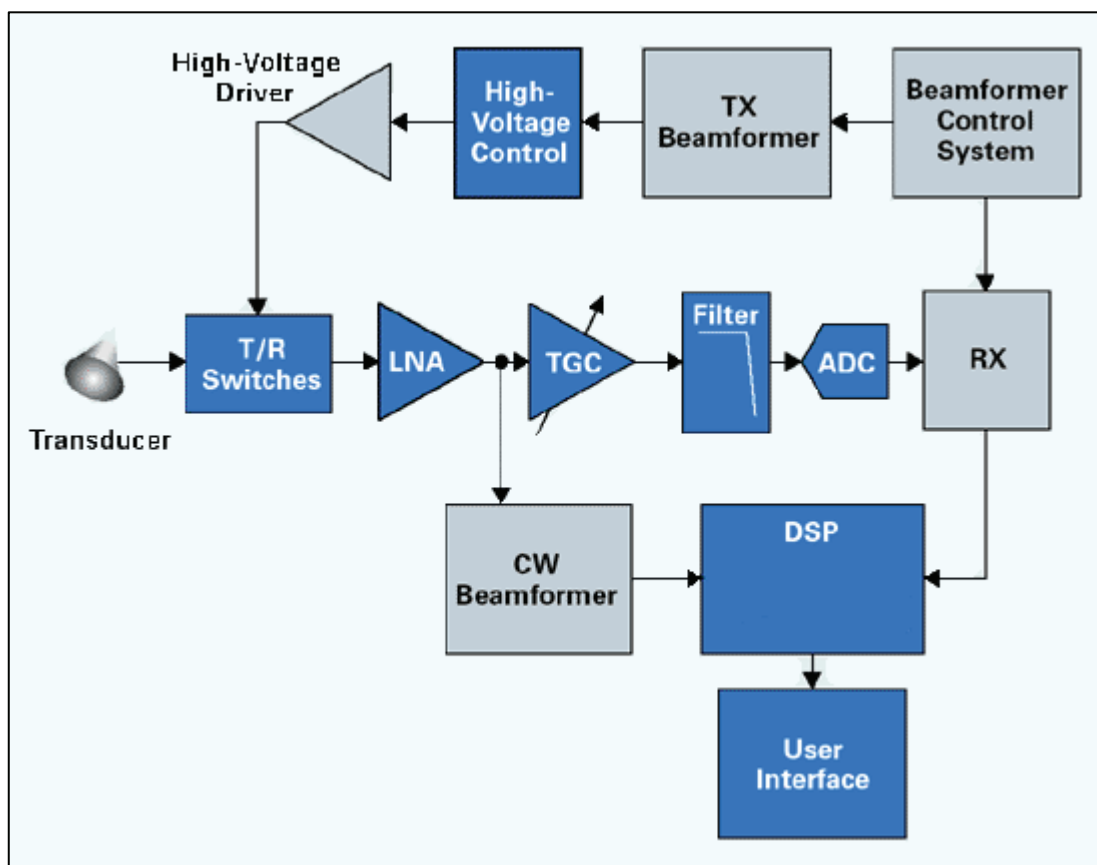

A Primer on Ultrasound System Electronics For Biomedical Engineering Professionals

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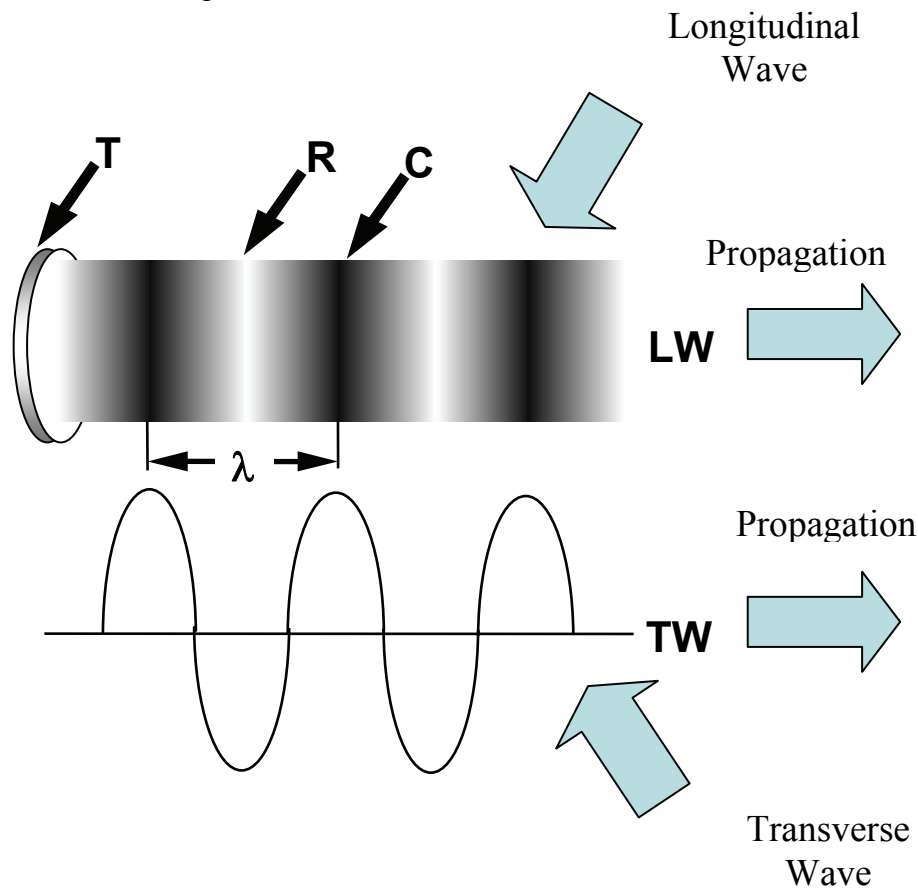
INTRODUCTION

As more and more hospitals rely on their internal biomedical engineering resources to provide “first-look” services for the ultrasound assets within the hospital or hospital system the need for a comprehensive, and accessible technical “White Paper” describing the core operation and system architecture of full-modality diagnostic ultrasound devices common to all Original Equipment Manufacturers is apparent.

This Primer, then, is intended to address that need. The reader will be introduced to the basic architecture of today’s ultrasound systems, and will gain an understanding of the constraints on system design and configuration. Human physiology and the laws of physics dictate the general design criteria of most of the subsystems that make up the aggregate ultrasound system architecture. Further, the variances in body habitus encountered in the clinical setting necessitate a large number of user controls that attempt, with various degrees of success, to compensate for the ultrasound beam distortions and attenuation created by these variances. In the field of medical imaging, the operator (in the case of ultrasound the sonographer) interaction with ultrasound systems is far more intense than with any other imaging modality. Understanding the basic operation of the system and its various modes of operation will facilitate better communication and trust between the Biomedical Engineer and the clinical user and will greatly enhance the troubleshooting process.

A BRIEF OVERVIEW OF ULTRASOUND PHYSICS

Sound waves are mechanical waves that travel at different speeds through different media. The average speed of sound waves through seawater is 1531 meters per second (m/s).¹ Because humans are basically “bags of water” it is not surprising then that the average speed of sound through the human soft tissue is generally accepted to be 1540 m/s (**Note:** all diagnostic ultrasound systems in the United States are calibrated to 1540 m/s). 1540 m/s is a limitation in ultrasound transducer and system design. It defines the timing for transmitting and receiving signals from different depths.



Of the mechanical waves, we are interested in two: the longitudinal and the transverse waves. This graphic shows events within a longitudinal wave and a transverse correlate. The longitudinal wave forms changes in pressure in regions and correspond to the changes in transverse wave amplitude. T is the transducer, R is rarefaction, C is compression. Soft tissue imaging and Doppler only involve longitudinal waves. Transverse waves occur in steel and other materials

¹ CRC Handbook of Chemistry and Physics, CRC Press

The distance sound travels during a period of time is a function of the media that it travels through. $d = c \times t$, so $t = d/c$ and since $c = 1540$ m/s in the body, we can calculate when to “listen” for a return echo at any given distance. Of course the return signal has traveled both directions between the transducer and a reflecting structure, so the time to receive is $t = 2d/c$. This means that the elapsed time for an echo to return to the transducer tells us the depth of the reflecting structure.

There is a relationship between the speed of sound in the body, the frequency of the sound wave, and its wavelength. This relationship can be expressed by; $c = f \lambda$, where c is acoustic velocity (i.e., 1540 m/s), f is the frequency of the wave (basically the transducer frequency), and λ is the acoustic wavelength.²

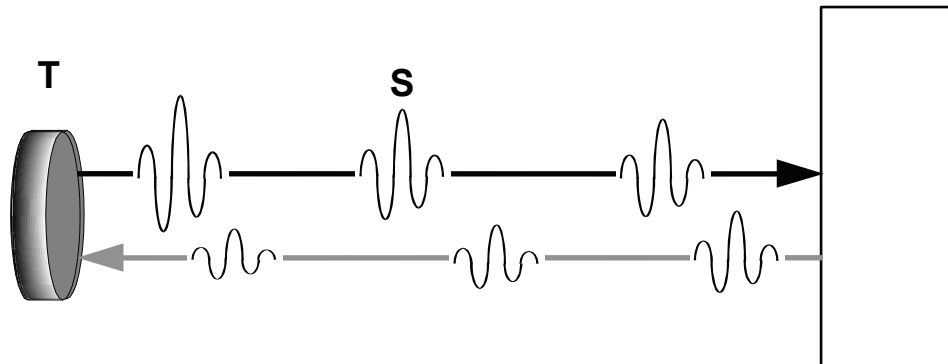
The relationship $c = f \lambda$ limits a key component of ultrasound imaging resolution, specifically spatial resolution. If $c = 1540$ m/s and $f = 2$ MHz, then $\lambda = 0.77$ mm. In contrast, if $f = 20$ MHz, then $\lambda = .075$ mm. Imaging resolution is a function of the wavelength; so it is easy to see that resolution increases as frequency increases, and lower frequencies produce poorer resolution. Axial resolution (along the beam line) is limited by the frequency or bandwidth of the pulse. Lateral resolution (across beam lines) is limited by frequency and the size of the array. Different frequencies also play a role in attenuation, or signal loss through the body.

Signal loss of ultrasound energy through the human body is approximately 0.5 dB/cm/MHz (see **diagram on following page**). For every -3 dB, $\frac{1}{2}$ of the power is lost. Let’s look at a couple of examples to see the importance of this relationship. At 12 cm depth, operating at 5 MHz, there is a 30 dB signal loss in one direction, for a 60 dB total “round trip” loss. That translates to only one millionth of the initial energy. At 20 MHz, the same signal loss occurs at only 3 cm. One can see the importance and critical trade-offs of frequency in the physics of ultrasound as it relates to attenuation in tissue, and image resolution.

Acoustic impedance differences cause signal loss, but also contribute to the success of ultrasound imaging. Different materials or different tissue structures have different acoustic impedance characteristics. This is beneficial because the interface boundaries of these different tissues are acoustic mismatches, and acoustic mismatches produce echoes. Acoustic impedance is a function of the velocity of sound and the density of a material. Bone, for example has much different acoustic impedance than soft tissue.

² Academic Press, Physical Acoustics Series, Volumes 23 and 24, Contributors; A. Goldstein, R. Powis

Ultrasound Medium Attenuation

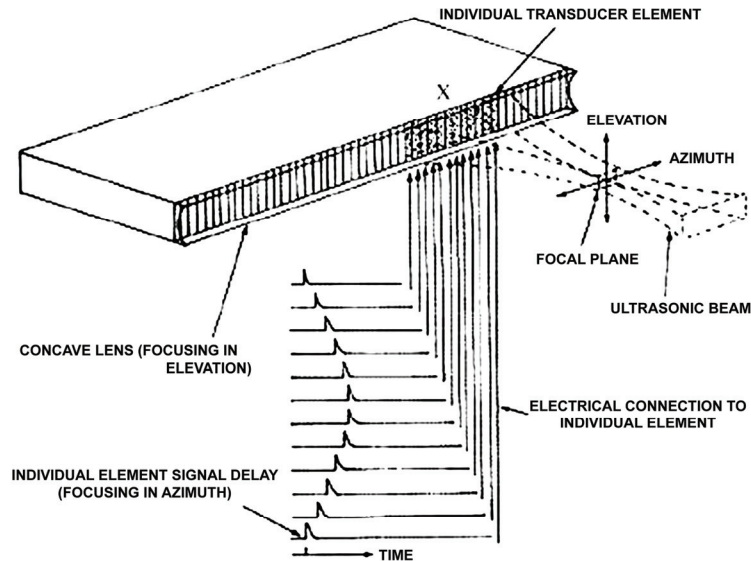


As the ultrasound travels through the tissues out and back, the tissue absorbs the ultrasound energy by converting the acoustic energy to heat, and also loses energy due to scattering. In effect, the compressional waves perform work on the particles of the medium carrying the waves. T is the transducer and S is the signal amplitude.

TRANSDUCERS

Transducer design is driven by the structures of interest and their location in the body. Transducers are also built around the limitations of ultrasound physics. This leads to different shapes, sizes, curve radiuses, elements, apertures, and frequencies in various designs.

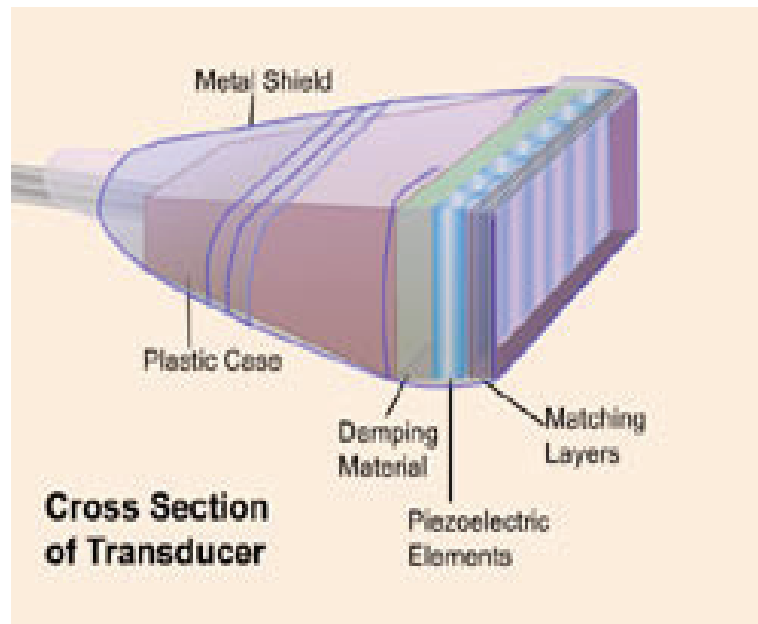
Lateral resolution is a function of beam width. And beam width is a function of aperture and frequency. Aperture is a term that refers to the transducer's dimensions. It is comprised of "elevation", which is the height of the crystals and "azimuth", which is the width of the array of the active portion of the array at any given time.



A large transducer aperture produces greater lateral resolution and a smaller one produces less. We know that higher frequencies produce finer spatial resolution. So, a large aperture transducer operating at a low frequency could be roughly equivalent in lateral resolution to a small aperture transducer operating at a higher frequency.

Transducer construction, simply put, consists of a lens, piezoelectric crystals, and backing material. The crystals in their entirety are referred to as the "array" and individually as "elements". The lens provides impedance matching between the array and human skin. It also acts as a wear layer and protection from electric shock hazard. The backing material acts as a

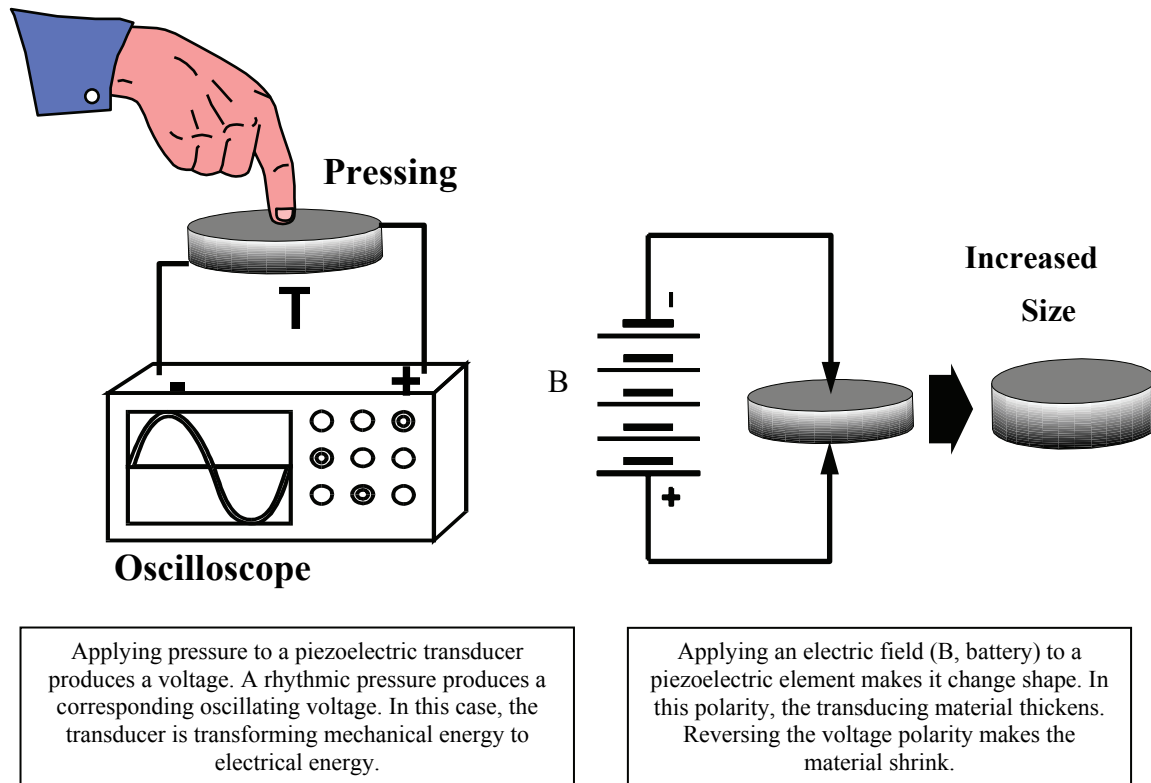
dampening agent so the oscillating crystals produce ultrasonic waves that travel into the subject and only in that direction.



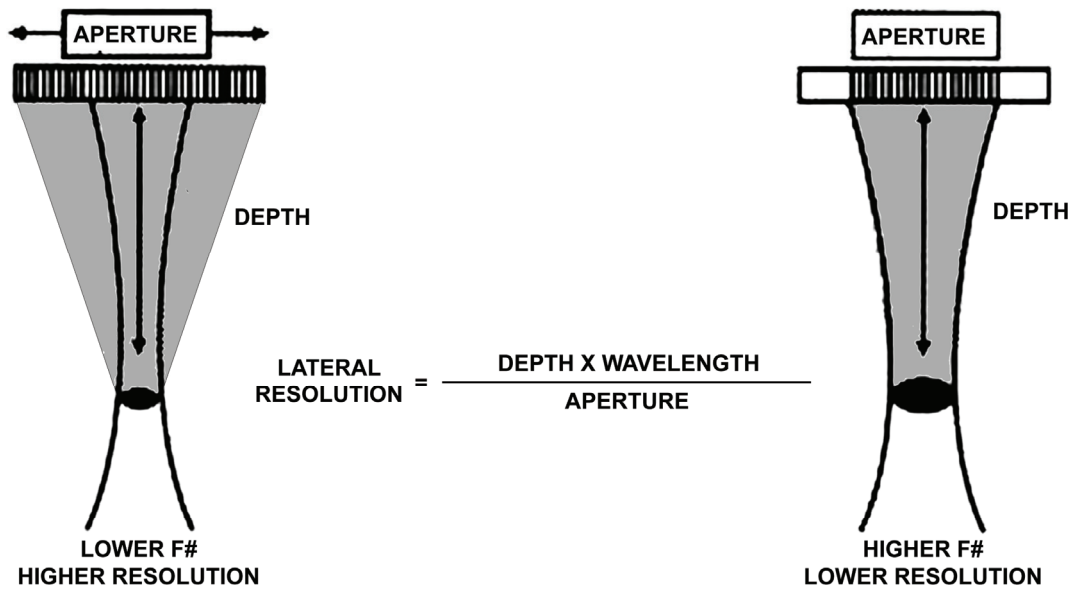
Piezoelectric crystals are the key enablers to ultrasonic imaging. They embody properties that link voltage and mechanical movement. When a voltage is applied across one of these crystals it mechanically deforms, thus producing ultrasonic sound waves. Likewise, when the crystal is mechanically deformed it produces a voltage (**see diagram on following page**). Thus, a voltage pulse is applied across the crystal producing an ultrasonic pressure wave. That wave travels through the lens and into the body where it is reflected. After a period of time it returns to the crystal and deforms it. That mechanical deformation then produces a voltage. This is the raw data that is then processed to produce a diagnostic image.

Many different types of probes have been developed to accomplish specific imaging objectives. They include the linear array, curved linear array, phased array, and the Doppler continuous wave (CW) single split element transducer. Each of these probe types have further variations such as different frequencies, curve radiuses, and dimensional options. For example, a large, low frequency, large radius curved array probe may be used for deep abdominal imaging and a small, high frequency, small radius curved array probe would be used for Endocavity imaging.

Piezoelectric Effect



Linear and curved linear array probes are common to all ultrasound systems. They are the least expensive probes and both function in a linear fashion. A group of elements is used to transmit and receive in order to produce one axial image line. In an array comprised of 128 elements, for example, typically 32 to 64 elements would be used as the active aperture. Once a beam line is formed, the active aperture moves over one element (e.g. ...#10 thru #42, #11 thru #43, #12 thru #44, and so on...) and repeats the transmitting and receiving functions to produce the next beam line. When the walking aperture has traveled completely across the array all of the beams created form a complete frame. It is possible to produce systems that only need 32 channels of hardware to support these linear transducers, which is a design cost savings.



Those familiar with ultrasound system operation are aware of transmit “focus caret” that can be selected for a particular depth on the display. In order to focus ultrasound beams to improve the diagnostic image at a certain distance, we use transmit timing delays. Note that the lens on the probe is not entirely like an optical lens – it does not necessarily focus beams (elevation focus), but it performs other key functions as well. Azimuthal focus is accomplished by introducing slight delays in firing adjacent elements. Delays have the effect of making the linear array appear as a concave curve as beam lines are created. Concave curves (even virtual ones) produce a focal zone or focal point. The shape of the virtual curve (different timing delays) determines the depth of the focal zone.

Adding a second focus caret to the display requires the creation of another shaped virtual curve for the new focus zone, which in turn cuts the frame rate in half. Two complete sets of beam lines are required for each frame when two transmit foci are selected because a transmit pulse can only have one focal point. Note that this is different than receive focus delays, which are introduced continuously for all depths and will be discussed later in this document.

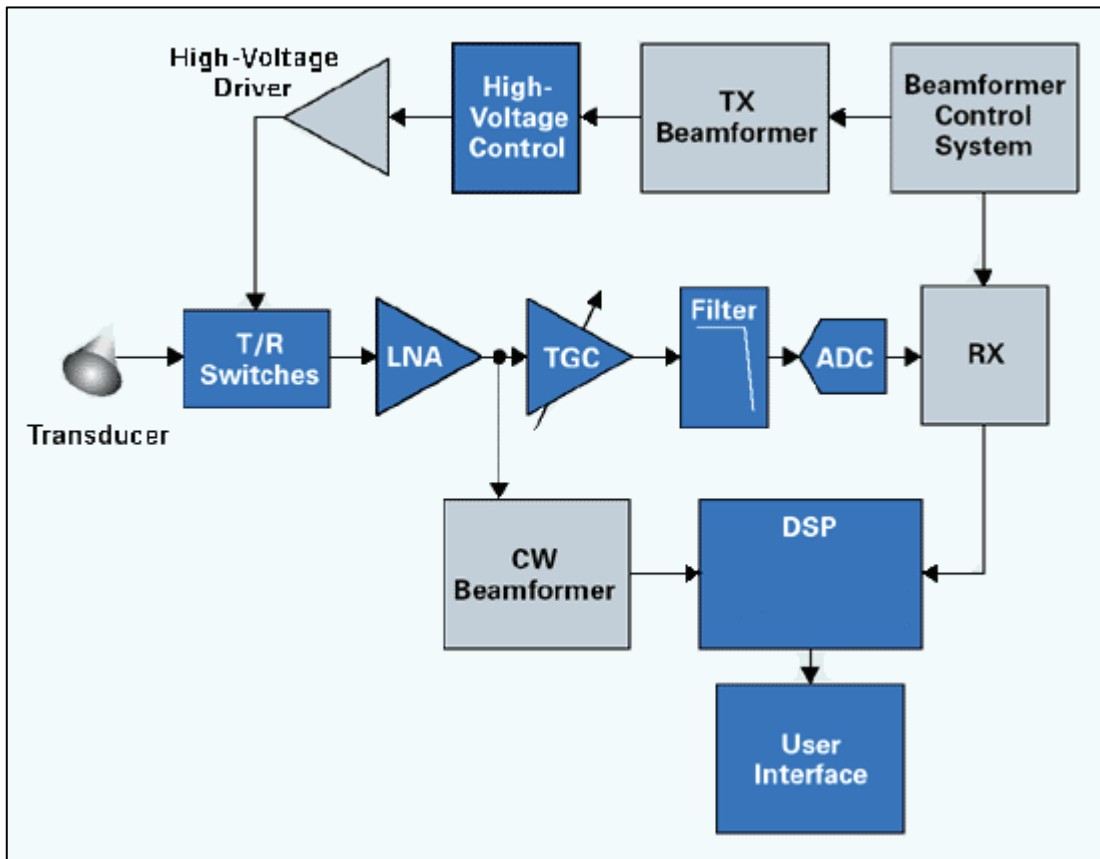
Phased array probes were initially developed for cardiac imaging, although today there are phased array probes designed for other clinical applications as well. In order to collect the data required to form an image of the heart, one needs to be able to scan between the ribs (i.e., the intercostal space) and to “steer”, or “angle” the beams. Because of this acoustic access limitation, the aperture size of a cardiac phased array probe is typically smaller than most linear probes, and often has fewer elements.

Phased arrays also use receive delays to accomplish steering – which are also referred to as “steering delays”. Steering delays are much longer than focus delays because the beam angles required are much more acute. Steering delays are used in both transmit and receive. In phased array probes the active aperture is always comprised of all the elements for transmit and receive. Phased arrays also make use of focus delays during transmission and reception.

Because of the electronics required to perform beam steering ultrasound systems that offer phased array capabilities are more expensive and complex in design than linear and convex array only machines. There are still a few systems on the market that use a mechanically steered single crystal or an annular array transducer to form a sector image. Thus avoiding the necessity of using complex beam steering electronics and the costs, but these system generally have other issues that compromise clinical performance and reliability, and prevent them from being more widely accepted.

ULTRASOUND SYSTEM ARCHITECTURE BLOCKS

All ultrasound systems are comprised of similar system architecture blocks. The three primary functions within each system are echo transmission, acquisition and display processing. Echo transmission and acquisition is often referred to by the term “scanner”. This circuitry is located in the “front end” of the system. Display processing takes place in the system “back end” otherwise known as the scan converter.



FRONT END/SCANNER

The transmit side of the front end is made up of transmit and receive (T/R) switches, a transmit beamformer, high voltage control, and high voltage drivers. The receive side of the echo acquisition front end also uses the T/R switches, a low noise amplifier, time gain control, filter, and analog-to-digital converter.

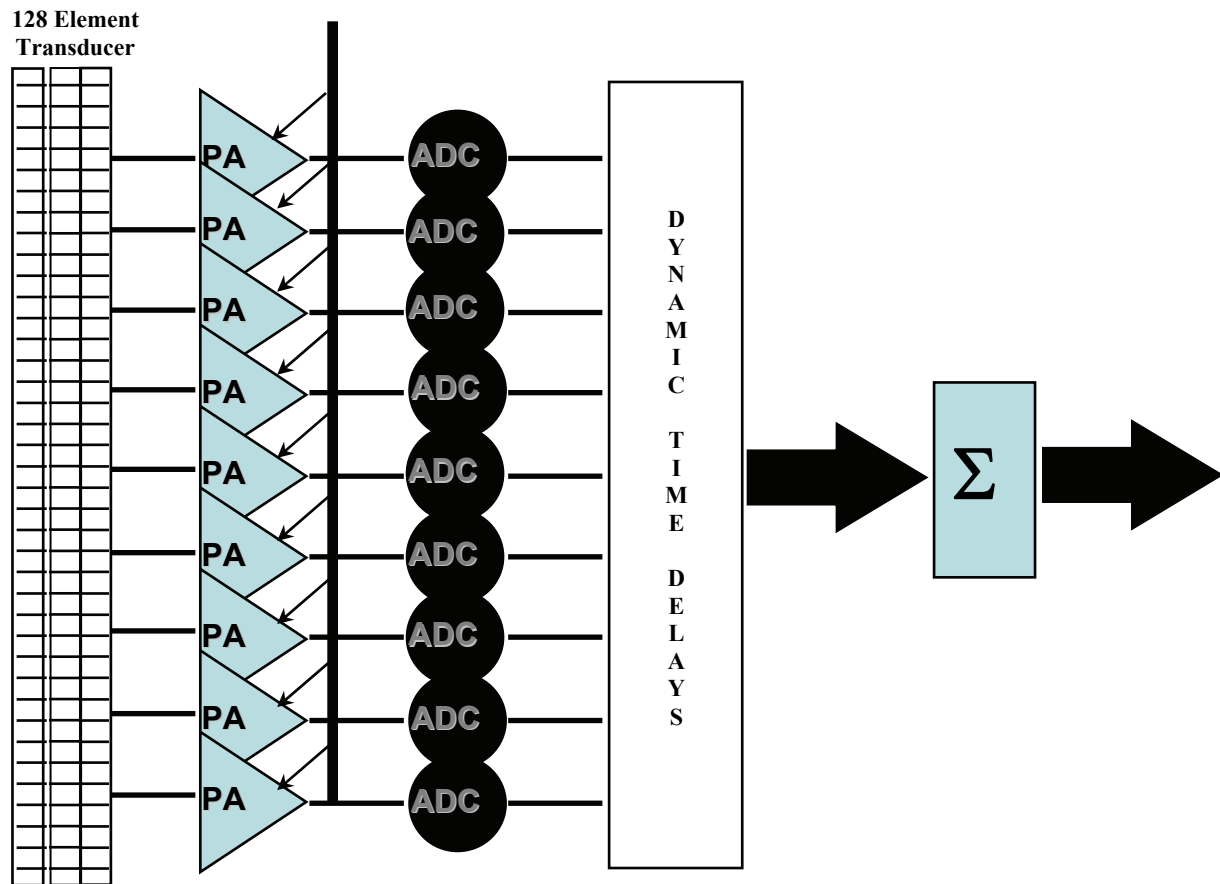
T/R switches isolate the sensitive receive channels from the high voltage transmit pulses as they connect first transmit and then receive lines to the individual conductors for each element. There are separate switches for each channel, and dedicated channels for the maximum number of active elements at any given time for each particular system. This could range from 32 to 128 and more, with most modern systems using at least 128 channels.

The transmit beamformer selects which elements to fire, and when. This is the aperture selection function. The time delays are chosen for transmit focus, beam steering, or both. The beamformer provides the input to the high voltage driver section. It is also tied to the high voltage control from the “power” or “output” adjustment on the user interface panel. So, the transmit section includes a user controlled high voltage power supply that supplies the high voltage drivers that are driven by the transmit beamformer.

On the receive side of the system front end, immediately past the T/R switches are the low noise amplifier (LNA) channels. These preamplifiers have a separate channel for each of the maximum number of elements that the system might use. Once the analog return signals are amplified, they enter time gain control (TGC) or depth gain control (DGC) channels. Because of the effect of tissue attenuation, returning signals must be amplified depending on depth of field. Greater gain for returns from greater depth, or more gain for signals that take longer to return helps to provide a homogeneous displayed amplitude of the image. The TGC or DGC circuitry is tied to the slide potentiometers found on the user interface panel.

Digital Beam Former

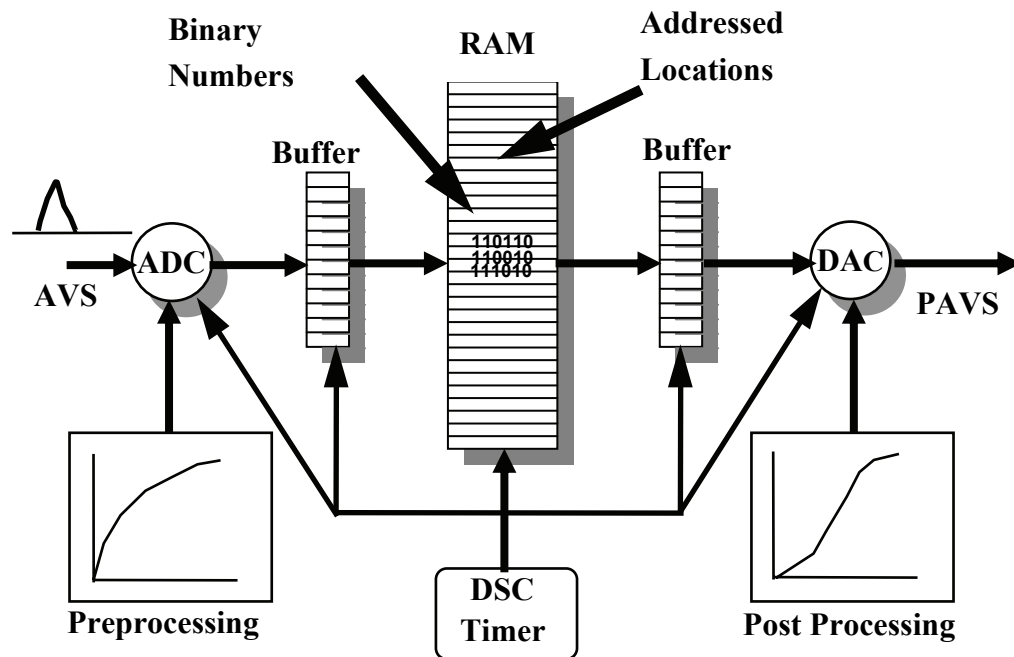
TGC Application



An anti-aliasing filter follows the TGC before the analog to digital converter. The digitized data is fed to the receive beamformer, which executes a continuous focus in receive mode. The same transmit time delays used to focus the transmit pulses (virtual lens) is used in receive mode, except the time delays are introduced after the data is received and so can be employed to provide greater focus at all depths.

BACK END/SCAN CONVERTER

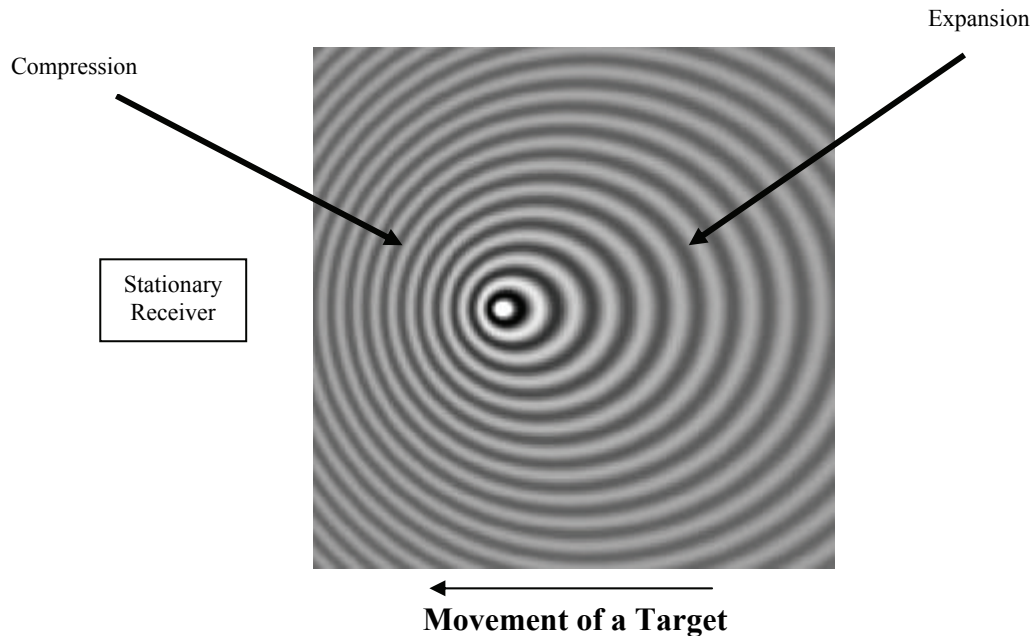
In older systems, the back end of the system is where all of the digital signal processing (DSP) is accomplished. Newer system designs are pushing parts of the DSP further into the front end. The signal from the receive beamformer is amplitude detected and then sent through a display compression algorithm (logarithmic compression). This is to optimize display intensity. Another image enhancement process is scan conversion. When the return signals are received through their individual channels the data is in the form of beam lines. Data points along different beams or rays are in the form of polar coordinates (r,θ) . The scan conversion process manipulates the data set to the Cartesian coordinate format (x,y) . This is necessary to provide an image that can be displayed on the user interface monitor, or printed or stored in digital format.



DSP works on an analog video signal (AVS) to produce a processed video signal (PAVS). ADC analog to digital converter, RAM random access memory, DSC timer synchronizes events, DAC is digital to analog converter. Preprocessing sets the assignment of amplitudes to the available binary numbers, while post processing assigns numbers to gray scale.

DOPPLER

Transmitting acoustic energy towards a reflecting object in motion produces a shift in frequency of the return signal that is related to the velocity of the object and the angle of energy incidence. This phenomenon is known as the Doppler effect, or Doppler shift. It is useful in ultrasound because it allows us to measure blood flow velocities and to observe areas of interest for relative velocity directions and speeds.



The characteristics of the Doppler shift have led to three primary clinical ultrasound implementations that have various benefits and drawbacks. They are Continuous Wave (CW), Pulsed Wave (PW), and Color Flow. Also, those limitations make it difficult to integrate Doppler into conventional two dimensional ultrasound electronics. For that reason, Doppler is being treated as an independent modality in this section of the paper.

The Doppler shift frequency is expressed as; $f_d = 2f_t v \cos\theta/c$; where v is the velocity of the target, θ is the incident angle between the beam line and the flow direction, f_t is the transmit frequency, and c is the speed of sound in the body. One can immediately see the importance of the angle, θ . When the beam line is perpendicular to the flow vector, θ is 90° , and $\cos\theta$ is zero! In that case, or when the beam line is close to perpendicular to the blood flow there is no Doppler shift frequency toward the stationary receiver, and the impact of a small angle error on the velocity calculation is very large.

Conversely, at a theta of zero degrees cosine theta is one, and small angle errors have very little effect on the accuracy of the calculated velocity. At 45° cosine theta is 0.707. One should avoid making Doppler velocity measurements at angles over 60°, therefore, as large errors may occur. In most systems the operator places a line on the believed axis of flow and the system calculates theta and after measuring the frequency shift can calculate velocity.

Continuous Wave (CW) Doppler as its name implies, transmits continuously. It also receives continuously and so the ultrasound system either “divides” the array to perform the two separate functions or uses a non-imaging “split D” crystal – half for transmit and half for receive. Split D crystals are encapsulated in so called “Pencil Probes”.

Many modern digital beamformers do not have the dynamic range necessary to adequately detect the full range (often as high as 100dB) of Doppler shifts encountered when using CW in cardiac applications. Because of this limitation the system designer typically utilizes separate transmit and receive circuitry.

The great clinical benefit of CW Doppler is that there is no practical limit on the maximum velocity that it can measure. The significant limitation is that you can only measure along one beam line and at an indeterminate depth, that is, there is no range resolution. The CW Doppler receive signal is processed by using a very steep notch filter at the transmit frequency, because of the constant transmission.

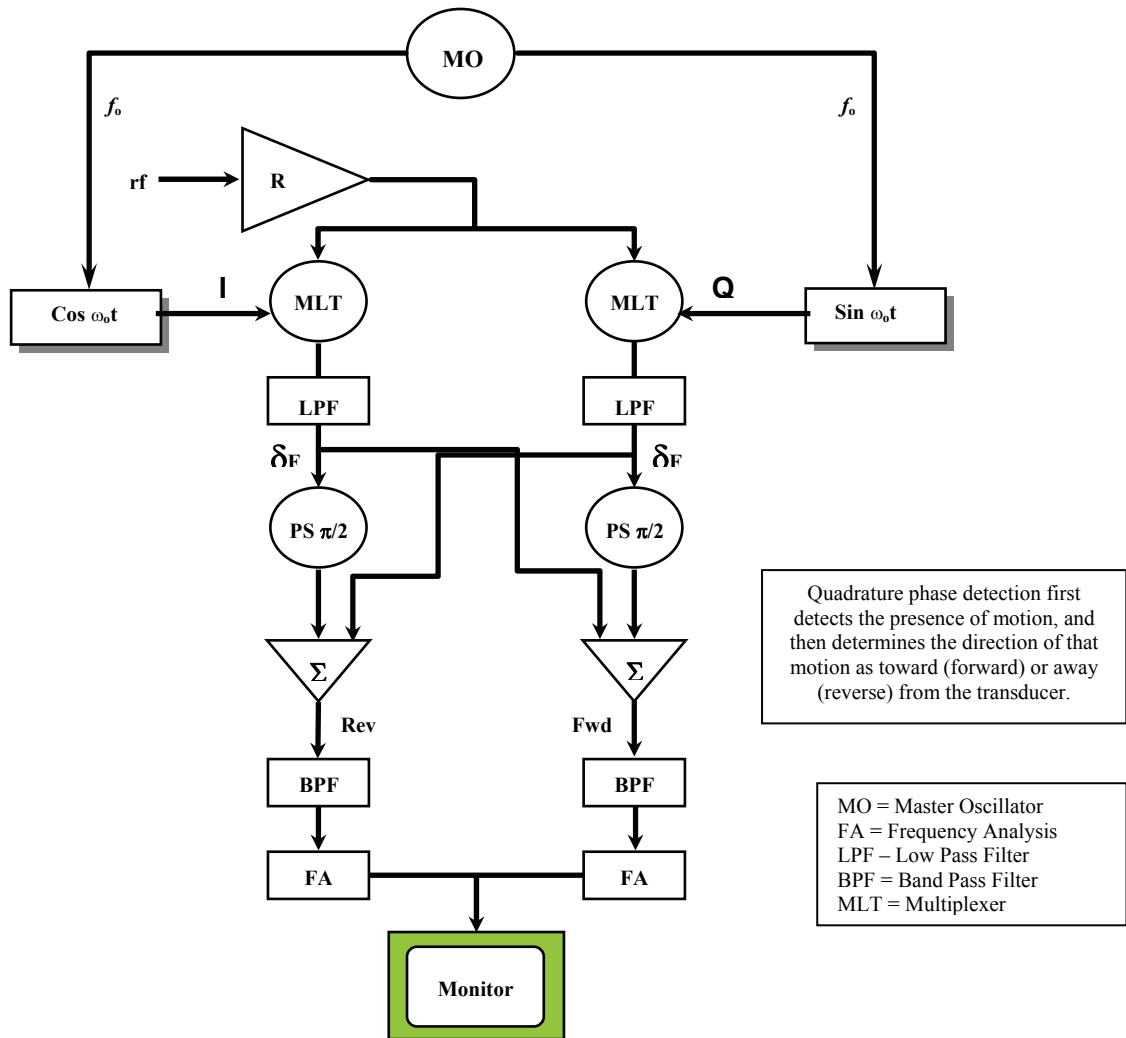
Pulsed Wave Doppler (PW) is similar to CW Doppler in that it collects data along a single beam line. PW Doppler adds the benefit of a selectable sample location, i.e. depth, sometimes called a “gate” or “range gate”, this provide range resolution. This comes at the trade-off of limited maximum frequency shift, which translates into a limit on maximum velocity that can be measured.

PW uses a pulse stream with each pulse typically made up of 3 to 25 cycles. Spacing between the pulses determines the pulse rate frequency (PRF), which, in turn, determines the maximum detectable velocity without ambiguity. PW Doppler signal processing requires not only a steep notch filter at the transmit frequency, but also a low pass filter to remove the PRF frequency component.

Color Flow is the third type of Doppler imaging. It has characteristics similar to PW Doppler, but uses several beam lines and limits the pulse train to a burst of pulses as it scans the area of interest. So, while PW utilizes a continuous pulse stream, Color Flow uses bursts of pulses. For each beam line, a burst of typically 16 transmit pulses is emitted. Along each beam line a series of sample volumes, just like the one used in pulsed Doppler, are taken for each color display element. The modal frequency for each display element is then converted to a color to communicate the 2D frequency/relative velocity information to the operator.

Frame rate is limited by the width of the window of interest and by the depth of the window, since the system must wait for the return echoes. This results in a Color Flow frame rate that is usually fairly low limiting the real time motion image in some clinical applications. However, since different colors are assigned to frequencies with the most energy, a clinically useful image is produced. The image shows flow directional differences, various velocities and flow characteristics which is useful for diagnosis, even though the frame refresh rate does not lend itself to good absolute velocity measurements.

PW/Color Flow transceiver electronics include a burst transmitter to drive the elements. The receive side incorporates an RF preamplifier followed by a sample volume gate. The signal is then split to be modulated with the sine and cosine of the transmit frequency. Each signal passes through a high pass “thump” filter, low pass filter to reduce noise, and through ADC into the processor (DSP). **See diagram on following page.**

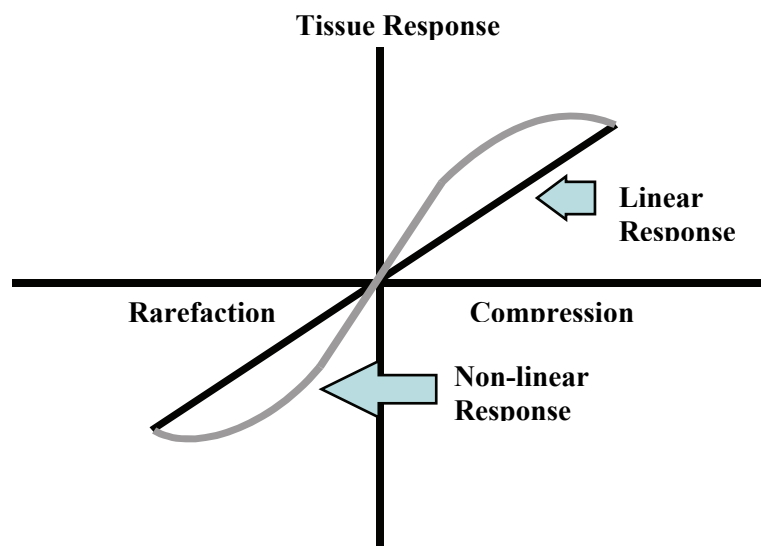


TISSUE HARMONIC IMAGING AND THREE DIMENSIONAL IMAGING

Modern advances in ultrasound imaging include tissue harmonic (THI) and three-dimensional imaging. Three-dimensional imaging is fairly straightforward and may be accomplished with a motor driving a standard array that produces two-dimensional images through the “z” axis. With additional signal processing a three dimensional image is displayed with the larger data set. It may also be achieved with a two dimensional array of crystals in the transducer head.

Harmonic imaging improves the resolution of ultrasound imaging and is possible due to some unique principals of physics as they apply to ultrasound (**see diagram below**). When a pure acoustic wave produced by the probe is high enough in intensity it becomes distorted as it passes through tissue. The waves are comprised of alternating positive pressure and negative pressure. Because water is essentially not compressible, and the human body is largely water, distortion occurs. This non-linear result produces frequency harmonics, which are useful.

The harmonic imaging mode filters out the fundamental transmit frequency and only analyzes the harmonic frequencies that are produced. Several benefits accrue; harmonics are higher frequencies and so have better resolution, harmonics are produced within the tissue and so only have to travel half the distance (one way, not round trip) which reduces attenuation, and by filtering out the transmit frequency, large near field return echoes, i.e. reverberation, are greatly diminished. Harmonic imaging requires more acoustic power to create the non-linear propagation, so it is not typically used – or needed – in imaging applications such as fetal studies. It is useful in studies such as cardiac imaging on large patients where near field echoes (reverberations) could result in poor imaging. In all cases the ALARA principal must be observed, power must be As Low As Reasonably Achievable.



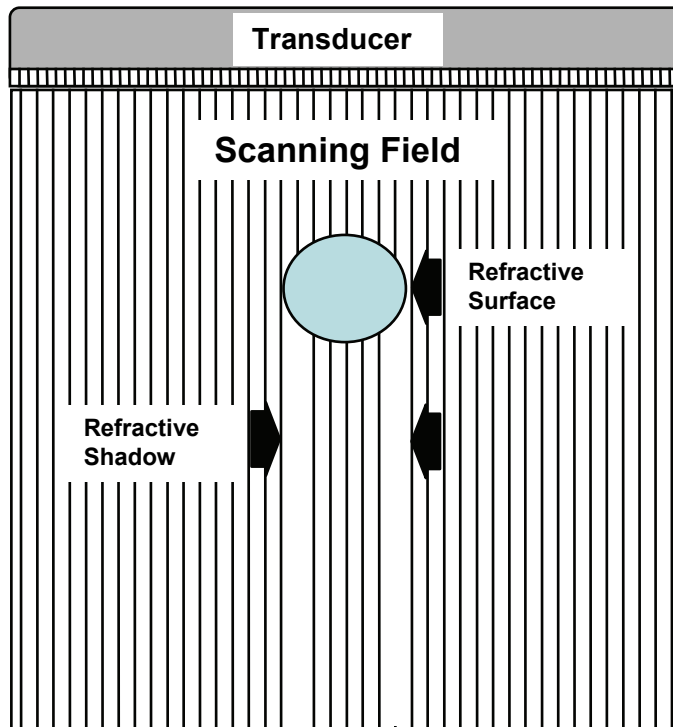
Non-linear versus linear tissue responses are central to tissue harmonic imaging (THI). Linear responses will disappear from the image in tissue harmonic imaging.

SYSTEM FAILURE MODES

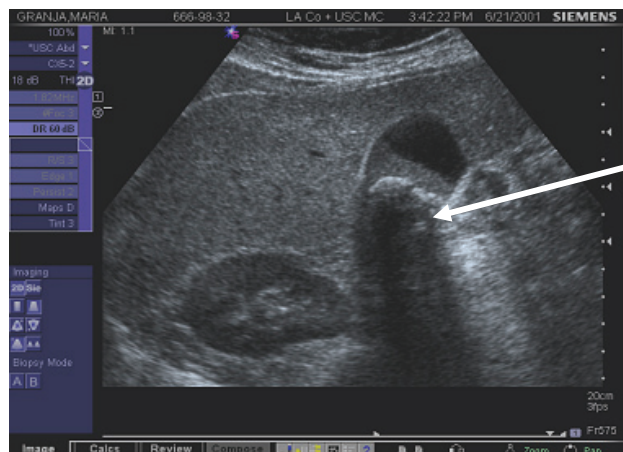
Failure modes will occur in many varieties: hard, soft, intermittent, software, and operator error. Hard failures are catastrophic and are generally fairly straightforward to diagnose. They can usually be traced to a certain board, power supply, monitor, or transducer, for example. Their symptoms (e.g., smoke coming out of a power supply!) will often help direct troubleshooting efforts. Be sure to round up the usual suspects: something that people touch, components that normally run hot, software, and connectors. The clinical operator usually provides clear reports on soft failures, e.g., broken knobs, casters, etc. Intermittent problems, software anomalies, and failure reports stemming from operator error are always more difficult to diagnose and require higher levels of troubleshooting.

Image artifacts may occur within the transducer field of view displayed on the monitor and do not represent actual anatomical structures (**see example diagram and clinical image on the following page**). They are often reported as system or transducer failures. However, image artifacts may also be caused by system limitations or operation, as well as the laws of physics. Artifact types include reverberation, attenuation, transducer, refraction, spectral, and color. They are often caused by too much power, too much power at a certain depth, muscle movement during imaging, acoustic shadowing such as the area behind a gall stone, large boundary mismatches, and noise interference from outside electromagnetic sources.

Refraction Boundary Shadowing Artifact Example



The basic event that contributes to this artifact is the refraction (Snell's law) of the incident ultrasound beams that intercept the cystic boundaries at a critical angle. Importantly, there must be a change in ultrasound propagation velocity at the interface.



Shadow
Artifact

SONORA SERVICE TRAINING INSTITUTE (SSTI)

Sonora is the industry's leading provider of ultrasound service technician and MRI training. Sonora holds general ultrasound service training and MRI classes. In addition to general training, Sonora offers system-specific training classes on GE, Toshiba, and Philips for MRI. And Philips (Agilent/HP) Sonos 4500/5500/7500 and ATL HDI 3000/5000; Siemens Aspen, Sequoia, and Acuson 128XP-10; and GE Vivid 5, Vivid 7, Voluson, Logiq 7, and Logiq 9; and the Philips iU22 and iE33 ultrasound systems.

Located at the base of the beautiful Rocky Mountains near Denver, Colorado, Sonora Medical Systems offers the world's premier ultrasound service training. Ranging from our popular basic ultrasound course to our system-specific courses, each student can receive the training relevant to their individual needs. Students also obtain all reference materials with extensive and accessible post-training technical support.

Sonora training delves into areas not taught at factory courses such as advanced probe maintenance, comprehensive system and board-level troubleshooting techniques, and Total Ultrasound Quality Assurance. Ongoing research and development efforts maintain our competitive advantage in system interfacing, software access and control, probe testing and repair as well as other advanced system maintenance techniques.

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