lens in front of the transducer can influence the elevation aperture. Axial resolution, perpendicular to the transducer face, is set by the pulse duration - the shorter the pulse the higher the axial resolution.

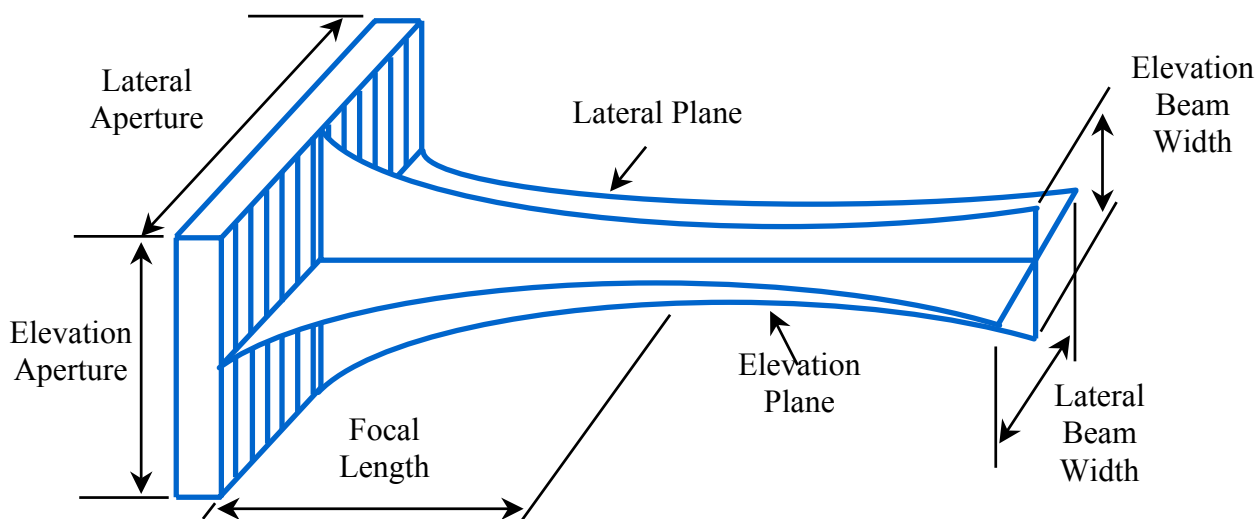


Figure 5. Focused Beam [7]

Figures 6 and 7 show in a more plastic way what has been revealed in Figure 3. Note how the focal plane is perpendicular to the array in the Linear Array, while it is on a curved surface for the phased array. In the linear array the beam is moved back and forth by selecting the group of elements via a switch matrix and stepping them one transducer at a time. For a given delay pattern across the active group of transducer elements, the focal plane stays fixed. In a phased array, the focal plane lies along a circular arc, it should be obvious now why the phased array will only generate a sector scan image. It is important to reiterate that **all elements are active in a phased array**, and therefore such a system typically needs more hardware than a linear array. That being said for explanation purposes of the principals of a linear stepped and phased array, it should be pointed out that in high end modern systems a “linear array” with, for example, 192 or more elements might use a 128 element “phased array” sub-section. In that case a **compound linear scan** as seen in Figure Figure 2 is generated and doesn’t look like the typical fan shape.

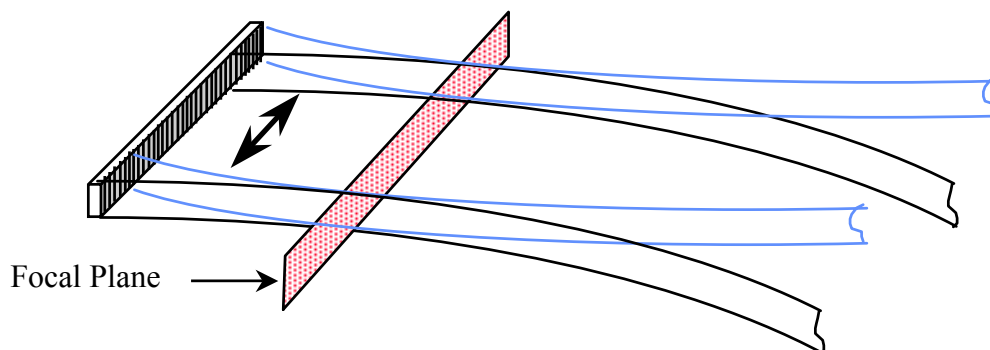


Figure 6. Linear (Stepped) Array Scanning [3]

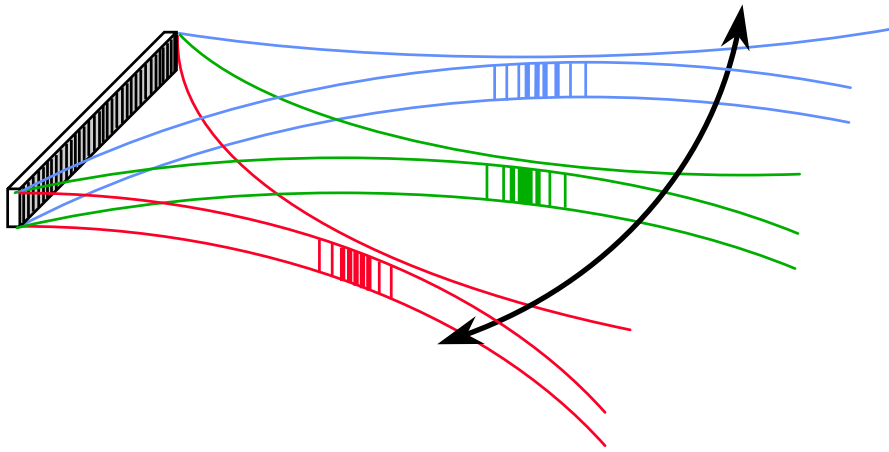


Figure 7. Phased Array Scanning [3]

What is Apodization?

An ultrasound transducer introduces a sampled region in space (aperture). If all of the elements in the transducer are excited simultaneously with equal pulses, then a spatial rectangular window will be generated. This produces a spatial $\sin(x)/x$ response as shown in Figure 8. To reduce the sidelobes of this response, the pulses are shaped by a **tapering or apodization function** like a Hamming, Hanning, Cosine, etc. The main reason for doing this is to concentrate all the energy in the central lobe and thereby increase the directivity of the transducer. The drawback is that the main lobe becomes wider with consequent reduction in lateral resolution.

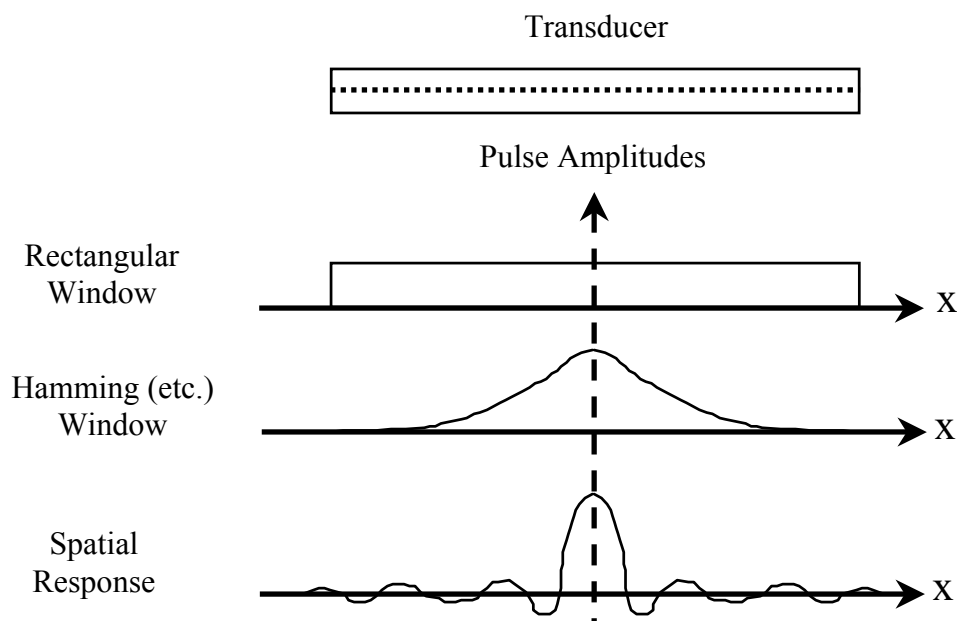


Figure 8. Apodization

Localization

There are two types of localization: (1) **Object Localization** (Fig. 9) and (2) **Axial Localization** (Fig. 10). A single transducer element cannot resolve two objects that are an equal distance away from the element. The reason for this is that the ultrasound waves propagate just like water waves and therefore reflections from O1 and O2 in Figure 9 will arrive at the same time at T1; T1 can't distinguish O1 and O2. One needs at least two elements to get both range and angle (azimuth) information [10]. As the number of elements increases the position of each object becomes better defined; i.e. the resolution increases. Although images can be generated by activating one transducer

element at a time to form the individual scan lines of the image, poor resolution and sensitivity plus long acquisition time make such a system not very viable. That's why small groups of elements are connected, and then moved one element at a time. By using a group of elements, the active area of the transducer is increased which gives better sensitivity and resolution in the far field where the beam starts to diverge.

The far field starts at:

$$x = \frac{d^2}{4\lambda}$$

Where 'd' is the width of the aperture, and 'λ' is the wavelength of the ultrasonic wave [3][7].

Using a group of elements, however, is detrimental to resolution in the near field since the beam gets wider. This is one of the reasons why the aperture gets narrowed when generating images of locations close to the transducer array.

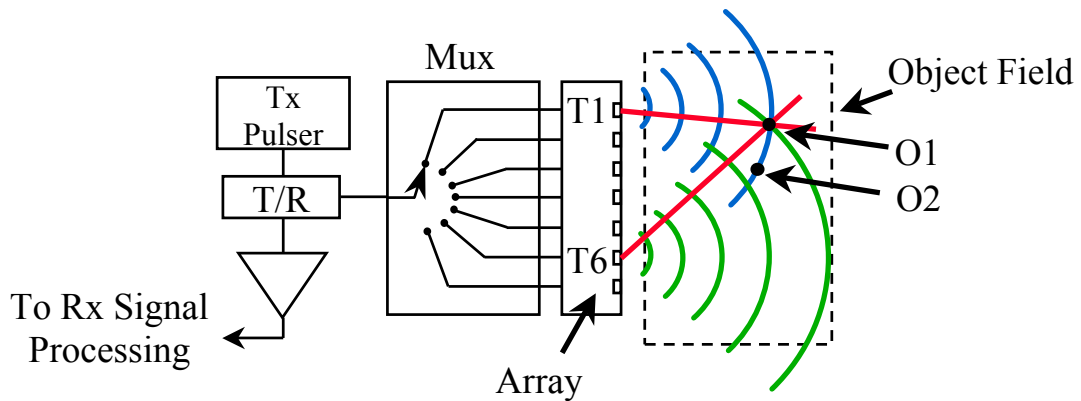


Figure 9. Object Localization

Figure 10 shows a hypothetical transmitted PW signal and received echo for a single transducer element. During transmit, the burst is repeated every $1/\text{PRF}$ seconds, the Pulse Repetition Rate. As soon as the transmit burst is over, the same element can be used to listen to the echoes. At first, the echoes will be very strong and then rapidly diminish as the time-of-flight increases. For example, the first echo might be from a blood vessel close to the surface of the skin, while the last echo might be from a kidney. By gating the receive signal, one can select where along the beam one wants to evaluate the signal. For a given returning pulse the time of flight is $2 \cdot t_1$ and the Rx gating window is $t_2 - t_1$.

As pointed out earlier, the Pulse Repetition Frequency (PRF) limits the maximum Doppler frequency shift that can be measured. But at the same time, the PRF together with the carrier frequency also determines how deep one can look in the body. Most ultrasound OEMs are using multiple pulses in flight at once (high PRF) to help increase the measurable Doppler frequency shift, but now one has to resolve multiple echoes and it becomes more difficult to determine where an echo comes from. Furthermore another problem emerges with high PRF. Since the time of return of an echo is random a transmit pulse might mask a received echo, this is called **eclipsing** [8].

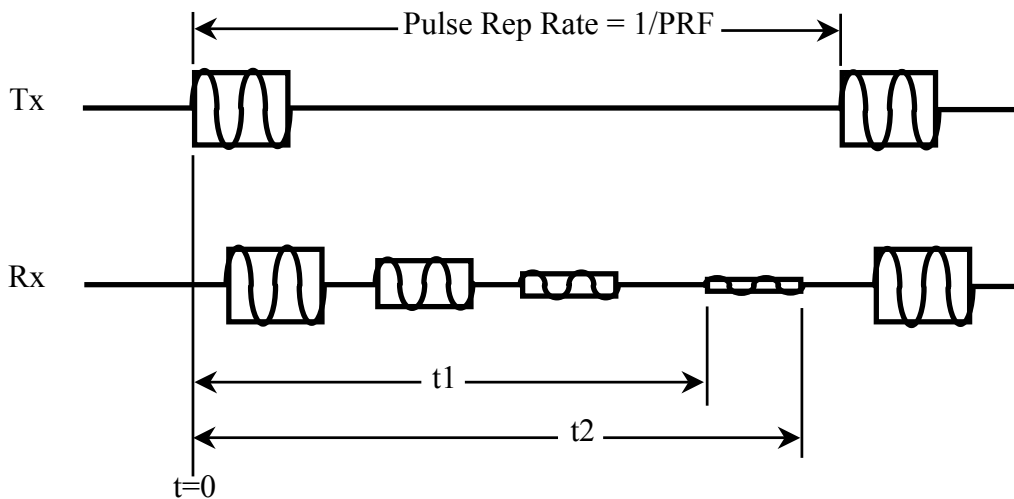


Figure 10. Echo Localization along Beam

Figure 10 also shows why a VGA is needed in the front-end of an ultrasound system; the VGA compensates the attenuation versus time of the reflected pulses. This is called Time or Depth Gain Control – TGC or DGC – and often ultrasound engineers will refer to the **TGC amplifier** that is just a linear-in-dB or exponential VGA. The need for a linear-in-dB VGA for TGC is that the signal is attenuated logarithmically as it propagates in the body; an exponential amplifier inverts this function.

Analog vs. Digital Beamforming

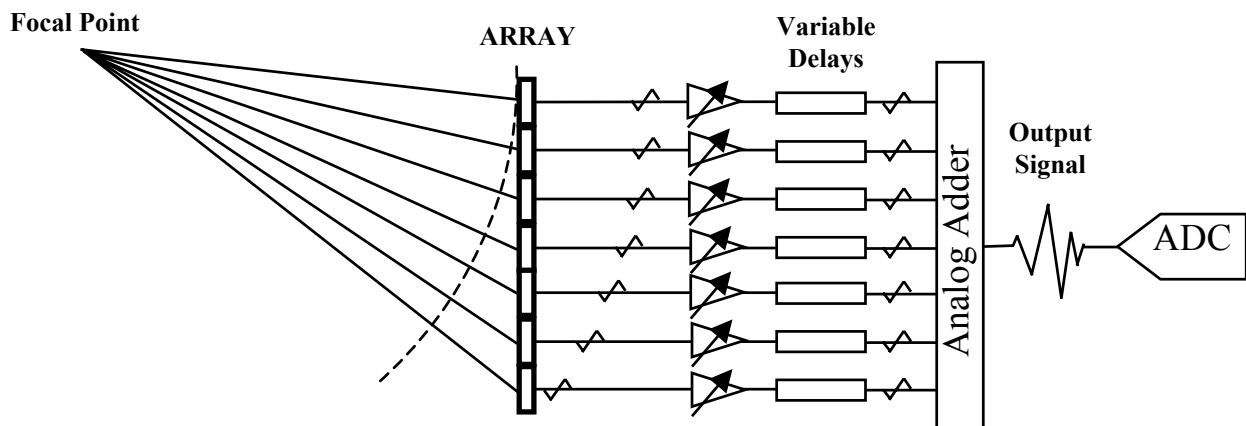


Figure 11. Simple Block Diagram of ABF System [4]

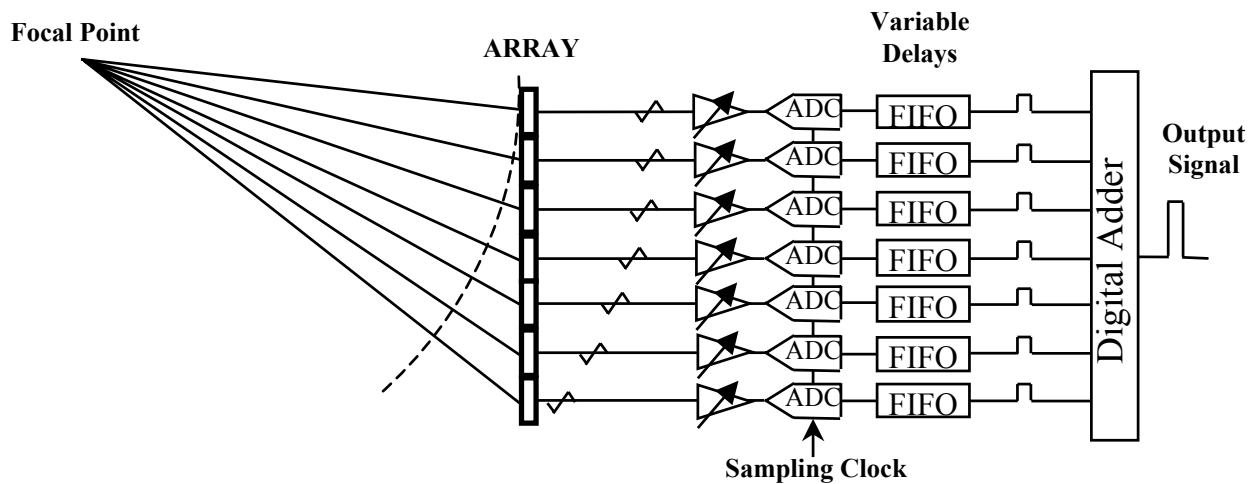


Figure 12. Simple Block Diagram of DBF System [4]

Figures 11 and 12 show basic block diagrams of an ABF and DBF system respectively. Note that the VGAs are needed in both implementations at least until a large enough dynamic range ADC is available (see Dynamic Range section). The main difference between an ABF and DBF system are the way the beamforming is done, both require perfect channel-to-channel matching. In ABF, an analog delay line and summation is used, while in DBF the signal is sampled as close to the transducer elements as possible and then the signals are delayed and summed digitally. In ultrasound systems, ABF and DBF, the received pulses from a particular focal point are stored for each channel and then lined up and coherently summed; this provides *spatial processing gain* because the noise of the channels are uncorrelated. Note that in an ABF imaging system only one very high resolution and high speed ADC is needed while in a DBF system ‘many’ high speed and high resolution ADCs are needed. Sometimes a logarithmic amplifier is used in the ABF systems to compress the dynamic range before the ADC.

Advantages of DBF over ABF:

- Analog delay lines tend to be poorly matched channel-to-channel
- Number of delay taps is limited in analog delay lines; the number of taps determines resolution. Fine adjustment circuitry needs to be used.
- Once data is acquired, digital storage and summing is “perfect”; i.e. in the digital domain the channel-to-channel matching is perfect
- Multiple beams can be easily formed by summing data from different locations in the FIFOs
- Memory is cheap, therefore the FIFOs can be very deep and allow for fine delay
- Systems can be more easily differentiated through software only
- Digital IC performance continues to improve fairly rapidly

Disadvantages of DBF over ABF:

- Many high speed, high resolution ADCs are needed (to process PW Doppler one needs about 60 dB of dynamic range which requires at least a 10 bit ADC)
- Higher power consumption due to many ADCs and digital beamformer ASICs
- ADC sampling rate directly influences axial resolution and accuracy of phase delay adjustment channel-to-channel; the higher the sampling rate (need correspondingly deeper FIFO for a given image depth and frequency) the finer the phase delay. Ideally ADCs with >200 MSPS would be used to get fine delay resolution [9], but because it isn’t possible to get ADCs with low enough power and high enough resolution for those speeds, most systems use digital interpolation in the beamforming ASICs instead

Practical Front-End Circuit Considerations

Transducer Interface

To appreciate the difficulty of achieving an optimal transducer interface it is necessary to discuss the type of transducers first. Figure 13 shows a slide of a presentation that summarized the four main transducer types [13].

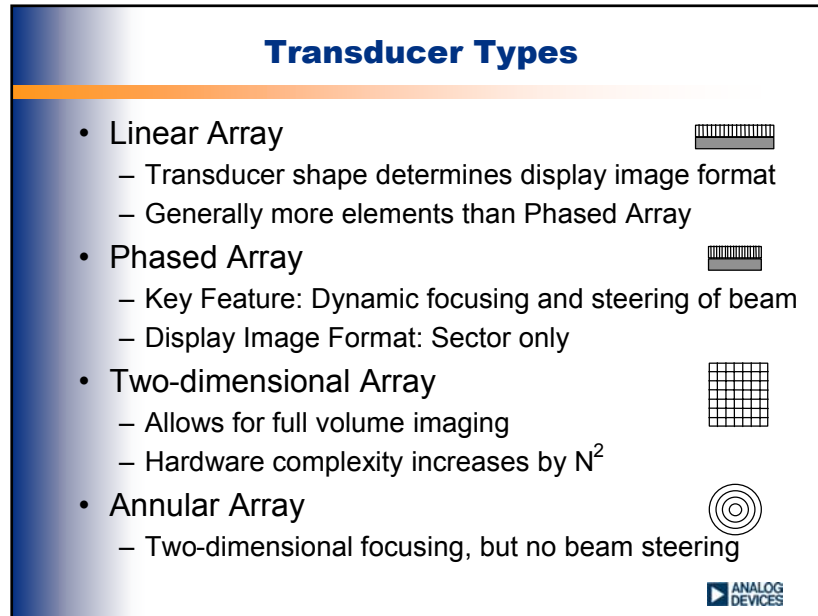


Figure 13. Transducer Types

- **Linear Array**: The shape of the transducer is determining the image shape. For example, if the array is convex, then the image generated will be a sector just like in a phased array.
- **Phased Array**: Its main advantages are full electronic steering of the beam and small size. This makes it the predominant transducer in cardiac imaging since one needs a small transducer to scan in between the ribs. The sector format is the optimal solution in cardiac imaging since the heart is far away from the surface and the beams all bundle near the skin, which makes them fit easily between the ribs. However, as pointed out earlier, the linear and phased array technology can be mixed and used to generate **compound linear scans**.
- **2-D Array**: This (theoretically) is the most versatile transducer since one doesn't have to move the transducer to scan a volume if a phased array approach is used. The biggest drawback of the 2-D array is that the complexity increases by N^2 . People have tried using sparse arrays, to alleviate the complexity increase, but so far nobody has made a commercially successful electronically steerable 3-D system. The only practical systems so far use mechanical motion to generate a volumetric image or use "1.25D or 1.5D" arrays; these sub-2D electronic arrays reduce complexity by restricting beam steering to only the lateral plane. They are sort of like an annular array that is sliced only in one direction. Another major problem is also the need for much larger cables to access the additional elements. The cable is one of the most expensive items in an ultrasound system, plus they become very stiff and unwieldy for 256 and more micro-coax cables. Because of this, high voltage multiplexers need to be used in the transducer handle to reduce the number of cables.
- **Annular Array**: This type of array allows for 2-D focusing, but no beam steering. With this type of array one can focus both in the lateral and elevation plane and produce a nice round beam. The focal point is determined, just like in the phased or linear array, by the delay pattern to the circular elements. As already mentioned above under 2-D arrays, a 1.25D or 1.5D array can be built which allows 2-D focusing but only 1-D beam steering. Full explanation of this technology is, however, beyond the scope of this article.

Transducer impedances can vary from less than 50Ω to $10k\Omega$ for single element and 2-D transducers respectively; most common are impedances of 50 to about 300Ω . With single element transducers one has more latitude in designing the transducer, this allows for customized impedances. In array transducers spacing between the elements is important to minimize grating lobes, therefore the spacing needs to be less than $\lambda/2$ (i.e., $250\text{ }\mu\text{m}$ at 3 MHz) [12]. Note that ultrasound is coherent just like laser light; therefore optical artifacts like grating lobes due to diffraction are present. Grating lobes are a problem in that they generate gain away from the main beam; if a strong undesired signal is coming along the direction of the grating lobe it could mask a weak signal along the main lobe. The main effects are ghost images and reduced SNR in the main image. The lateral size restriction makes it more difficult to design low impedance transducers; this problem gets compounded in 2-D arrays since one now also has the size restriction in the elevation plane in addition to the lateral plane. Lastly, as the transducer frequency increases, the wavelength and consequently the area decrease, which results in an increase in element impedance (reduction in capacitance; increase in real part). Increased transducer element impedances have the strong disadvantage that it becomes ever more difficult to drive the cable directly. I.e., a typical 2m cable might have a capacitance of 200 pF, while a transducer element could have capacitance on the order of 5 pF. This makes for a large capacitive attenuator; there are only a few possible solutions: (1) try to reduce the element impedance; (2) use a preamplifier in the transducer handle; (3) use a more sensitive LNA in the system. Solution (2) would be ideal, but it brings with it many problems like: how to protect these amplifiers from the high voltage Tx pulses; power consumption in the handle (heat); and area constraints.

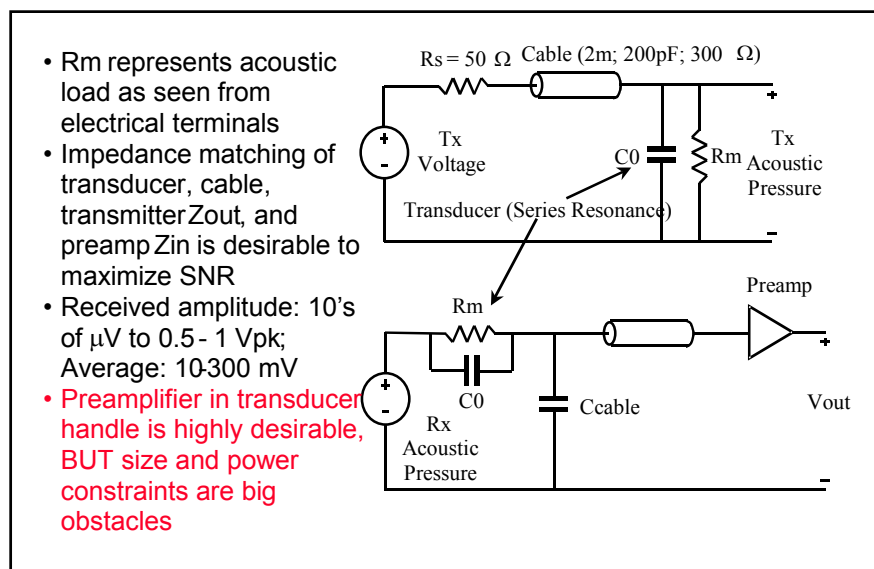


Figure 14. Transducer Interfacing

Figure 14 [2] shows a simplified equivalent circuit of an ultrasound front-end at series resonance of the transducer element. The upper circuit represents the electrical equivalent of the transmitter. A high voltage pulse ($>100\text{V}$) is generated on the left by a source with possibly 50Ω source impedance. This pulse is typically transmitted to the transducer element via a >2 meter micro-coaxial cable. A capacitor and resistor represent the model for the transducer. Resistor, R_m , is the electrical equivalent of the transducer plus body resistance. This resistor is **REAL** and therefore **NOISY**; ideally this resistor should limit the noise performance in an ultrasound system. The transducer element converts the electrical energy into acoustic pressure. The lower circuit represents the electrical equivalent of the receiver. Acoustic pressure is converted into electrical signals by the transducer. Cable loads the transducer and one normally tries to tune it out with an inductor, but since this is a narrow band solution and the ultrasound signals are broadband bandpass signals a resistor is needed that de-Q's the tuning resonator formed by C_{cable} and the inductor. This is best done with a resistive input preamplifier to minimize the degradation in receiver noise figure (NF). However, if the cable capacitance does not need to be tuned out then it is best to not set the input

resistance of the LNA; this gives the lowest noise performance. The worst thing to do is to shunt terminate with a resistor at the input of the LNA; this creates the worst NF performance.

Modern ultrasound optimized ICs like the dual AD8332 and the quad AD8334 allow for the input resistance to be set via a feedback resistor. The first generation of VGAs like the AD600 presented a ladder attenuator network at the input and needed a Low Noise Amplifier (LNA) preceding it for best noise performance; the second generation of ultrasound VGAs like the AD604 already included an integrated LNA, but still required dual supplies to achieve the ultralow input noise and also didn't provide the option of setting the input resistance; the third generation of ultrasound VGAs like the AD8332 are single +5V supply only components which integrate the LNA at the same noise performance as the AD604, but now at almost half the power. These new VGAs are differential to gain back dynamic range that is lost by going to a single supply; a differential signal path also allows for a more symmetric response, which as pointed out earlier, is extremely important for Doppler signal processing. Furthermore, since the author has learned more about the importance of excellent overload recovery and the need for perfect limiting in the receive signal chain this has also been attended to in the third generation VGAs for ultrasound by Analog Devices. The same can be said for the latest high speed ADCs (10b: AD9214, AD9218 and 12b: AD9235, AD9238) by ADI, they are primarily designed for communications applications, however, as has been pointed out multiple times already, the performance improvements required for ultrasound are also important in other receive signal applications like communications or test and measurement and vice versa.

Dynamic Range

The front-end circuitry, mainly the noise floor of the LNA, determines how weak of a signal can be received. At the same time, especially during CW Doppler signal processing, the LNA needs to also be able to handle very large signals, therefore maximizing the dynamic range of the LNA is most crucial since in general it is impossible to implement any filtering before the LNA due to noise constraints. Note that this is the same for **any receiver** - in communications applications the circuitry closest to the antenna doesn't have the advantage of a lot of filtering either and therefore needs to cope with the largest dynamic range.

CW Doppler has the largest dynamic range of all signals in an ultrasound system because during CW one transmits a sine wave continuously with half of the transducer array, while receiving on the other half. There is strong leakage of the Tx signal across into the Rx side and also there are strong reflections coming from stationary body parts that are close to the surface while one might want to examine the blood flow in a vein deep in the body with resultant very weak Doppler signals. CW Doppler signals cannot currently be processed through the main imaging (B-mode) and PW Doppler (F-mode) path in a digital beamforming (DBF) system, this is the reason why an analog beamformer (ABF) is needed for CW Doppler processing in Figure 1 [5]. The ABF has larger dynamic range, but the "holy grail" in DBF ultrasound is for all modes to be processed through the DBF chain and there is ongoing research in how to get there. It seems that at least a *perfect* 15 bit SNR ADC with a sampling rate of >40 MSPS will be needed before CW Doppler processing through the 'imaging' channel is possible.

Just to get an idea of the current (March 2002) state-of-the-art in high speed, high resolution ADCs, Analog Devices' AD6644 14b/65MSPS ADC has an SNR of 74 dB, this is 12 dB below (!) the theoretical SNR of a 14 bit ADC. Because actual ADCs will always be below the theoretical SNR, it is very likely that until a 16b/>40MSPS ADC exists CW Doppler will still need to be processed through a separate analog beamformer. One more thing that needs to be pointed out in this regard, even if such an ADC would exist today, one still has to figure out how to drive such a device! **ALL** components before the ADC need to have at least the same dynamic range as the ADC, and since in most systems the input signal level at the antenna (transducer) is not perfectly mapped to the input range of the ADC, one will possibly need an amplifier that maps the level optimally onto the ADC

input range. This is currently done through a LNA + VGA combination like in the AD8332 dual channel device for 10 and 12 bit ADCs.

Yet another wrinkle in the pursuit of a single receive signal channel for all modes in ultrasound is the need for the beamformer to be able to utilize all the bits provided by the ADCs. Current state-of-the-art beamformers are 10 to 12 bits per channel, going to 16 bits will certainly increase the complexity and memory requirements of the beamformer ICs significantly.

Power

Since ultrasound systems require many channels, power consumption of all the front-end components – from T/R switch, through LNA, VGA, and ADC, to the digital circuitry of the beamformer – is a very critical specification. As has been pointed out above there will always be a push to increase the front-end dynamic range to hopefully eventually be able to integrate all ultrasound modes into one beamformer – this will tend towards increasing the power in the system. However, there is also a trend to make the ultrasound systems forever smaller – this tends towards reducing power. Power in digital circuits tends to decrease with reduced supply voltages, however, in analog and mixed signal circuitry this is not the case. Therefore, there will be a limit in how low the supply voltage can go and still achieve a desired dynamic range.

In the current state-of-the-art CMOS ADCs (AD9235 and AD9238), the supply voltage for a 12 bit/65MSPS device is 3V. For the AD8332 LNA + VGA, the supply voltage is 5V. Already the newest devices use differential signaling throughout to increase headroom, but in analog circuits there are eventually limits to reducing power by reducing supply voltage. For example, most of Analog Devices' ADCs now use a full-scale (FS) input of 2Vpp. It should be immediately obvious that if the FS is fixed, but the resolution increases, the only way to go is down – in noise that is. Lower noise circuits tend to consume more power and at some point it may be better to stick with a larger supply voltage to reduce the need for ultra-low noise front-ends. It is the author's opinion that for input referred noise voltages of less than 1 nV/ $\sqrt{\text{Hz}}$ and dynamic ranges of >14 bits it doesn't make much sense to go below a single 5V supply due to headroom and power constraints.

Integration

Another important consideration is how to best integrate the front-end components to optimize space savings. Looking back at Figure 12, one can see that there appear to be two possible integration paths: (1) across multiple channels of VGA, ADC, etc.; or (2) along channels VGA + ADC + beamformer. Each option has advantages and disadvantages depending on the system partitioning.

To show the trade-offs in IC integration, Figure 15 shows a few simple diagrams of some partitioning options of the front-end. Typical trade-offs are:

- Isolating analog and digital signals
- Minimizing connector pin counts
- Reducing power and cost by using fewer ADCs

The most versatile and therefore costly option is shown in the uppermost picture, it uses a full DBF approach with access to all transducer elements and thereby can implement a phased array. In this example, there is a separate Tx and Rx board, plus a digital board. This is a good partition if one wants to minimize connectors between the Rx analog board and digital board. Having the ADCs on the digital board also allows easier connection of the many digital lines, i.e. $128 \times 10 = 1280$ lines if 10 bit ADCs are used, from the ADCs to the beamformer ASICs. The high digital line count indicates one of the integration options – **integrating the ADCs into the beamformer ASICs**. However, there are some serious problems with this approach: (1) typically high performance ADC processes are not compatible with smallest feature size for digital design, and the beamformer ASICs

can have very high gate counts and therefore should use a process that is optimized for digital design; (2) putting the ADC onto the DBF ASIC doesn't allow an easy upgrade to improved ADCs over time. This approach is attractive for low performance systems where an ADC of 10 bits or less is integrated; there are some manufacturers of portable ultrasound systems that have done this; (3) beamformer implementation contains significant intellectual property by the ultrasound manufacturers plus every manufacturer has their own approach to implementing the beamformer. Therefore these facts would make it difficult for a company like Analog Devices to make a part that could satisfy all customers.

The middle picture in Figure 15 shows an ABF system, in that case the low noise pre-amplification and TGC function is still needed to compensate for the signal attenuation in the body. The same is true for the lowest picture, but it shows a further way of reducing cost and power of a system (both for DBF and ABF, even though an ABF systems is shown) by reducing the **active channels (defined as the number of VGAs needed)** from 128 to 64 by an analog multiplexer. Most commonly a high voltage multiplexer is used before the T/R switch since that way the number of active channels is reduced in both the Tx and Rx circuitry.

From the above discussion it should become clear that the most sensible approach is to **integrate multiple channels of VGAs and ADCs**. Furthermore, it might appear that integrating the VGA and ADC together is a good idea, yet again this is not ideal since it restricts the usage of the component in a system partitioning where VGA and ADC are supposed to be separated. Because of these reasons, Analog Devices decided to focus on multiple channel devices like the dual AD8332 and quad AD8334 VGAs, and dual ADCs like the AD9218 (10b) and the AD9238 (12b). Similar arguments can be made for the Tx side as well; in that case there is an even bigger discontinuity in process technology because of the very high voltages that are required to drive the piezoelectric elements. Ultrasound which is still a 'relatively' low volume application can greatly benefit in terms of low cost by utilizing 'standard' components that are designed for other applications, in particular communications.

One last comment on the optimum number of integrated channels, four seems to be a very good number since for higher channel counts the area required by external components and wiring may make it very difficult to gain a significant area advantage – one of the main reasons to go to higher integration in the first place. There is of course still the benefit of potential improved reliability in more highly integrated components.

Conclusion

This paper attempted to show the trade-offs required in front-end ICs for ultrasound by explaining the basic operation of such a system first and then pointing out what particular performance parameters are needed to insure optimal system operation. For the reader that wants to find out more about ultrasound systems, I'd recommend the book in reference [6], it gives a good overview without getting into too much detail. Lastly, the author would like to thank the many people that have helped him in understanding ultrasound systems, especially the engineers at GE Medical Systems in Milwaukee, WI, USA and Norway.

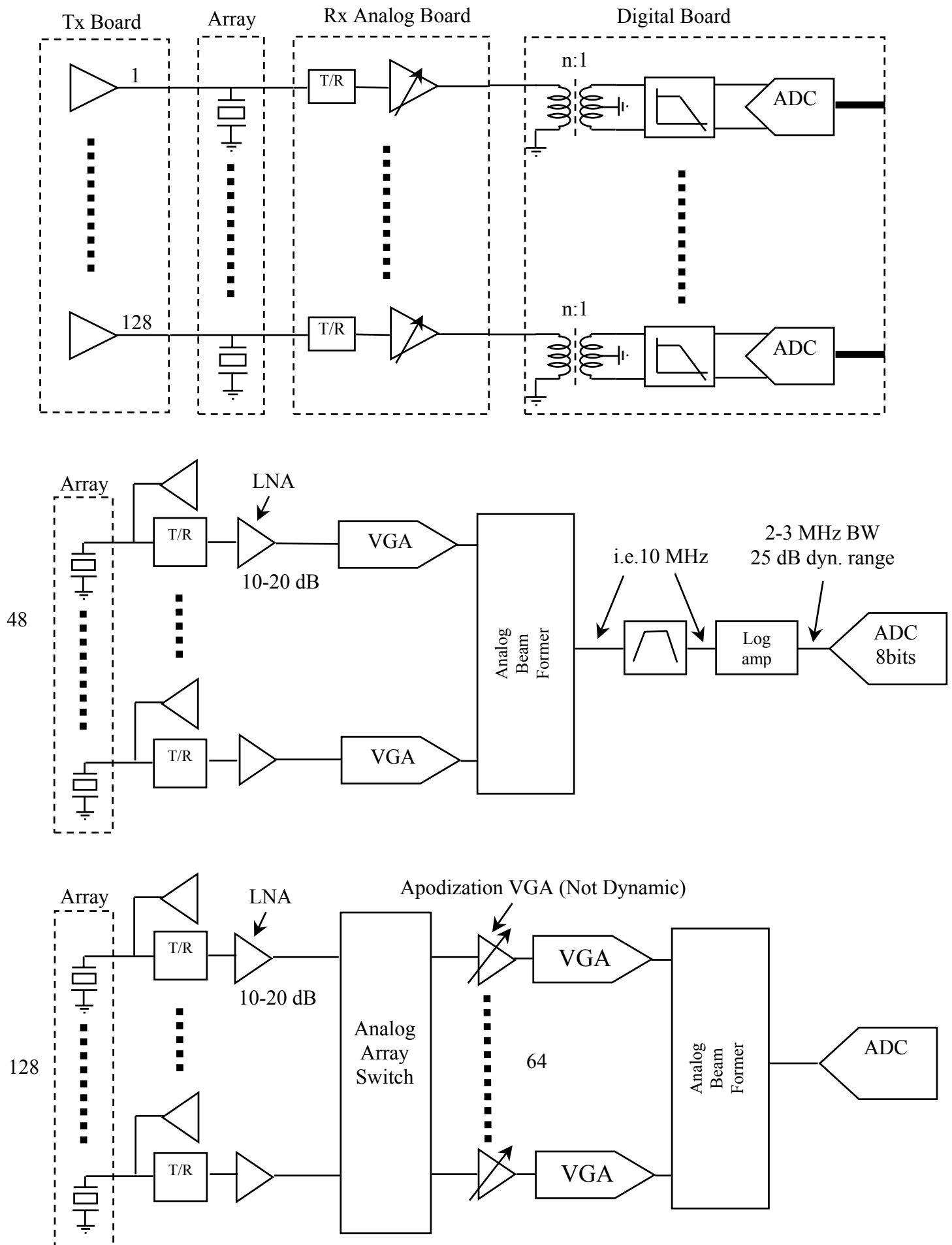


Figure 15. Some System Partitioning Options

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