Biomedical Ultrasound: Fundamentals of Imaging and Micromachined Transducers

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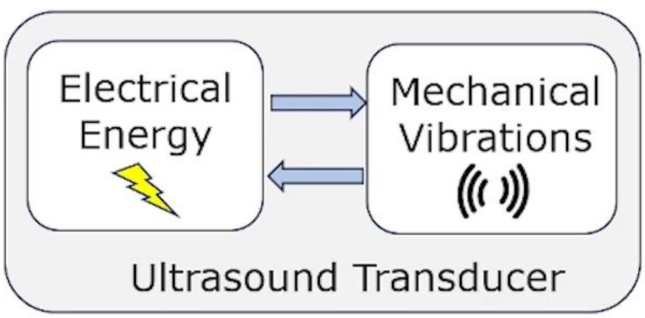
Lecture: 10

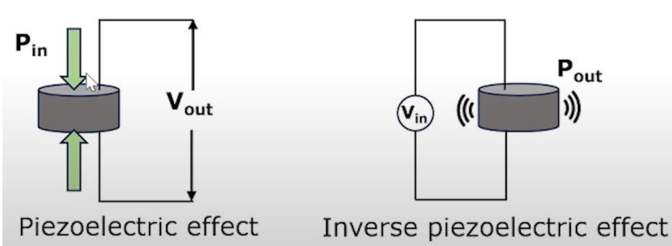
Transducers and transducer arrays

Welcome to this lecture on transducers and transducer arrays. I am Professor Himanshu Shekhar from IIT, Gandhinagar. So, you have already heard about ultrasound transducers. Transducers convert variations in a physical quantity into an equivalent electrical signal or vice versa. And you have heard about the piezoelectric effect and the inverse piezoelectric effect. Transducers in ultrasound rely on electromechanical energy conversion.

So, if you give electrical energy to an ultrasound transducer, it will result in mechanical vibrations. And conversely, when mechanical vibrations are sensed by the transducer, they can be converted to electrical energy.



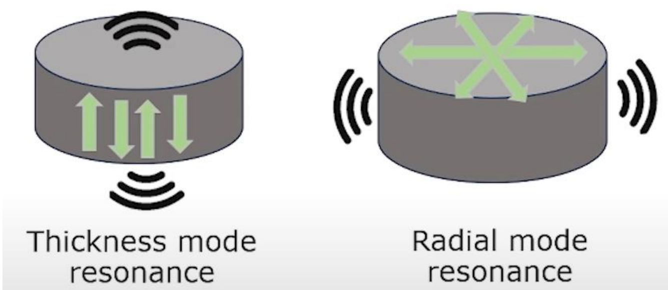
So here is a schematic in which a voltage is generated across a piezoelectric element when some kind of pressure is applied across its planes. And similarly, if you apply a voltage to a crystal, then it will generate a pressure field or a strain across it, which is the inverse piezoelectric effect.



We also discussed that there are some natural piezoelectric crystals, including quartz, and there are other synthetic crystals such as lead zirconium titanate and PVDF, which is a membrane polymer material.

So, let's discuss the resonance modes of ultrasound transducer crystals. So, these piezoelectric materials are actually capacitive in nature, and they exhibit resonance at certain frequencies. So, the resonance of these crystals can be of two types. So, imagine this type of crystal geometry.

It's a disc. when you provide strain to it than it can generate resonance in either thickness mode which is the first case in which the pressure will be transmitted in this manner, or it can be in the radial mode in which the oscillations happen in this manner. So, for ultrasound, we would like the pressure to be propagated along the axis of the transducer and therefore we prefer the thickness mode resonance and not the radial mode resonance. So, the thickness mode resonant frequency can actually be calculated by knowing the speed of the sound in the crystal. So, this you would know from the material properties of the crystals.



It can be measured also. And then the thickness of the crystal will be chosen in such a way that you get the right resonant frequency. So, for example, if you want a transducer resonant at a frequency of f, you know the speed of sound in the crystal material, then using the formula C equal to F lambda you can get the wavelength lambda and then you can set the thickness of the crystal to lambda by 2. In this way you will be able to get a crystal resonant at a particular frequency.

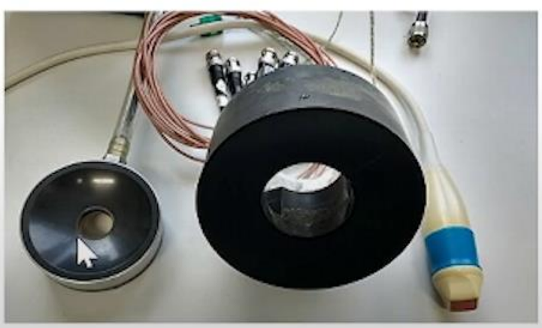
$$Thickness\ mode\ resonant\ frequency = \frac{c_{crystal}}{2\ x\ thickness}$$

So here is an image of an ultrasound transducer. If you have visited the clinic, you may have seen this. And there are several different types of ultrasound transducers. There are single element transducers which have a single crystal. It could be a small crystal, or it could be a larger aperture crystal. Then there are linear array transducers which are most common and here the transducer that you see it is composed of a large number of small crystals and therefore the name linear array.

"File:UltrasoundProbe2006a.jpg" by The original uploader was Drickey at English Wikipedia. is licensed under CC BY-SA 2.5.

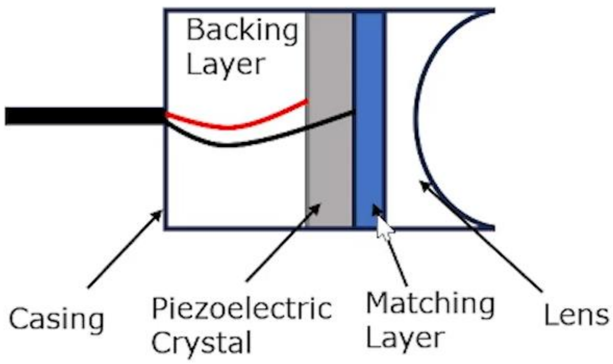
There are curvilinear arrays which are like linear arrays, but they have a curve aperture and then there are also phased array transducers about which we will discuss a little bit later.

And here are some large transducers which are used in therapeutic ultrasound. These are not the subject of this particular course, but just to show you that there can also be single element large aperture transducers for used in medical applications.



"HIFU Transducers Compared with Cardiac Imaging Probe" by Chris Adams is licensed under CC BY-SA 4.0.(public domain image)

So here is a schematic of a single element transducer. You first can see this in grey is the piezoelectric crystal and across the interfaces of the piezoelectric crystal we can apply the voltage and then that will make this crystal oscillate.



So as this crystal oscillates, there's a matching layer. The purpose of this matching layer is impedance matching. We'll discuss more on this later. And then there's a lens. Here you see a concave lens, which is used for focusing.

So, in ultrasound, the focusing lens properties are a little bit different from optics. In optics, generally you know that if you want to focus, we use a convex lens. But here we will see that a concave lens will serve the same purpose. So more on that later. And then behind the piezoelectric crystal you have something called the backing layer.

So literally it is in the back of the piezoelectric crystal, and this can either be just air filled or it is filled with some kind of epoxy or dense material which we will see in a minute. So, the matching layer, like I said, reduces impedance mismatch between the layers, the crystal and the lens and the tissue. And it helps in transferring maximum mechanical power. It avoids the reflection of energy back into the transducer material. The lens helps focus the wave to get better resolution in our images.

And the backing layer, it actually controls the manner in which these oscillations happen. So, we'll discuss that more now. So why do we really need a matching layer? Well, the acoustic impedance of PZT, lead zirconium tetanate, a particular very commonly used piezoelectric material, is about 30 Mega Rayls. So, the unit of impedance, as you will recall, is Mega Rayls, and Rayl is after Lord Rayleigh, a very prominent contributor in the field of acoustics. So, the acoustic impedance of this crystal is 30 Mega Rayls, but the average acoustic impedance of tissue is only about 2 Mega Rayls.

So, when there is acoustic impedance mismatch, what do you expect? The energy will be reflected, and very little energy will actually be transmitted. So to avoid this what we do is we add a third layer in between these two layers meaning in between the crystal and the tissue which is called a matching layer and this matching layer improves the coupling of the crystal and the tissue material and in this way if you just had this two media so if the signal is originating in the transducer and it is propagating into the tissue the large majority of the signal will be reflected back and thus this is not a very efficient design but we could add a matching layer and after adding this matching layer the majority of the energy is actually coupled through the matching layer into the medium. So, you may have heard about this impedance matching also in the context of electrical engineering and in circuits. So how do we design this matching layer? The impedance of the matching layer has to be intermediate compared to the input and output layers for minimal reflection. So here the input and output layers would mean the layer through which the input is passing, that is the crystal, and the output is tissue in this case.

So, for a perfect matching layer, the impedance of the matching layer has to be the square root of Z_1 and Z_2, where Z_1 and Z_2 are the acoustic impedance of the input and output layers, in this case the crystal and the tissue.

$$Z_{mt} = \sqrt{Z_1 x\, Z_2}$$

And also the thickness of the matching layer is typically maintained at lambda by 4. This is what enables impedance matching. There are mathematical proofs which show that if the layer is lambda by 4 and the impedance is square root of Z_1, Z_2, then you will get perfect matching. Now it turns out if lambda is too small, say for example, lambda can be of sub-millimeter dimensions for Megahertz transducers, then lambda by four in some cases will become too thin or too flimsy.

$$Optimum\ Thickness = \frac{\lambda}{4}$$

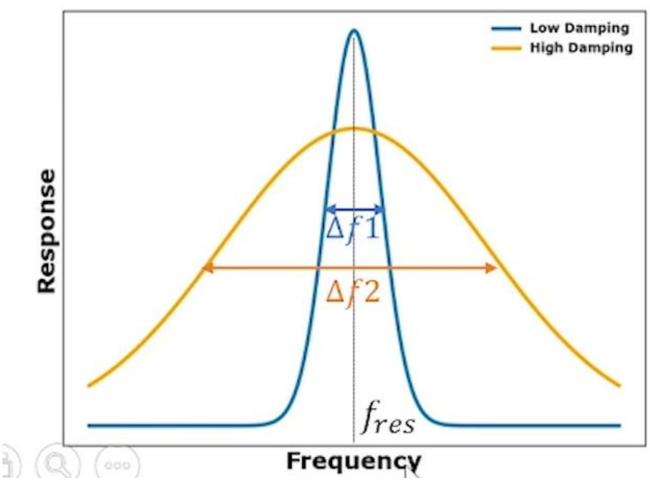
In such cases, we go for n lambda by four, where n can be an integer. And the advantage here is you get better resistance to strain, but the disadvantage is now there'll be more attenuation as the wave passes through this matching layer. So, some commonly used materials for matching layers are alumina epoxy composite, tungsten epoxy composite, and graphite. Now in practice, we need typically two matching layers because a single matching layer with the right properties is not really available because you need a material which has this kind of impedance which is square root of Z_1 Z_2 and these kind of materials are not naturally available and it is also difficult to synthesize them so some have jokingly called these materials unobtanium and for that reason, typically two matching layers, two layers are applied and so the matching happens in a cascaded manner. But nonetheless, from a conceptual point of view, we can think of matching layer as a single layer with a quarter wavelength thickness and an impedance of square root Z_1, Z_2.

Let us discuss the backing layer now. So, ultrasound transducers are typically driven by continuous sinusoids or short bursts based on the application. Typically for imaging, short bursts are used. Now the transducer or the crystal exhibits a ringing behavior at resonance. What do I mean by ringing behavior? Well, ringing is a byproduct of the inertia of the crystal.

For example, if I have an ultrasound crystal and I drive it using a two-cycle sine burst, even after these two cycles have ended, the transducer continues to oscillate because of its inertia before these oscillations are damped out. So, this is actually not very good for imaging because you would want to have a transducer which quickly damps so that now you can receive the signal from the tissue. So now the extent of ringing can be adjusted by selecting a suitable backing layer for damping the crystal oscillations. So, like I said, for continuous wave applications, we don't care that much about ringing.

Ringing is acceptable. What we need is very good power transfer for continuous wave applications. So, therefore, we go for either no backing, which means its air backing or some lightly backed transducers. On the other hand, if we consider pulse wave applications which are used for imaging, ringing is not preferred because we would like the transducer to be able to receive the signals immediately after the transmit is over. And for such transducers, we have a heavy backing material which has high damping. So, this damping leads to some loss of energy, but this produces broadband transducers.

And the backing material which is typically common for these types of transducers is either epoxy or rubber. So here we look at the quality factor of our crystal and if the resonance is sharp, so on the x-axis you see the frequency and on the y-axis you see the crystal response.



So, if the resonance is very sharp in the vicinity of the resonance frequency, then we say the transducer has high quality factor, which can be computed as the resonant frequency by the width of this resonance peak.

$$Quality\ factor = \frac{f_{res}}{\Delta f}$$

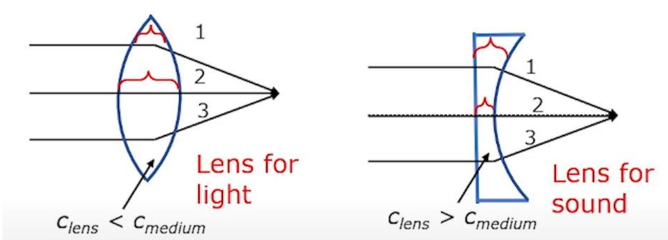
So this is a high quality factor, which is preferred for continuous wave applications. But for pulsed wave applications, we want high damping and thus the quality factor will decrease because the resonant frequency may remain the same in this case, but the Δf or the width of this resonance will drop.

In fact, when you have damping, the resonant frequency also shifts down a little bit, but that can be compensated for by design. Now let us discuss the lens and its use in ultrasound imaging. So, you are familiar with lenses in the context of optics. Lenses are used for focusing. For example, the glasses I am wearing right now, they have lenses because I have myopia.

So, lenses are typically used also in ultrasound as a last layer in a transducer to focus ultrasound to get sharp images. A focused ultrasound has better lateral resolution and higher focal intensity. What do I mean by that? Well, if you focus sharply, then the image formed will be sharp. Further, the focal intensity is high. That means that the energy concentration at the focal point is high.

So, you will get a good signal to noise ratio. So typically, in optics we have convex lenses for focusing. However, in ultrasound focusing lenses are generally concave in shape or the transducer crystal itself can have a concave geometry. And why that is the case? Well, typically in optics, the lens is made of glass or some other material in which the speed of light is slower. The speed of light is faster in the air and slower in the lens.

So, say we have these three rays, 1, 2 and 3, and we want all of these rays to focus at a point, to sum up coherently at a point, to arrive together at a point in phase. So how is that possible? Because ray 1 has to take a longer path. Similarly, ray 3 has to take a longer path than ray 2. That is only possible if ray 1 and ray 3 are allowed to speed up. How is that going to happen? Well, ray 2 travels through the thicker part of this lens and this is an optical glass lens we are talking about.



So, what happens to ray 2 is it slows down as it passes through the thickest part of this lens. However, ray 1 and ray 3 pass through the thinner part of the lens, so their time of flight in the slower speed of light medium is less. So, they don't have to slow down too much, they quickly get out of the slow medium whereas ray 2 travels for a longer time in the thicker part of the lens and thus this retardation that we provide to ray 2 if we compute it properly it will allow all of them to arrive together at the focal point in phase.

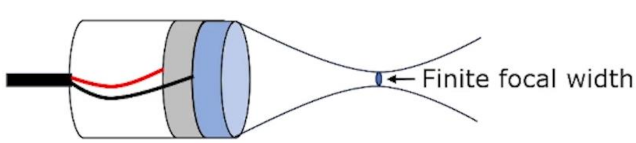
Case in point, ultrasound. In ultrasound, the lenses that are used are typically made of aluminum or some composite materials.

And the speed of sound, so now we talk about speed of sound rather than the speed of light. The speed of sound in air or in tissue is much less than the speed of sound in aluminum. So now if you pass sound or ultrasound through this lens material, it will speed up, the ray will speed up. So, let's consider ray 1. So again ray 1 has to travel a longer path compared to ray 2 for them to arrive in phase and combine coherently at the focal point.

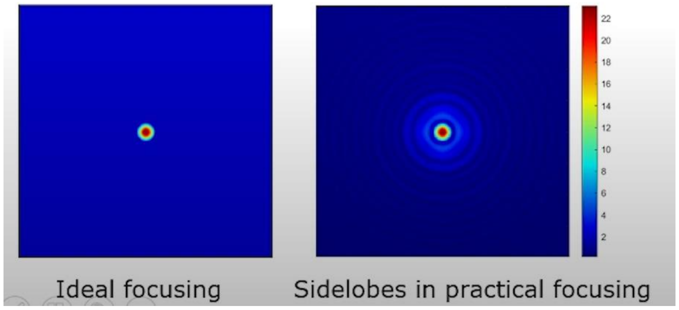
Now how is that possible? If I make ray 1 pass through a medium which has faster speed of ultrasound, then it will get that advantage which will help it cover that additional distance and arrive at the focal point at the same time. So I hope this point is clear why a concave lens will actually serve to focus ultrasound. Now instead of having a concave lens, the crystal itself can be deformed to have a concave geometry and that will also have a focusing effect. The other aspect is how much focusing we can achieve. So theoretically, if we talk about ray optics, we ignore the wave properties of light, then we can focus to a point.

But we have studied diffraction because of which we cannot really focus to a point. Similar concepts apply here itself. The width of the beam at the focus is actually proportional to the size of the aperture and of course it's also related to the lens. Now, the aperture diameter of the transducer, the bigger it is, the sharper the focus will be.

For example, I'll give you an analogy in optics. You may all have seen cameras. So, you may take cameras in your cell phone, or you may consider the professional cameras used by photographers which have larger lenses. So, the larger lenses produce sharper focus. Similarly, here, a larger aperture will lead to sharper focusing. Now focusing to a single point is possible only if the transducer has an infinite aperture but the transducer has a finite aperture so you can think of this as a truncation in the aperture and because of which you will not get an ideal focusing to a very small point, but you will get side lobes in focusing.



So, if I take a cross section of this, you will see a main lobe and you will also see side lobes. So, this kind of ringing behavior you will see and this is what is called the point spread function. We will discuss more about it in later classes. So here you can see the focusing is best at the focus point. If you move away from the focus point, then the focusing is not so sharp and thus the quality of the image will degrade.



Now what if the transducer is unfocused essentially it does not have a lens right then we can call a transducer unfocused so neither does it have a lens nor does it have a concave geometry to provide focusing however this is an extended aperture so it will have a natural focus although it will not be a very sharp focus but this is not a point source so this is a directed source this has a finite aperture and because of reasons explained by diffraction which we can discuss in further classes you will see that it does have a focal point. So, how do we get to the focal point? Well, there is a mathematics behind it where you can calculate the field of an ultrasound transducer by considering it as a combination of very large number of points. And so, when you do that, you will find that for this kind of a transducer consider a disc transducer with a diameter of d. The focal length, which is also called the natural focus, is calculated by aperture diameter square by 4 lambda. Now this length is called the natural focus, and this is the focal width.

$$Focal\ length = \frac{Aperture\ diameter\ (D)^2}{4\lambda}$$

(only for unfocused transducer)

Now as you can see in this case the focal width is pretty wide, the focus is not very sharp and that is because there is no lens or there is no concave geometry. So, it's a bit confusing because we are saying this is an unfocused transducer, but then we talk about its natural focus. I guess it is what it is. But let's remember that this formula that I have discussed, focal length is aperture diameter square by 4 lambda.

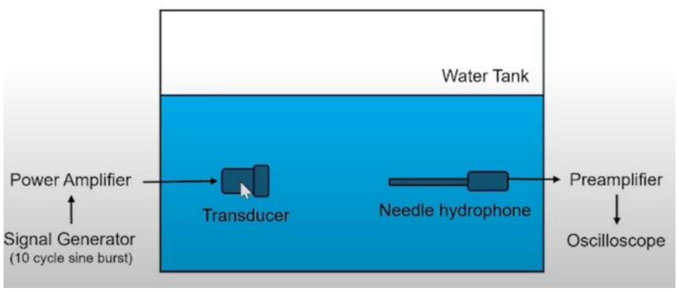
This is applicable only for unfocused transducers. If we apply a lens or we apply a concave geometry, then these properties will change. So now the focal width can also be calculated as lambda times focal length by aperture diameter. This is also an approximation, but this works pretty well. An F number, which actually tells you how sharp the focus is, is calculated as the focal length by aperture diameter. So these metrics you will see often when transducers are sold or transducers are designed to specification, these numbers are given so that the designers can act accordingly.

$$Focal\ width = \frac{\lambda\ x\ Focal\ length\ (d)}{Aperture\ diameter\ (D)}$$

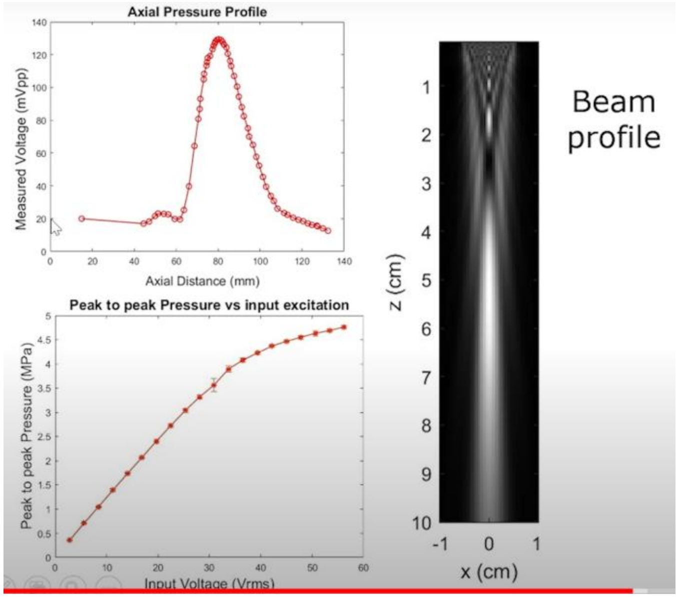
$$f\ number = \frac{Focal\ length\ (d)}{Aperture\ diameter\ (D)}$$

Now a smaller F number indicates a tightly focused transducer. So, for imaging, we have reasonably well focused transducers. For therapeutic ultrasound applications, such as using ultrasound for ablation, destroying cancerous tissue, etc., very sharply focused transducers are used. We won't be discussing so much in this class, but you can look them up if you are interested.

So, before we use the transducers, we ought to characterize the transducers. Fabricated transducers are characterized to ensure reliable parameter values and for this the typical measurements that are made are the axial and lateral pressure profiles and the intensity profiles. And for this, as Professor Karla Mercado-Shekhar would have taught you in an earlier lecture, pre-calibrated transducer with a small aperture, which is a hydrophone, it is like a microphone but typically used underwater in immersion experiments, is used. And a hydrophone is placed in front of the transducer and moved around spatially to map the entire beam. And particularly, it's common to measure the signal at the focus, but you can move the transducer in the 3D grid and figure out the pressure or the intensity profiles at all these points in the grid.

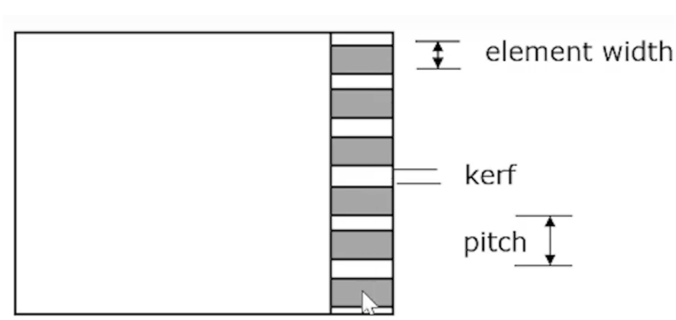


So, this is an image of the axial pressure profile.



So, on the left you can assume that the transducer is located at the left at 0, x equal to 0 and as you move further along the axis, you see that the pressure starts increasing as evidenced by the higher voltage and this is the focal point. So, if you reach the focal point of about 80 millimeters, you get the strongest response because that is the location of the focus. Similarly, you can change the pressure at the focus by changing the voltage. So, if you drive the transducer more strongly, let's say the input voltage changes from let's say 10-volt RMS to 20-volt RMS, the peak pressure as you can see here almost doubles to about 2 Mega Pascals. This is a peak-to-peak pressure. Now as you start going higher and higher to 40, 50, 60-volt RMS, you start seeing some saturation and some non-linearity sets in, in the response of the transducer. And here is a beam profile of the transducer showing the main beam and some side lobes and some patterns in the near field which we will discuss later. Now let us discuss transducer arrays. So far, we talked about single element transducers but transducers can also be arranged in an array geometry or rather crystals can be arranged in an array geometry to get a composite transducer which is used a lot in medical imaging these days.

So, consider this schematic of a transducer.



The gray rectangular boxes represent individual crystals which are arranged in a row and the white rectangular spacing between them represents the insulation layer. So, the width of this gray rectangle is the element width, and the width of this white rectangle is called the curve and the distance between the center of one crystal to the other crystal is called the pitch. So here is a schematic of a linear array transducer in which these crystals are arranged in a row, a curvilinear transducer in which the crystals are arranged in a curve and a phased array transducer in which these crystals are arranged in a special way in a row which allows this transducer to steer the beam in different directions. So for linear arrays, the pitch has to be less than the wavelength and for phased array, the pitch has to be less than wavelength by 2 to avoid what are called grating lobes. These grating lobes are a specialized type of side lobes which can appear in the image and degrade the quality of the images.

For linear arrays

Pitch $< \lambda$

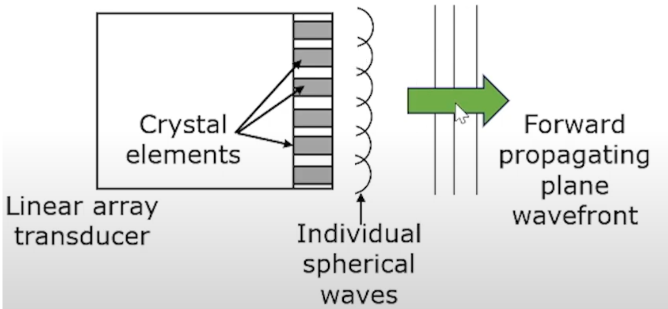
For phased arrays

Pitch $< \lambda/2$, for avoiding grating lobes

So, to ensure that these grating lobes don't affect your image, you have to follow the lambda criteria for linear arrays and lambda by 2 criteria for phased arrays. So for phased array we can do what is called beam steering which means we can move the beam in different directions. Linear arrays are typically used for superficial applications for imaging organs which are not located at high depth to give you an example carotid artery in the neck which supplies blood to the brain. Curvilinear array is usually lower frequency arrays which are used to image wide fields such as abdominal organs which are located deeper.

And phased arrays are used when you have a very small acoustic window. That means that due to the presence of ribs for example, bone for example, you don't have a very wide region through which you can image. So, you have a narrow region but you would like to steer your beam in different directions and be able to cover a large volume of imaging and thus that is where phased arrays are used.

So now these transducer arrays can typically contain 32, 64, 128 or 256 crystals. And these are typically available in powers of two because it helps with some algorithmic and hardware implementation related costs and efficiencies. Now, individual elements, because these are smaller than lambda or lambda by two, depending on whether this is a linear array or a phased array, these individual elements will not create directed fields, they will create spherical waves.



But when all these elements emanate spherical waves which interfere with each other, what you get is a forward propagating wave front and if all the elements are fired together it will approximate a plane wave front. Plane wave front means that all the points in which the waves are in phase lie in a plane. So, plane wave imaging has this advantage because all the transducers are fired together and not in a sequence. It has very high temporal resolution, that means it can acquire images faster.

But the lateral resolution is poor because of the large beam width. Plane wave essentially means no focusing on the lateral direction and because of which the beam width is wide, which leads to poor resolution in images. So just recall Huygens principle which you may have studied in optics or in your high school. This is just to give an intuitive understanding of how an array transducer generates its fields. So here is a plane wave which is coming and interacting with a slit.

This slit you can think of as a single crystal. This single small crystal smaller diameter than wavelength will produce this kind of a spherical wave front. Now if we have an extended slit which you can think of as being composed of a large number of transducers denoted by these yellow dots here then the signal will be caused by the interference of these waves and we get a pattern here which can be approximated as a plane wave. So now just as I mentioned, to get the frame rate, which means the time taken to acquire a single frame, you need to calculate the pulse repetition period.

$$PRF = \frac{1}{PRP}$$

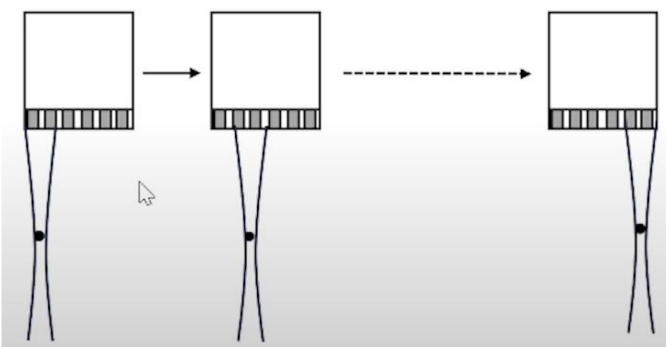
$$Time\ to\ obtain\ one\ image = number\ of\ \ A\ lines\ x\ PRF$$

$$Frame\ rate = \frac{1}{Time\ to\ obtain\ one\ image}$$

In ultrasound, we are sending pulse one after the other and the time duration between two consecutive pulses that are applied to the transducer is called the pulse repetition period. So, this pulse repetition period depends on the maximum depth of imaging required because once you fire ultrasound, you would have to wait for the adequate period of time for the sound waves to get back scattered from the targets that you are interested in and come back to you. And once you have waited enough for the adequate depths so that the signal comes back to you and some of the residual signal dies out, then you can send the next pulse. So now, like I said, the pulse repetition frequency can be calculated as the inverse of the pulse repetition period. The time taken to acquire one image depends on the number of A lines times the pulse repetition period. So typically, if you are firing, let's say 128 times to get one image, then your frame rate will be calculated as one by the time taken to acquire one image, which will be 128 times the time required to acquire one single A-line.

So, a higher frame rate typically improves temporal resolution, but there are some disadvantages in terms of artifacts, which you will learn from in a subsequent lecture by Professor Karla Mercado Shekhar. In linear array imaging, which is one of the most commonly used, a few elements are grouped together and fired to get an A line. So here, consider there's 128 element linear array, and we are grouping 32 elements and firing them at a time. So first, let's say element one to element 32, all these fired together, then we shift by one element.

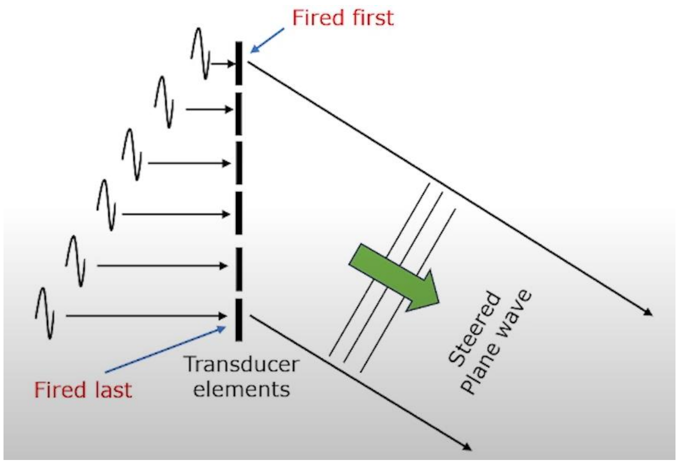
So, then we take the second element all the way to the 33rd element and we fire. Then we again shift by one element. We fire from the third to 34th element and so on. So, by doing this, we are activating the sub aperture at a given point in time.



We can also apply delays here to focus the beam. More on this later. And then once we get the signal back, we combine all these signals. We do a process called beam forming, which will have its own separate lecture. And then we are able to make an image. So, the number of A lines to make an image is equal to number of elements minus number of grouped elements plus 1. So, for example, if in this case we have 120 elements and we are firing 32 elements at a time, then the number of A lines required to make this one linear array image will be 128 minus 32 plus 1.

$$\begin{aligned}&Number\ of\ A-lines\ to\ make\ an\ image\\&\qquad = number\ of\ elements - number\ of\ grouped\ elements + 1\end{aligned}$$

Next is phased arrays. So, in phased array transducers, the phase of the individual elements can be electronically controlled to get desired directionality in the ultrasound beam. So, what's going on here, you have signals emanating from these individual transducers shown as black lines. They would typically fire in the forward direction, but here you are steering the wave to the side by applying this delay profile. What this means is, this wave electrically is excited first.

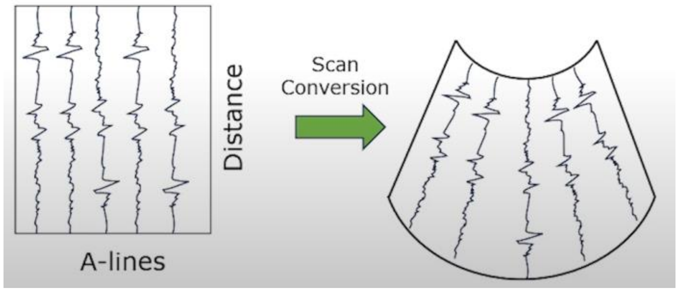


After some delay, the second wave comes and excites this transducer. And then the third wave comes and excites this transducer element. The fourth wave comes and excites this transducer element. So, there is a delay profile. This delay profile actually creates a wave front which is steered in a particular direction. So why would a delay profile cause this steering? Imagine you are doing march past Independence Day with your friends.

Now your group has to turn right, for example. How would you do that? The person who's marching rightmost would have to slow down and the person who's in the same row but marching leftmost would have to speed up. So in this case, if we have to steer the beam, just the manner in which we turn in march past, we cannot change the speed of sound because speed of sound in the media is fixed. But we can accelerate or retard certain beams by providing these delays. So, when we provide larger delays, we fire this transducer element last and this transducer element first, it has the effect of having steering of this beam.

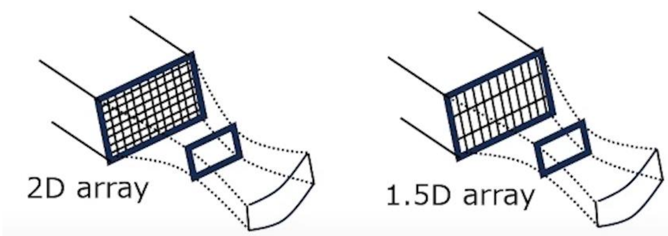
And by changing these delays, we can steer in any particular direction. Further, steering can be combined with focusing by changing the delay profile. So we'll discuss more on this in the topic of beamforming in a subsequent lecture.

Let us discuss sector or curvilinear arrays next. As I mentioned, in sector arrays, the elements are arranged on a curve rather than in a row and this enables a wider area of imaging. This is typically used for low frequency imaging and deeper abdominal organs are visualized using this technique.



The beam pattern is in the shape of a sector and not rectangular. However, the data is typically acquired by the sector array. When it is stored, it is stored in the form of 2D matrix because these are the individual A lines, and they will have their own values when it is converted to a digital matrix. And then to get into the right geometry, you have to convert this into the right coordinate systems and then you obtain the sector in the right format.

Let us now discuss the two-dimensional and 1.5-dimensional arrays which are also available. In one-dimensional arrays which are the most common, the elements or the transducer crystals, they are arranged in a row. But in two-dimensional arrays, they are arranged in a matrix and the number of rows and columns of the matrix are comparable. They need not be the same, but they are comparable.



In 1.5 dimensional arrays, the number of rows and columns are disparate. For example, you can have 8 columns but 128 rows. And so, because one dimension is much larger than the other dimension, that's why they are also called 1.5 dimensional arrays. Now these matrix arrays or the 2D arrays can scan volumes while imaging and therefore they can perform 3D imaging.

And now you can apply delays along either direction or thus you can focus on both lateral axes. Now 1.5 dimensional arrays can also focus in the lateral and elevational direction but not to the same extent as two-dimensional arrays. So, a 1.5-dimensional array can be considered as a compromise between a standard one-dimensional linear array and a two dimensional matrix array. The trade-off here is that if you have a two-dimensional array, it requires a large number of elements and thus a large number of electronic circuits, interconnects, etc.

But 1.5 dimensional arrays have few rows of elements in one dimension. Therefore, you save up on the complexity in terms of the electronics. So let us summarize what we have learned in today's lecture. We discussed ultrasound transducers, individual crystals, and the construction of an ultrasound transducer. We discussed the resonance frequencies of the transducers, quality factor, etcetera. We discussed focusing lens, how we can focus in ultrasound either by using a concave lens or by using a concave geometry of the transducer elements themselves.

We also discussed impedance matching, why impedance matching is important for better coupling so that there is better energy transfer between the crystal and the tissue. We also discussed transducer arrays, both linear and phased arrays, including curvilinear arrays. So, with this, I'll conclude this lecture. I hope this sets your foundation for advanced topics in image reconstruction. Thank you.