The discussion of the process by which sonographic images are created has so far been limited to utilizing a single crystal sampling a single line of sight. While this method of obtaining images does form the historical and technological foundation of sonographic technology, it does not accurately present the complexity of contemporary sonographic imaging systems. In modern-day systems, multiple piezoelectric crystals contained in a single probe are arranged, fired, and listened to in complex ways that create dynamic images and permit real-time interrogation of both human anatomical and hemodynamic states.

Real-time imaging is, in actuality, an automated B-scan imaging process in which the motion of the ultrasound beam is under mechanical or computer control. In such a scheme, the transducer – or, more commonly in contemporary imaging systems, the ultrasound beam – is moved repeatedly and automatically in pre-determined ways. As a result of this beam motion, a large number of scan lines can be produced by sweeping the sound beam through the patient's anatomy. Each of these scan lines provides echo information along a line of sight; by storing this information sequentially in the computer memory, a single two-dimensional image is produced during each sweep. When this process is repeated rapidly and the image is refreshed with each beam sweep, the illusion of motion is produced much as a series of still, sequential photographs will produce a movie when projected onto a screen. This display of echo information as it is received is called real-time, a term borrowed from other information-processing techniques that display information as it is received.

General Considerations

Real-time ultrasound imaging techniques have been the industry standard for nearly twenty years. In fact, they have become so integrated and accepted as the only method of performing ultrasound examinations that few sonographers stop to think about the actual nature and importance of dynamic imaging techniques. In comparing the acquisition and analysis of static sectional diagnostic images such as CT and MRI to dynamic imaging, several important differences are apparent. These differences are most notable in the area of the educational and psychomotor skills required by the individuals operating the equipment and acquiring the visual information that forms the basis of a clinical diagnosis or assessment. Examining a patient with real-time sonography is most similar to examining a patient with a radiographic fluoroscope; it requires constant, dynamic operator interaction with the patient and the structures being examined. And it requires independent judgment and critical thinking skills in acquiring and recording static images that will form the permanent record of this dynamic examination process.
Real-time imaging differs in several key ways to static imaging. These include:

**Demonstration of Organ Motion**

Observing motion during sonographic examination is important for two reasons: (1) certain types of normal and abnormal anatomy can only be appreciated by seeing organs in motion and (2) the need to suspend respiration for adequate examination is eliminated, therefore, patients who cannot suspend respiration can still be satisfactorily examined with real-time imaging. With static imaging alone, indistinct images or distortions of organ size and contour may occur because of organ motion during image acquisition.

Types of motion identifiable by real-time imaging are:

**Respiratory motion.** Typical examples of respiratory motion are the changes of diaphragmatic position between inspiration and expiration; or the excursion of one organ over the other, such as the movement of the liver over the kidneys. Absence of organ motion during respiration often implies a disease process. For example, lack of diaphragm motion may indicate phrenic nerve injury or an inflammatory process in the lung base or sub-diaphragmatic region. Failure of the liver or kidney to excursion normally with respiration suggests either an inflammatory or neoplastic process that adheres the organ to its surrounding structures.

**Vascular motion.** While it is frequently possible to recognize vascular structures by their size, location and anatomic relationship to adjacent parenchymal structures, this distinction is not always possible with two-dimensional imaging alone. While contemporary sonographic systems virtually all have pulsed and color Doppler imaging capabilities that allow easy discrimination of vascular from non-vascular structures, the rhythmic pulsatility of arteries, and the respiratory phasic change in the size of most veins, allow real-time differentiation from non-vascular structures when Doppler modalities are absent.

**Bowel motion.** The presence of peristalsis is important in distinguishing fluid-filled bowel loops from extraluminal pathologic collections such as abscesses or seromas in the abdomen. It is also frequently useful in differentiating normal pelvic intestinal structures from adnexal or cul de sac masses.

**Fetal motion.** The presence or absence of fetal cardiac activity on real-time sonography has become the gold standard in detecting fetal demise after the second trimester. Also, using endovaginal sonography, embryonic cardiovascular activity can be definitively demonstrated when the patient’s serum βhCG titers reach certain levels. Later in gestation, fetal limb and body motion can also be identified and quantitated as part of a fetal biophysical profile examination. Because real-time sonography is very reliable in demonstrating motion in obstetrics, it has come to play an integral role in providing care for pregnant patients.

**Anatomical Surveying**

All sonographic examination protocols should begin with a real-time survey of the body parts or cavities under investigation. By doing this, large anatomic areas can be rapidly surveyed to demonstrate normalcy or to detect and generally locate areas of pathology. It provides a “lay of the land” to the examiner. In performing this preliminary survey, the time
required for a complete sonographic examination can be markedly reduced as it frequently
can direct the examiner’s attention to the areas of interest. Surveying prior to beginning the
formal examination protocol also enables the operator to optimize imaging parameters and
to prepare a strategy for conducting a complete and tailored study individualized to each
patient.

**Optimizing Imaging Windows**

The ability to optimally visualize organ surfaces and structural interfaces depends upon
the correct orientation of the ultrasound beam relative to the area or structure of interest. Because specular reflections provide the strongest echo information, visualization of a
structure is best when the beam is oriented perpendicular (90°) to its surface. A change in
beam orientation of even several degrees from the perpendicular can produce a significant
decrease in intensity of returning signals and, in many instances, significant image quality
degradation. Since the angle and position of the sound beam can be freely and subtly
adjusted during real-time imaging, optimal beam orientation can be accomplished by
adjusting the transducer position in a variety of ways. Basic real-time transducer
movements include:

- **Moving**: sliding the transducer longitudinally or side to side
- **Rotating**: turning the transducer along its central axis without changing position
- **Heel-toe**: rocking, or slightly angling, the transducer along its long axis without
  changing position
- **Angling**: pointing the transducer top-bottom or right-left without changing position

**Real-Time Transducers**

Key to the development and application of real-time sonographic imaging systems in
clinical practice were alterations and improvements made to the single-crystal transducers
utilized in older, static B-scan systems. A transducer is simply a device that converts one
type of energy to another. In real-time imaging, the device that emits and receives the
ultrasound pulses is more properly called a probe. A probe incorporates hundreds of
individual transducers (piezoelectric crystals), as well as other hardware, into a complex
and sophisticated design that is engineered to emit a highly controlled and precise
ultrasound beam. Essentially, the transducers are the crystals within the probe.

Real-time probes assume many shapes and designs, and each may be specifically
engineered for particular clinical applications. Probes are specially designed for
extracorporeal applications such as cardiac, abdominal, vascular, small parts, etc.
Intracavitary probes may be used in endovaginal (EV), transrectal or transesophageal (TE)
applications. Despite the multitude of possible clinical applications and designs of
sonography probes, there are two basic types of real time transducers: **MECHANICAL** and
**ELECTRONIC**. Each produces “live-action” scanning; each, however, employs a different
method for achieving it.
Chapter 5. Real-time Scanning

Simply stated, if a probe has moving parts that oscillate the crystal or other components within the scanhead, it is a **MECHANICAL** probe. If there are no moving parts and the illusion of motion is produced by electronic steering and moving of the beam, it is an **ELECTRONIC** or **PHASED** probe.

**Mechanical Probes**

The simplest type of real time transducer device is a mechanical sector probe that contains a single frequency piezoelectric element. The crystal is attached to a motor that mechanically moves it back and forth fast enough to produce enough scan lines in a short enough period of time to create the illusion of motion. The ultrasound beam geometry is that of a sector, and it is usually swept in a given arc at a single or variable frame rate. Another configuration of a mechanical sector probe is the **rotating head** transducer assembly. In this design, one or several transducer crystals are mounted on a cylinder that is rapidly rotated over 360° using a small servo-controlled motor. The cylinder sits in a fluid-filled chamber that allows controlled rotation of the cylinder and transducers. The fluid path container cap is often shaped to conform to the circular motion of the rotor.

Another mechanical method of moving the ultrasound beam consists of mounting the transducer crystal in the scanhead and holding it steady. A rotating or oscillating mirror is positioned so that as the crystal emits the sound beam, it is reflected off the mirror and directed out of the probe.

Other configurations of real-time probes have been used over the years to produce mechanical real-time scans, the commonality of all these methods is the presence of moving parts to steer the ultrasound beam.

**FOCUSBING**

Focusing the beam in a mechanical sector probe is similar to focusing the beam in a single element, static scan transducer. This is usually accomplished by using a curved crystal or an acoustic lens and creates a beam with a fixed focal length that is specific for each probe. To change the focal length, it is necessary to change probes. In any individual mechanical probe, the focus is fixed and not electronically changeable.
In electronic real-time ultrasound probes, a series of piezoelectric crystals are housed in very close proximity within a single scan-head. This is called an array. Each crystal in the array is wired to the pulser and the receiver so that each may be fired and silenced individually. By applying sophisticated timing patterns to the array of crystals, precise beam patterns can be created due to Huygens principle. Simply stated, Huygens principle posits that a wave front emanating from multiple sound sources travels as a unified whole and is the result of complex patterns of constructive interference between the individual wavelets. The timing patterns used to produce this wavefront typically consist of varying the firing phase of the crystals and, therefore, these devices are frequently called phased-array probes. Unlike mechanical sector probes, phased-array probes contain no moving parts; the sweep, focusing and steering of the beam are accomplished by controlling and altering the timing among transducer elements, and focal length can be changed by the operator. There are a variety of types of electronic probes and they include:

**SEQUENTIAL LINEAR ARRAY**

Sequential linear arrays are simple types of electronic real-time probes and are rarely found in clinical practice these days. Historically, linear arrays were the first type of electronic real-time probe manufactured and consist of a series of transducer elements aligned in a straight line. The firing sequence is simple; each crystal is fired sequentially down the line, resulting in a linear sweep of the ultrasound beam and the formation of a rectangular shaped image, or field of view (FOV). Each transducer element creates a single line of sight. Linear arrays use conventional, mechanical focusing techniques, therefore, the focal length is fixed for each probe.
SEGMENTAL LINEAR ARRAY
Similar to a sequential linear array probe, the segmental linear array probe fires a groups of several elements almost simultaneously to form a single wavefront.

CONVEX SWITCHED ARRAY
Similar to a linear array, a convex array has a series of piezoelectric elements arranged in a line, but the face of the probe is curved into an arc shape. The crystals are fired sequentially and create a sector-shaped image with a blunted footprint. Since returning echoes are detected more readily when they fall perpendicularly, rather than obliquely, on each crystal face, a convex array probe has higher sensitivity and preserves resolution at the sides of the sector. As with linear arrays, convex arrays are mechanically focused and the focal depth is not changeable by the operator.

ANNULAR ARRAY
Circular-shaped piezoelectric crystals arranged concentrically within the scanhead are used in an annular array probe. Such a transducer has five to ten of these concentric crystals that are energized sequentially from inside out. Each concentric crystal produces a focused beam; the smaller, inner circular elements focus in the near field; the larger, outer elements provide focus in the far field. By using multiple firings of each crystal, a form of electronic focusing can be achieved, with variable focusing of the ultrasound beam along the central axis. Like the mechanical sector probes, annular arrays are rarely found in contemporary clinical sonography.
Phased Array Probes

Phased array real-time probes all contain a group of piezoelectric crystals arranged in a row, but the beam geometry and field of view vary based on timing sequence of crystal excitation. The description of the probe is usually based on the appearance of the field-of-view created by the phased firing of the crystals. If the image is pie-shaped, it is a **phased sector** probe; if the image is rectangular in shape it is called a **phased linear** probe. When the crystals are aligned in a **curved linear** configuration, the probe may be designated as such or as a **convex** or **curvilinear** probe. One type of ultrasound probe combines characteristics of both sequential linear and phased array technology. These devices are called **vector** probes.

**DYNAMIC FOCUSING**

Like the single element transducer, a phased array probe must focus inside its effective near field. The degree of focusing is still partly a function of the transducer diameter and the distance of the focal point from the face of the array. However in phased array probes, variations in the timing and number of crystals fired allow the beam to be focused and steered in ways that dramatically improve imaging applications and quality.

Almost all contemporary real-time ultrasound imaging systems now employ a dual type of focusing: focusing when the beam is transmitted from the receiver, and focusing when the echoes are received after interacting with human soft tissue.

**TRANSMIT FOCUSING**

Focusing during transmission is produced by introducing delays in the excitation pulse to each element of the array. If the elements are pulsed in an arc or curved sequence, the wavefront diagonal versus curved will have a concave shape and will converge to a focal point. Altering the time delays in the firing sequence of the elements can produce a change in the focal depth. As the firing delay between the center and the outer elements of the array increases, the focal point decreases, or is closer to the transducer. Conversely, as the timing delay decreases, the focal point will be deeper. (See accompanying schematic) In general, a curved configuration of the electronic signals transmitted to the array will produce a focused beam.

Transmit focusing can also be accomplished by changing the number of crystals fired at one time. For example, rather than firing five crystals as a group, two crystals could be fired together to produce a narrower beam with a shorter near field. Typically, when larger groups of crystals are fired together, the focal range increases.

Electronic focusing created by applying a “curved” configuration to the excitation pulses to the arrayed crystals.
Chapter 5. Real-time Scanning

BEAM STEERING
Just as altering the firing sequence of the piezoelectric elements in a curved configuration will create a focused beam, alteration in a diagonal, or sloped, format will create beam steering. Beam steering provides many advantages in producing high-resolution two-dimensional sonographic images by increasing the specular echoes returning from the body. Beam steering also enhances echo and scattering information used in producing pulsed and color Doppler displays. The accompanying schematics demonstrate the relationship between sloped signal transmission and the resultant beam steering.
RECEIVE FOCUSING

Just as sound beams diverge when they propagate away from the transducer after crystal excitation, they also diverge after reflecting from an acoustic interface and return to the transducer. Focusing the received echoes helps improve the overall resolution of the image. Similar to focusing on transmit, receive focusing uses time delays. The echoes received from the body can be selectively assigned to a focal region and a line of sight resulting in better image quality.

MULTIPLE ZONE FOCUSING

Multi-zone transmit focusing allows several levels of focusing to occur in a single image. To achieve this, multiple pulses are sent out from each transducer element, and each one is focused at a different depth. While the overall resolution and image quality is significantly improved using multiple focal zones, the image acquisition and display rate are slowed because of the increased amount of time necessary to create each frame of information. Data must be collected for all the lines of sight for each individual pulse before the next series of pulses can be transmitted.

APODISATION

Another technique to improve the overall quality of an ultrasound image is one called apodisation. As will be discussed in the section on image artifacts, most of the ultrasound beam energy is concentrated in a main beam lobe; however, there can be some leakage of power into little pools of ultrasound energy hanging off each side of the transducer, which are called side lobes. Side lobes can lead to spurious echo registration that can cause artifactual information to be displayed on the image. A process called apodisation can reduce the intensity of these side lobes. This is achieved by reducing the amplitude of the vibration of the crystals toward the edge of each transducer crystal in the transmit mode, and by reducing the sensitivity toward the edge in the receive mode. While this process is typically used in annular arrays, it can be tailored and can be applied to other types of transducer arrays.
Image Creation

PULSES, LINES AND FRAMES

Time is an important factor when considering the production of real-time ultrasound images. Since individual, single images must be produced rapidly enough to give the illusion of motion, the amount of time required to build a single image directly affects scanning ability. Like a static B-scanner, a real time system builds a two dimensional image from a series of individual pulse-listen cycles, storing the acquired data on each cycle. Each pulse produced by the ultrasound scanner creates a single line on the display. If, for example, there are 120 lines per image, then there must be 120 separate pulses processed to create a single static image. This would theoretically be accomplished by incorporating 120 separate crystals into the probes. Using the soft-tissue average propagation velocity of 1,540 m/sec, the echo transit time is 13 μs for each centimeter of tissue depth for each pulse. Obviously, increasing the depth of field will increase the pulse listen time. Consequently, the maximum time it takes for the most distant echo to reach the transducer in any field of view can be represented by the following:

\[
\text{PLT} = 13 \text{ μs/cm} \times \text{FOV}
\]

Where: \( \text{PLT} = \) pulse listen time
\( \text{FOV} = \) field of view (depth in cm)

Example: The pulse listen time for a 10 cm field of view is:

\[
13 \text{ μs/cm} \times 10 \text{ cm} = 130 \text{ μs per line.}
\]

This formula calculates listen time for a single element. If there are \( n \) number of lines of sight in a single image then, the time it takes to build a complete frame (or single picture) is called the:

\[
\text{FT} = \text{PLT} \times n
\]

Where: \( \text{FT} = \) frame time
\( \text{PLT} = \) pulse listen time
\( n = \) number of lines of sight

Continuing the example from above, if there are 120 lines in a single image, the time to build a single frame then will be:

\[
\text{FT} = 130 \text{ μs} \times 120
\]
\[= 15,600 \text{ μs}
\]
\[= 0.0156 \text{ s}
\]

If it takes this long to build a single image, then the maximum rate at which the image can be renewed is the reciprocal of the frame time and is called the frame rate.
Physical Principles of General and Vascular Sonography

Chapter 5. Real-time Scanning

\[ FR = \frac{1}{FT} \]

where: \( FR = \) frame rate
\( FT = \) frame time

If one frame requires 0.0156 seconds of time to be created and displayed, then (still using the example above) in one full second, 64 \( (1 \div 0.0156) \) images can be displayed. This means that 64 images can be flashed across the screen each second. Calculations will show that as the field of view, or depth, increases, the maximum available frame rate decreases. In order to produce a flicker free image, at least 25 frames per second must be displayed. This is known as the **Flicker Fusion Rate**.

In dynamic focusing, the focal zone is moved to several positions within the field of view, and the image is constructed only from the signals contained in the focal zone. If a system utilizes four of these dynamic focusing zones, for example, then each line of sight in the display is created from four separate pulse-listen cycles, where the focal zone is moved into a new region on each cycle. Reviewing the above calculations, dynamic focusing over four zones would quarter our frame rate to 16 Hz (frames per second). Dynamic focusing, then, can provide high-resolution images, but the sacrifice entails a lowered frame rate and a lessened ability to detect motion within the real time system. The ability to accurately display motion as it occurs over time in an imaging system is called **Temporal Resolution**.


Chapter 5. Real-time Scanning

Exercises

1. Calculate the frame rate of a real time system that has 128 channels and is set to display a 20 cm deep image.

2. List four types of body motions identifiable with real-time ultrasound.

3. A "diagonal" firing sequence applied to phased array probe will result in beam ________.

4. A "curved" firing sequence applied to phased array probe will result in beam ________.

5. Explain the difference between frame rate and frame time.

6. What is apodisation?
Chapter 5. Real-time Scanning

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Types of motion identifiable by real-time imaging are:

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Real-time probes assume many shapes and designs, and each may be specifically engineered for particular clinical applications. Probes are specially designed for extracorporeal applications such as cardiac, abdominal, vascular, small parts, etc. Intracavitary probes may be used in endovaginal (EV), transrectal or transesophageal (TE) applications. Despite the multitude of possible clinical applications and designs of sonography probes, there are two basic types of real time transducers: **MECHANICAL** and **ELECTRONIC**. Each produces “live-action” scanning; each, however, employs a different method for achieving it.
Simply stated, if a probe has moving parts that oscillate the crystal or other components within the scanhead, it is a MECHANICAL probe. If there are no moving parts and the illusion of motion is produced by electronic steering and moving of the beam, it is an ELECTRONIC or PHASED probe.

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The simplest type of real time transducer device is a mechanical sector probe that contains a single frequency piezoelectric element. The crystal is attached to a motor that mechanically moves it back and forth fast enough to produce enough scan lines in a short enough period of time to create the illusion of motion. The ultrasound beam geometry is that of a sector, and it is usually swept in a given arc at a single or variable frame rate.

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**FOCUSING**

Focusing the beam in a mechanical sector probe is similar to focusing the beam in a single element, static scan transducer. This is usually accomplished by using a curved crystal or an acoustic lens and creates a beam with a fixed focal length that is specific for each probe. To change the focal length, it is necessary to change probes. In any individual mechanical probe, the focus is fixed and not electronically changeable.
In electronic real-time ultrasound probes a series of piezoelectric crystals are housed in very close proximity within a single scan-head. This is called an array. Each crystal in the array is wired to the pulser and the receiver so that each may be fired and silenced individually. By applying sophisticated timing patterns to the array of crystals, precise beam patterns can be created due to Huygens principle. Simply stated, Huygens principle posits that a wave front emanating from multiple sound sources travels as a unified whole and is the result of complex patterns of constructive interference between the individual wavelets. The timing patterns used to produce this wavefront typically consist of varying the firing phase of the crystals and, therefore, these devices are frequently called phased-array probes. Unlike mechanical sector probes, phased-array probes contain no moving parts; the sweep, focusing and steering of the beam are accomplished by controlling and altering the timing among transducer elements, and focal length can be changed by the operator. There are a variety of types of electronic probes and they include:

**SEQUENTIAL LINEAR ARRAY**

Sequential linear arrays are simple types of electronic real-time probes and are rarely found in clinical practice these days. Historically, linear arrays were the first type of electronic real-time probe manufactured and consist of a series of transducer elements aligned in a straight line. The firing sequence is simple; each crystal is fired sequentially down the line, resulting in a linear sweep of the ultrasound beam and the formation of a rectangular shaped image, or field of view (FOV). Each transducer element creates a single line of sight. Linear arrays use conventional, mechanical focusing techniques, therefore, the focal length is fixed for each probe.
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Similar to a sequential linear array probe, the segmental linear array probe fires a group of several elements almost simultaneously to form a single wavefront.

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Similar to a linear array, a convex array has a series of piezoelectric elements arranged in a line, but the face of the probe is curved into an arc shape. The crystals are fired sequentially and create a sector-shaped image with a blunted footprint. Since returning echoes are detected more readily when they fall perpendicularly, rather than obliquely, on each crystal face, a convex array probe has higher sensitivity and preserves resolution at the sides of the sector. As with linear arrays, convex arrays are mechanically focused and the focal depth is not changeable by the operator.

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Circular-shaped piezoelectric crystals arranged concentrically within the scanhead are used in an annular array probe. Such a transducer has five to ten of these concentric crystals that are energized sequentially from inside out. Each concentric crystal produces a focused beam; the smaller, inner circular elements focus in the near field; the larger, outer elements provide focus in the far field. By using multiple firings of each crystal, a form of electronic focusing can be achieved, with variable focusing of the ultrasound beam along the central axis. Like the mechanical sector probes, annular arrays are rarely found in contemporary clinical sonography.
Chapter 5. Real-time Scanning

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Phased array real-time probes all contain a group of piezoelectric crystals arranged in a row, but the beam geometry and field of view vary based on timing sequence of crystal excitation. The description of the probe is usually based on the appearance of the field-of-view created by the phased firing of the crystals. If the image is pie-shaped, it is a phased sector probe; if the image is rectangular in shape it is called a phased linear probe. When the crystals are aligned in a curved linear configuration, the probe may be designated as such or as a convex or curvilinear probe. One type of ultrasound probe combines characteristics of both sequential linear and phased array technology. These devices are called vector probes.

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Like the single element transducer, a phased array probe must focus inside its effective near field. The degree of focusing is still partly a function of the transducer diameter and the distance of the focal point from the face of the array. However in phased array probes, variations in the timing and number of crystals fired allow the beam to be focused and steered in ways that dramatically improve imaging applications and quality.

Almost all contemporary real-time ultrasound imaging systems now employ a dual type of focusing: focusing when the beam is transmitted from the receiver, and focusing when the echoes are received after interacting with human soft tissue.

TRANSMIT FOCUSING

Focusing during transmission is produced by introducing delays in the excitation pulse to each element of the array. If the elements are pulsed in an arc or curved sequence, the wavefront diagonal versus curved will have a concave shape and will converge to a focal point. Altering the time delays in the firing sequence of the elements can produce a change in the focal depth. As the firing delay between the center and the outer elements of the array increases, the focal point decreases, or is closer to the transducer. Conversely, as the timing delay decreases, the focal point will be deeper. (See accompanying schematic) In general, a curved configuration of the electronic signals transmitted to the array will produce a focused beam.

Transmit focusing can also be accomplished by changing the number of crystals fired at one time. For example, rather than firing five crystals as a group, two crystals could be fired together to produce a narrower beam with a shorter near field. Typically, when larger groups of crystals are fired together, the focal range increases.

Electronic focusing created by applying a “curved” configuration to the excitation pulses to the arrayed crystals.
BEAM STEERING
Just as altering the firing sequence of the piezoelectric elements in a curved configuration will create a focused beam, alteration in a diagonal, or sloped, format will create beam steering. Beam steering provides many advantages in producing high-resolution two-dimensional sonographic images by increasing the specular echoes returning from the body. Beam steering also enhances echo and scattering information used in producing pulsed and color Doppler displays. The accompanying schematics demonstrate the relationship between sloped signal transmission and the resultant beam steering.
RECEIVE FOCUSING
Just as sound beams diverge when they propagate away from the transducer after crystal excitation, they also diverge after reflecting from an acoustic interface and return to the transducer. Focusing the received echoes helps improve the overall resolution of the image. Similar to focusing on transmit, receive focusing uses time delays. The echoes received from the body can be selectively assigned to a focal region and a line of sight resulting in better image quality.

MULTIPLE ZONE FOCUSING
Multi-zone transmit focusing allows several levels of focusing to occur in a single image. To achieve this, multiple pulses are sent out from each transducer element, and each one is focused at a different depth. While the overall resolution and image quality is significantly improved using multiple focal zones, the image acquisition and display rate are slowed because of the increased amount of time necessary to create each frame of information. Data must be collected for all the lines of sight for each individual pulse before the next series of pulses can be transmitted.

APODISATION
Another technique to improve the overall quality of an ultrasound image is one called apodisation. As will be discussed in the section on image artifacts, most of the ultrasound beam energy is concentrated in a main beam lobe; however, there can be some leakage of power into little pools of ultrasound energy hanging off each side of the transducer, which are called side lobes. Side lobes can lead to spurious echo registration that can cause artifactual information to be displayed on the image. A process called apodisation can reduce the intensity of these side lobes. This is achieved by reducing the amplitude of the vibration of the crystals toward the edge of each transducer crystal in the transmit mode, and by reducing the sensitivity toward the edge in the receive mode. While this process is typically used in annular arrays, it can be tailored and can be applied to other types of transducer arrays.
Chapter 5. Real-time Scanning

Image Creation

PULSES, LINES AND FRAMES

Time is an important factor when considering the production of real-time ultrasound images. Since individual, single images must be produced rapidly enough to give the illusion of motion, the amount of time required to build a single image directly affects scanning ability. Like a static B-scanner, a real time system builds a two dimensional image from a series of individual pulse-listen cycles, storing the acquired data on each cycle. Each pulse produced by the ultrasound scanner creates a single line on the display. If, for example, there are 120 lines per image, then there must be 120 separate pulses processed to create a single static image. This would theoretically be accomplished by incorporating 120 separate crystals into the probes. Using the soft-tissue average propagation velocity of 1,540 m/sec, the echo transit time is 13 μs for each centimeter of tissue depth for each pulse. Obviously, increasing the depth of field will increase the pulse listen time. Consequently, the maximum time it takes for the most distant echo to reach the transducer in any field of view can be represented by the following:

\[
\text{PLT} = 13 \, \mu \text{s/cm} \times \text{FOV}
\]

Where: \( \text{PLT} = \) pulse listen time
\( \text{FOV} = \) field of view (depth in cm)

Example: The pulse listen time for a 10 cm field of view is:
\[
13 \, \mu \text{s/cm} \times 10 \, \text{cm} = 130 \, \mu \text{s per line.}
\]

This formula calculates listen time for a single element. If there are \( n \) number of lines of sight in a single image then, the time it takes to build a complete frame (or single picture) is called the:

\[
\text{FT} = \text{PLT} \times n
\]

Where: \( \text{FT} = \) frame time
\( \text{PLT} = \) pulse listen time
\( n = \) number of lines of sight

Continuing the example from above, if there are 120 lines in a single image, the time to build a single frame then will be:
\[
\begin{align*}
\text{FT} &= 130 \, \mu \text{s} \times 120 \\
&= 15,600 \, \mu \text{s} \\
&= 0.0156 \, \text{s}
\end{align*}
\]

If it takes this long to build a single image, then the maximum rate at which the image can be renewed is the reciprocal of the frame time and is called the frame rate.

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**Frame Rate**

\[
FR = \frac{1}{FT}
\]

where:
- \( FR = \) frame rate
- \( FT = \) frame time

If one frame requires 0.0156 seconds of time to be created and displayed, then (still using the example above) in one full second, 64 \((1 \div 0.0156)\) images can be displayed. This means that 64 images can be flashed across the screen each second. Calculations will show that as the field of view, or depth, increases, the maximum available frame rate decreases. In order to produce a flicker-free image, at least 25 frames per second must be displayed. This is known as the **flicker fusion rate**.

In dynamic focusing, the focal zone is moved to several positions within the field of view, and the image is constructed only from the signals contained in the focal zone. If a system utilizes four of these dynamic focusing zones, for example, then each line of sight in the display is created from four separate pulse-listen cycles, where the focal zone is moved into a new region on each cycle. Reviewing the above calculations, dynamic focusing over four zones would quarter our frame rate to 16 Hz (frames per second). Dynamic focusing, then, can provide high-resolution images, but the sacrifice entails a lowered frame rate and a lessened ability to detect motion within the real-time system. The ability to accurately display motion as it occurs over time in an imaging system is called **temporal resolution**.
Exercises

1. Calculate the frame rate of a real time system that has 128 channels and is set to display a 20 cm deep image.

2. List four types of body motions identifiable with real-time ultrasound.

3. A "diagonal" firing sequence applied to phased array probe will result in beam ________.

4. A "curved" firing sequence applied to phased array probe will result in beam ________.

5. Explain the difference between frame rate and frame time.

6. What is apodisation?