Biomedical Ultrasound: Fundamentals of Imaging and Micromachined Transducers

Prof. Karla P. Mercado-Shekhar, Prof. Himanshu Shekhar, Prof. Hardik Jeetendra Pandya

IIT Gandhinagar, IISc Bangalore

Lecture: 11

Recap of Week 2

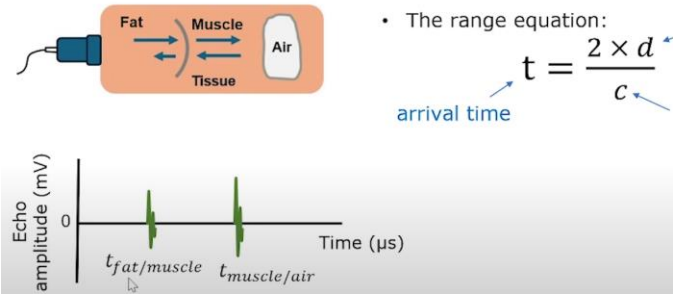
Hello and welcome to today's lecture. In today's lecture we will take a recap of what we learned in the last week. So, we have discussed ultrasonography or ultrasound imaging as we would say it and its ability to visualize internal body structures by different imaging modes. There is no one size fits all mode. There are modes which are applicable in certain clinical contexts based on the diagnostic information that you would like to receive. So first, the simplest one is A-mode in which you send a single RF line and when you get back the echoes, you take the envelope and that actually localizes the presence of the strong scatterers.

And this A-mode imaging is now very uncommon in diagnostic ultrasound imaging, but it is still very extensively used in non-destructive testing for looking at structures, beams, aircraft wings, railway tracks, etc. where mechanically scanned A-mode is very common. Then comes the brightness mode. Typically, if you have seen any ultrasound image, you may have seen any fetal image from one of your relatives, for example, or any other abdominal images, most likely the image you have seen is a B-mode image, which is called the brightness mode image, which can be created either using mechanically scanned A-mode or by using array transducers.

Next is the M-mode or the motion mode. It can be used to look at the motion of certain structures such as the heart. And then comes Doppler modes which are useful for visualizing and quantifying blood flow. And there are a variety of modes including continuous wave, pulsed wave, color and power Doppler. So, we will just revise them quickly.

So, first of all in A-mode, it is based on the pulse echo principle. It is essentially an echo location principle similar to that used by dolphins and bats. So, the idea is that you have the instantaneous echo signal amplitude versus time. This is what you will have on the oscilloscope should you do the measurement. and here is an example where you fire a transducer it goes through layers of fat and muscle and air and there are these different interfaces so the strongest reflection would actually come from these interfaces and

probably the maximum impedance mismatch you will see here will be between the air and muscle impedance and that echo will be the strongest.

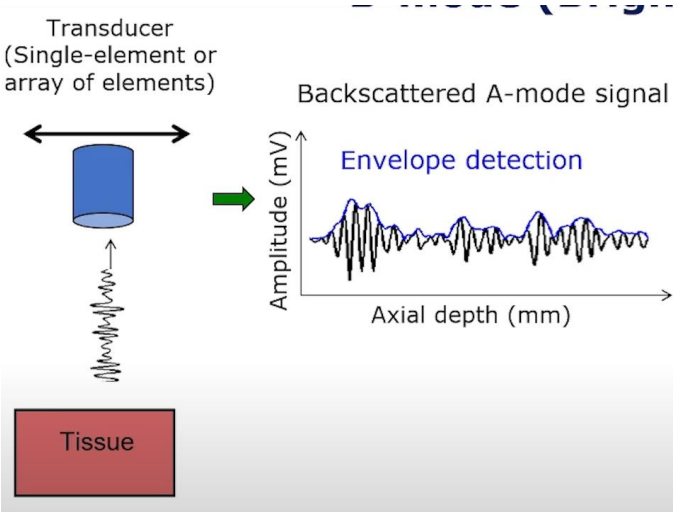


Now after how much time do you expect to get the echoes after firing the transducer? So that will be calculated using the range equation. The time is given by 2 times distance by c. This factor of 2 comes because this is pulse echo imaging. The same transducer is serving both to transmit the pulse as well as to receive the pulse. Therefore, the distance travelled is actually 2-fold.

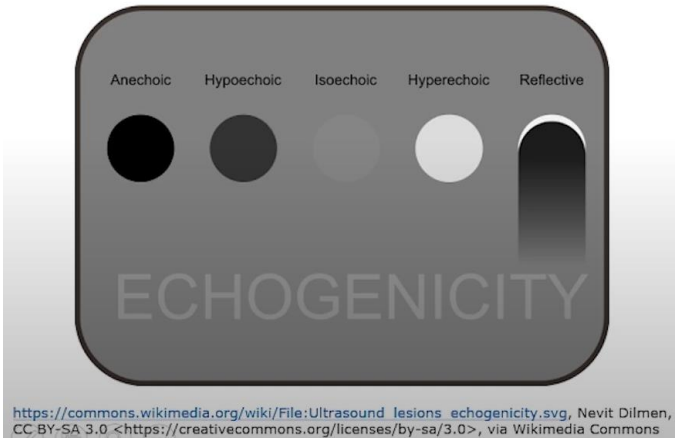
$$t = \frac{2\ x\ d}{c}$$

That's why the term 2d by c. Now once you get the echo amplitudes, let's say this is the first interface, the interface of fat and muscle and after some time you get that echo, and the second echo is from muscle and air and as you can see this echo is stronger in the schematic which makes sense because there will be a stronger impedance mismatch between muscle and air. Now this A-mode assumes that there is a single ray that is transmitted and is received back, and it only displays echo data from a single beam line but we do have to realize that there is a finite thickness to the beam so it's not just a ray but a finite thickness from which the signal is being received. So then if we talk about B-mode or the brightness mode, this is the most common mode and it can be used for visualizing structures such as the chambers of the heart, opening and closing valves, etc., as you can see in this image.

To create a B-mode image we need a grid or a matrix and then you can simply scan an A-mode transducer in one axis and then you already have the RF data in the other axis. So, with this you will have a grid of the information and if you look at the envelope of the scattered A-mode RF signal and you stack them up together you will actually get a B-mode image. You can also do this by using an array transducer where you don't mechanically scan the beam, but you electronically scan the beam. Now let's discuss B-mode echogenicity. The word echogenicity literally means the genesis of the echo or the origin of the echo.



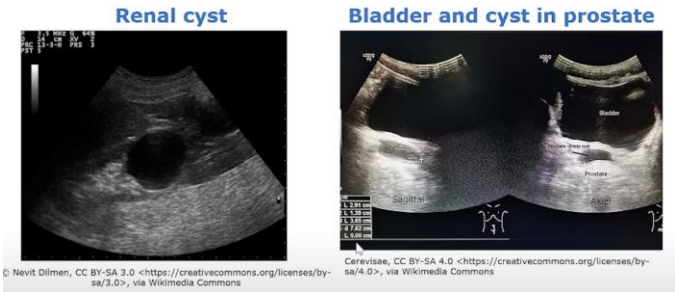
So, if we have structures that are anechoic, these would be the structures that appear black in the image, meaning that there's no echoes originating from these structures. And what kind of structures would have this kind of a response? Well, if you have structures which contain fluid where there are no cells, for example, that will lead to no internal echoes and would appear anechoic. Now there are structures which have weaker echoes than the background and these are called hypoechoic structures. So, they appear darker than the structures which are around them and here is an example in the schematic. There are isoechoic structures which exhibit the same brightness as their surrounding structures.



So, for example imagine having an isoechoic tumor which means that the echo signal coming out from the tumor is almost similar to the surrounding background. So, what would happen? This tumor would be very difficult to spot. So, let's assume this is a tumor against this background. This tumor in the presence of electronic noise would be very difficult to spot and would therefore be called an isoechoic tumor. And hyperechoic structures are ones that give out more signal than the background and therefore they are more echogenic or brighter than the surrounding structures which you can see here.

And it turns out sometimes you get a very strong echo at the interface when you have reflective structures, and you may get a shadowing behind them because most of the signals have been reflected back to the transducer. Now let us look at some anechoic structures in the body. So here is a cyst in the kidney or a renal cyst which is anechoic. There is not much signal emanating from the cyst because it is fluid filled. Here is an image of the bladder which is fluid filled and therefore anechoic.

Here is another image of the bladder and a cyst present in the prostate gland. So this cyst is also fluid filled and therefore this is also hypoechoic relative to the background. And now I'll show you some hyperechoic structures. So here is an example of a stone in the gallbladder. And this stone is a strong scatterer.



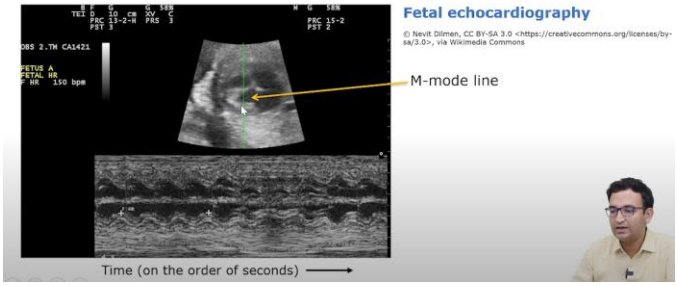
Therefore, there's shadowing behind it because it reflects most of the signal right there. Then we discussed 1D, 2D, 3D and 4D ultrasound. A-mode being the simplest is 1D ultrasound. It's not really an image, although sometimes you call it an image. It's just a one-dimensional matrix.

And then we have the two-dimensional ultrasound or B-mode ultrasound, which is the most common ultrasound that you have seen. And then if you have B-mode with a matrix array transducer, more on this later, but this is a two-dimensional array transducer as opposed to the standard one-dimensional linear array transducers. Or if you take a one-dimensional linear array transducer and mechanically scan it, you can get a 3D image, which is a three-dimensional representation. And if you are able to reconstruct 3D images in real time, essentially if you are able to make live 3D images, then these can be called four-dimensional images. So here are some 3D images of the fetus.

And here is a 2D image of the valve in the heart chambers. And here is actually a 4D image. This is actually a video essentially, not an image. But this is a 4D image where it is three-dimensional and also in real time. So, it can be called four-dimensional.

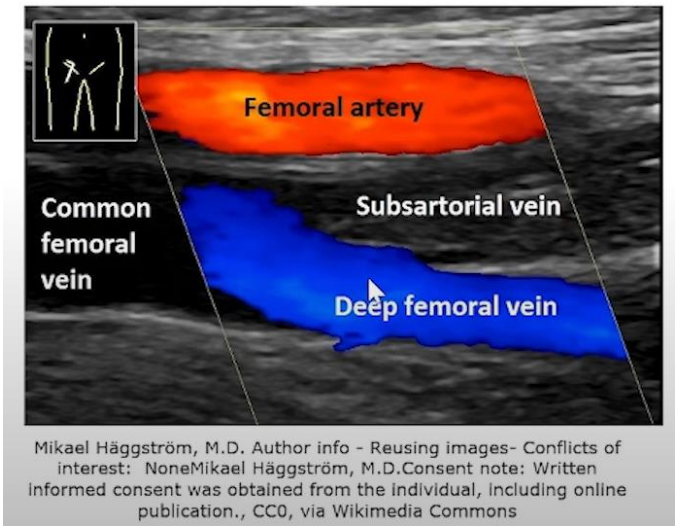
Now, when it comes to looking at the motion of certain organs, especially the heart, just looking at the motion of the heart wall gives a lot of diagnostic information. So, the idea is to look at the heart wall over time. So, what would be done is you have this is a traditional B-mode image and the chambers of the heart are visible here. You would take

a single line shown in green here and you would like to see the motion of structures along this line. So only one anatomical dimension is represented.



So, if I was to just draw a line here that would tell you about the signal along this line. But then I would like to repeat it over time. So, the same location you are taking that one line again and again and displaying it as a two-dimensional image. So, it's dynamic information that you are getting and these kinds of oscillation-like structures you see in these M-mode images, they represent the dynamic information because the heart is beating. So, this M-mode is also quite useful for assessing anomalies in the heart.

Then comes the color Doppler. So Doppler imaging allows us to get two-dimensional imaging of blood flow. We learned that the convention is such that if the flow is towards the direction of the transducer being used for imaging, it's called red shift and here actually the structure is depicted in red and if it is away from the transducer, then it is depicted in blue.

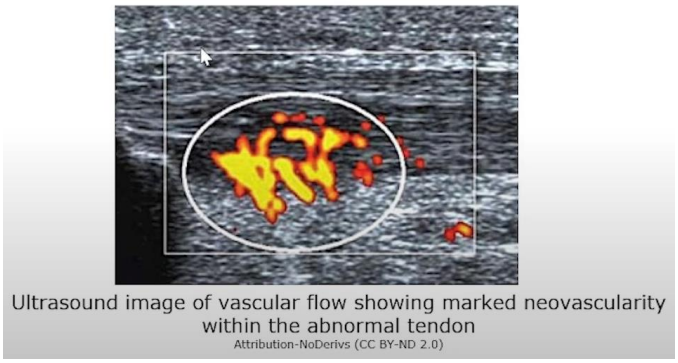


Now this is very interesting because you may have read about the Doppler effect in physics and in physics you have red shift when the structure is going away and blue shift when the structure is coming towards us. For example, we know that the universe is expanding because you have red shifts in the distant stars. So interestingly, the convention is opposite for ultrasound and for astrophysics, for example, but nonetheless

for historical reasons, this is the convention which is in place, and which is being followed.

So, this gives you not only anatomical information with traditional B-mode but also some functional information. It gives you direction and velocity of flow which can be quite useful to assess diagnosis. In some cases, you have regurgitation, or you have some kind of pathological flow, abnormal flow and that can be easily seen using this kind of Doppler images.

Next comes power Doppler. So, power Doppler is more sensitive to flow than the traditional color Doppler and is also known as the angio mode. And what's interesting about this power Doppler is that it doesn't give you any directionality of flow, but it only tells you the spatial regions where there is blood flow. So, the color coded image of blood flow is shown based on the intensity rather than the direction and it is useful tool for examining low velocity blood flow because this is more sensitive because of the averaging effect. Therefore, it can detect blood flow from relatively lower velocity blood vessels as well.

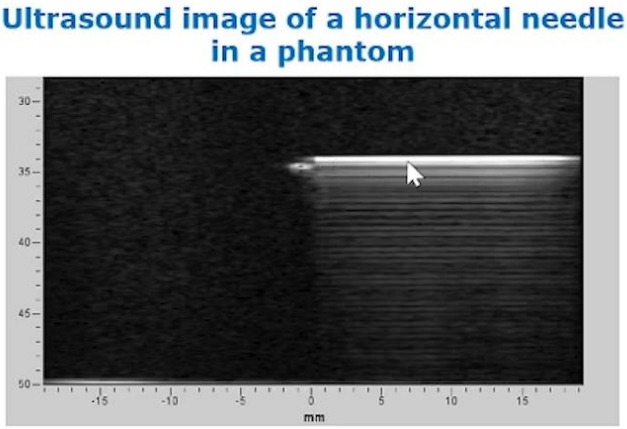

Ultrasound image of vascular flow showing marked neovascularity within the abnormal tendon

So here is an ultrasound imaging of a vascular flow. This is in a tendon which is abnormal and because it is abnormal and it is probably healing, you have angiogenesis inflammation and that can be seen in this angio mode power doppler.

Now you'll recall us also discussing imaging artifacts. Artifacts, like we discussed, are unwanted features in images that don't accurately represent the target being imaged. Artifacts can cause false portrayal of anatomy or function.

You can have structures which don't appear in images. You can have mislocated or distorted objects in the image. There can be appearance of fake objects, which we also call ghosts sometimes in colloquial language in ultrasound images. And there can also be incorrect flow speeds when doing Doppler imaging. We discussed how artifacts are different from noise.

Broadly speaking, we can call artifacts a subset of noise. However, while noise is typically random, artifacts are deterministic. Here is an example of the comet tail artifact caused by multiple reflection or reverberation.



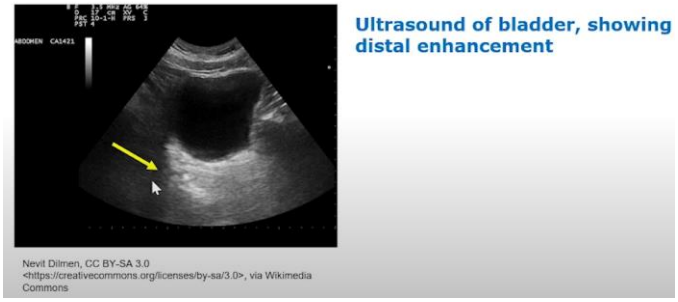
So, this is a needle inserted in a phantom and because this needle is such a strong scatterer being metal, there is a large acoustic impedance mismatch, and the needle signal bounces off the transducer phase, and you get these streaks or this reverb like artefact. So, this is very interesting because now the beam path actually includes two strong reflectors, one being the transducer surface itself and other being the needle and this fading artifact which looks like a comet tail is appearing here.

Now it turns out that because of this sometimes you can confuse this as some other target which is undesirable.

Next, let's discuss acoustic shadowing. So, when the ultrasound beam encounters a structure with high attenuation, rather higher attenuation than assumed in the image reconstruction, the distal structures appear darker than they should seem. So, what's going on here? Here you have an image where there is a stone in the gallbladder. So, this anechoic structure filled with fluid is the gallbladder and you have a stone that has formed here.

So now this stone is actually reflecting back most of the sound. So, in the distal region you get this dark hypoechoic region which is called shadowing. Now structures such as stones, bones, metal implants, calcification etcetera in the tissue can cause shadowing. A similar effect is acoustic enhancement, where if the ultrasound beam interacts with a structure with lower attenuation than assumed in image reconstruction, the distal structures appear brighter.

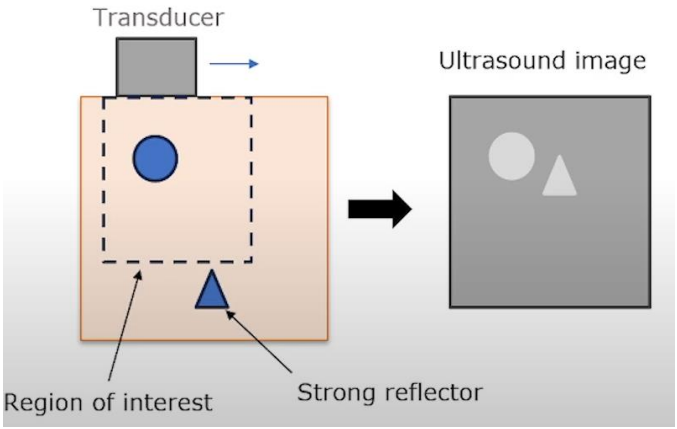
So here, you have some tissue which is a cyst.



So, there's not much attenuation. But your ultrasound reconstruction approach actually thinks that this tissue is attenuating the signal. And therefore, it enhances the signal. And this enhanced signal appears as a distal increase in the signal. It creates a hyperechoic region, which is called the distal enhancement.

And this is certainly an artifact. It's not that this region is a strong scatterer relative to the adjacent region. So, fluid-filled structures such as bladder or cysts can cause acoustic enhancement in distal structures.

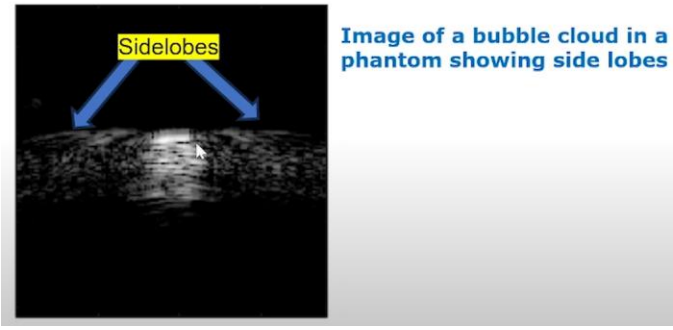
We discuss the range or the ambiguity artifact. So typically, we decide the depth at which we want to image. And we choose our pulse repetition frequency in such a way that we are able to receive signals from a certain depth.



And then we allow some gap to enable the signals to die down. And then we receive the next echo. Now it turns out if the pulse repetition frequency is not chosen appropriately and if it is too high, the echoes from deeper region do not fully die out before the next pulse is transmitted. So, to give you an example, consider this transducer in grey here and this structure in blue is our target and these dotted lines depict the region of interest. So, my image is only going to have this region of interest and nothing more.

But there is a strong reflector which is located beyond my region of interest and as I am scanning this transducer it turns out the echo from this strong reflector reaches me and I assume th is to be part of my region of interest and I may get an artifact like this where the strong reflector is actually present in the ultrasound image.

And then there are side lobe artifacts.



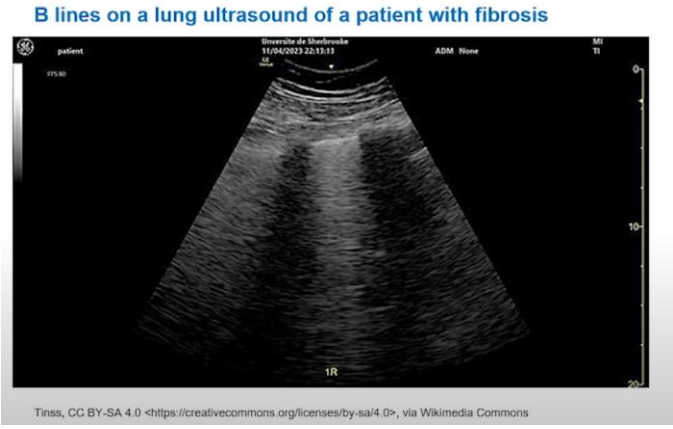
So, we discussed that ultrasound beam has a main lobe and it has side lobes. And for ideal imaging, we would like to suppress the side lobes. There are some specialized side lobes as well, such as grating lobes, which are present sometimes in arrays and they appear in addition to the primary sound beam. So, what happens if there are side lobes in the beam? If there is a strong scatterer, it will interact with the side lobes as well and then you get these kinds of artefacts. So here there is a bubble cloud here which is a strong scatterer and the streaks which you see on the side, these are actually caused by the side lobes. So, the side lobes are going to show some structures which are not present in the actual image.

We also discussed edge refraction artifacts.

So, when the ultrasound passes through this edge, because of the acoustic impedance mismatch, unless the incidence is entirely normal, there will be refraction. And this refraction prevents the signal from reaching back the transducer, and therefore these dark streaks are formed at the edges. So, the clinicians are actually trained to interpret these artifacts and not to get tricked. So, like I was saying, sonographers are trained on how artifacts arise and what they represent and some artifacts can in fact be useful to the sonographer because they avoid misdiagnosis and improve the accuracy of their interpretation. For example, if you have shadowing in a certain tissue which is supposed to be soft tissue, one can assess that this is some kind of foreign body, or this can also be calcification.

So, this is an example in the liver where there is calcification, and it shows distal shadowing and therefore it can easily be identified.

Another artifact is that in the lung. So the lungs is filled with air and therefore it creates a lot of artifacts when imaging is performed. Now imagine if the lung has some kind of pathology such as pneumonia or COVID, then the pattern of these artifacts will change which will help assess pathologies such as fibrosis or fluid buildup in the lung.



So changing tracks a little bit, in one of the lectures we discussed ultrasound transducers.

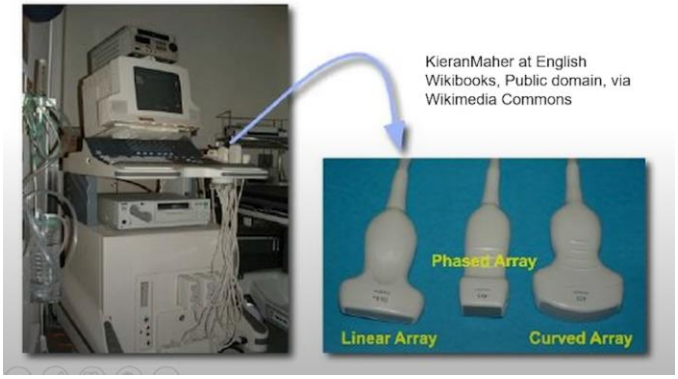
We discussed piezoelectric and the inverse piezoelectric effect. We discussed energy conversion from electrical energy to mechanical vibrations, as well as from mechanical vibrations to electrical energy. So first, when we want to probe a certain region in the

tissue, we provide electrical energy to the crystal. The beam is transmitted, the beam causes scattering and in the form of mechanical vibrations, I receive the signal from the transducer which converts it back to electrical energy. So, then we discussed that the crystal resonance frequency can be given by c by 2d where d is the thickness of the crystal and we discussed that we want the resonance to be in the thickness mode and not in the radial mode.

$$crystal\ resonance\ frequency = \frac{c}{2d}$$

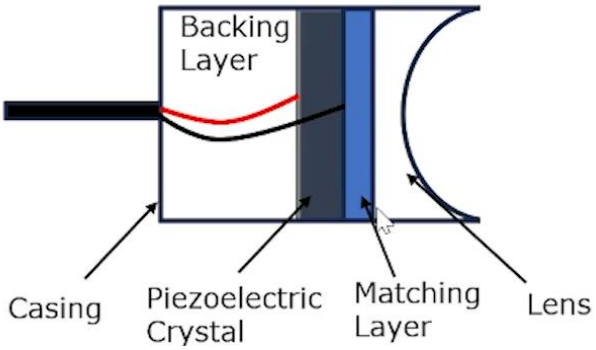
$$d = \frac{\lambda}{2}$$

Now if you want to have the transducer oscillate at a frequency f_o, employing the formula c equal to f_o times lambda and knowing the speed of sound c in the crystal material, I would set d equal to lambda by 2 in this formula, and I would be able to assess the thickness of the crystal which would give me a resonant frequency f_o. We discussed the different types of transducers, single element transducers, linear array transducers, curvilinear array transducers and phased array transducers. So, here's an example of a linear array, a phased array and a curvilinear array.



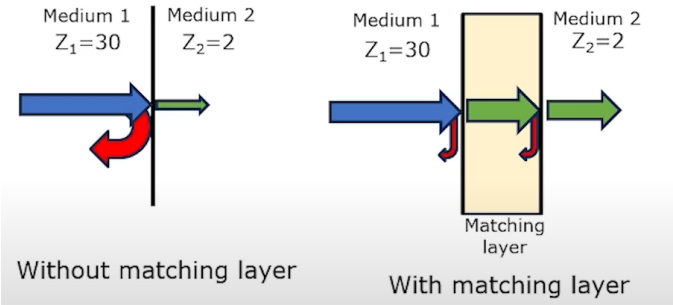
We also discussed that linear arrays are typically used for more superficial structures. Phased arrays are used for cardiac applications, applications where you have a narrow acoustic window, and curved arrays typically have slightly lower frequencies, and they are used for abdominal organs.

So here is a schematic of a single element transducer and the different parts of the single element transducers are indicated here.



You have the crystal itself which is shown here in grey. You have the matching layer which is there to match for impedance. Then you have the lens, and you have the backing layer which is typically filled with epoxy or some other dense material to damp the transducer for imaging transducers. Now the matching layer is needed because there is a striking impedance mismatch between the transducer and the tissue.

And we discussed that if you select a material with the acoustic impedance of square root of Z_1, Z_2 where Z_1 is the acoustic impedance of the input and output layers and Z_{ml} is the impedance of the middle layer or the matching layer then you are able to prevent a lot of signal reflection and loss of signal. So as shown in this schematic if you have Z_1 is 30 MegaRayl and Z_2 is 2 then majority of the signal is just reflected back at this interface but if you put this intermediate layer with a thickness of lambda by 4 or n lambda by 4, if lambda by 4 becomes too thin and brittle, you will choose n lambda by 4.

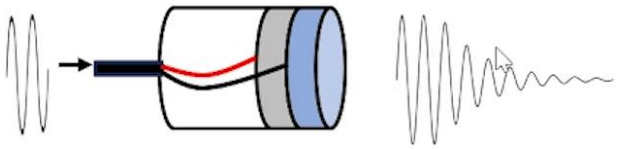


Then most of the signal is actually transmitted and there is very little loss because of reflection at the interfaces. So practically speaking, it's hard to find a single matching layer which has the properties such that its acoustic impedance is square root of Z_1 and Z_2.

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

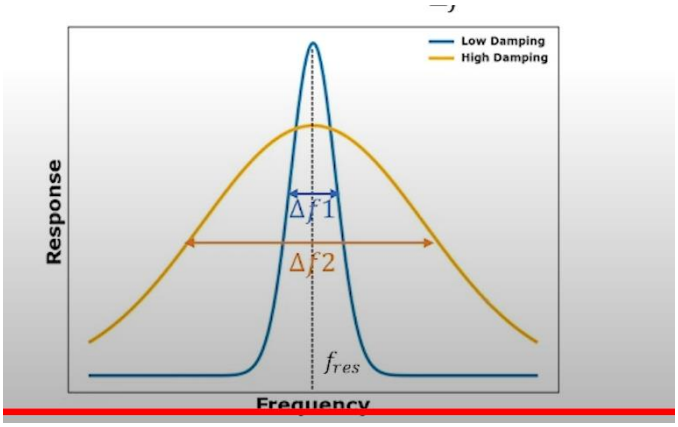
So typically, a cascaded matching layer is used or more than one layer, typically two layers of matching are used so that you can have good coupling of the signal coming from the transducer all the way into the tissue.

And as you may have heard or you may have seen if you ever had an ultrasound exam, gel is used. to eliminate any air between the transducer and the tissue in contact and that gel prevents acoustic mismatch because of presence of any air. We discussed the backing layer and its role for transducers driven by continuous sinusoids as well as short bursts. So, we discussed that continuous sinusoids are used in therapy transducers and short bursts are used when imaging is being performed. At resonance the transducer exhibits ringing behavior and for example here if we take a transducer and excite it with a two-cycle sine burst, even after this sine burst has ceased to exist in time, you still have signal coming out from the transducer.



This is because of the inertia of the crystal. And this is not good for imaging because this ringing increases the spatial pulse length which actually adversely affects the axial resolution. So, we would like the transducer to be damped in such a way that almost immediately the transducer oscillation ceases the moment the excitation pulse has ceased. And this can be done by selecting a suitable backing layer with the appropriate amount of damping. So, like I said, ringing is acceptable if it is a continuous wave transducer because transducers which are not damped, which have a sharp resonance or a high quality factor given by the resonant frequency by the bandwidth at the full width half maximum, they will have a high Q factor, they will have high amount of ringing. But transducers that are being used for imaging where axial resolution is important, where we would like to have low spatial pulse length, there we would like to have heavily damped transducers as shown in yellow here and in such transducers, we would like to have epoxy or other dense materials as their backing.

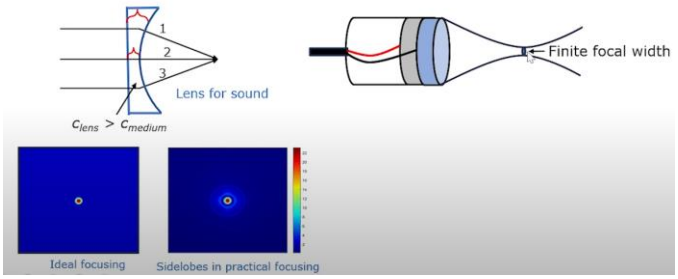
$$Quality = \frac{f_{res}}{\Delta f}$$



For therapy transducers where we want a high Q factor, we would like to choose a backing material of air. So, like I said for therapy applications ringing is adequate but ringing is not preferred for imaging applications. You would like to have heavy backing that will damp this ringing and materials such as epoxy and rubber composites are used for backing in imaging transducers.

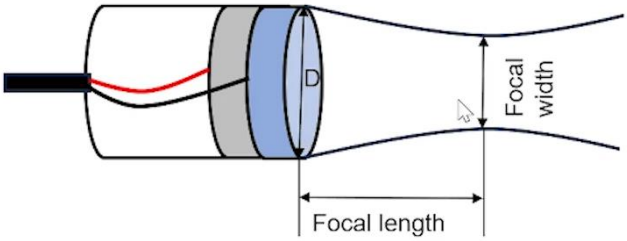
We also discussed lenses for the purpose of focusing in medical ultrasound. Lenses are used as a last layer in the transducer to focus ultrasound and typically the focus transducers have better lateral resolution and high focal intensity and therefore lenses are useful.

Now, ultrasound focusing can be produced with lenses where the material has high acoustic impedance when the lenses are generally concave in shape. We discussed how this is different from optical lenses where convex lenses lead to focusing and that is because of the difference in speeds of sound and light. In lenses that are acoustic in origin, the wave travels faster in the lens. However, for optical lenses light travels slower in the lens and therefore this difference is that acoustic lenses are concave but optical lenses for focusing are convex. So here you can see after focusing on the focal region you have some finite focal width because it's impossible to focus exactly on a point.



For doing that you would need a very large aperture. So in practical focusing you will not be able to focus very sharply and you will have the presence of some side lobes and this degrades image quality. We also discussed unfocused transducers. Well, as it turns out, if it is an extended source with an extended aperture, there will be some natural focusing.

But as you can see in this schematic, it is not very sharply focused. These transducers will have a larger focal width than focused transducers and also larger focal length.



And the natural focus, so again it's confusing because on one hand we are calling it unfocused transducer but typically we use the term natural focus. The natural focus is given by the aperture diameter square by 4 lambda. And this formula again remember is only valid for this type of transducer which does not have a lens or does not have a natural curvature which leads to focusing. So, there are other terminologies associated with transducers. One is the focal width which can be calculated as lambda times focal length by aperture diameter.

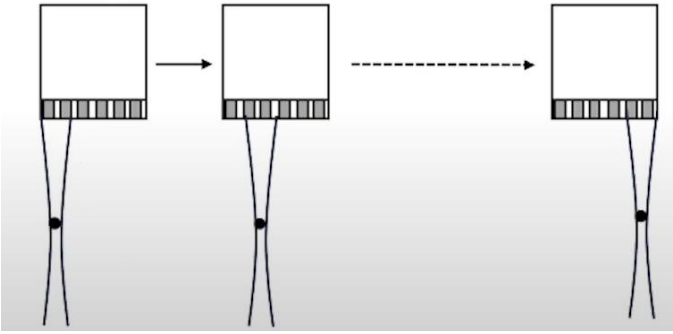
$$Focal\ length = \frac{Aperture\ dimeter\ (D)^2}{4\lambda}$$

(Only for unfocused transducers)

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

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

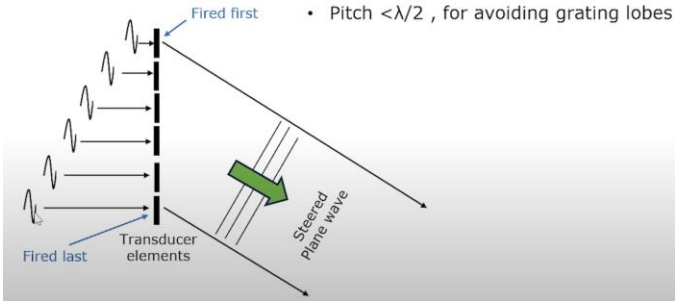
And then there is the F number which represents the focal length divided by aperture diameter. If the F number is low, that means the transducer is sharply focused. And if the F number is high, that means the transducer is moderately focused. We discussed transducer arrays, linear, curvilinear and phased arrays and we discussed how they have rectangular elements spaced by some insulating material and we discussed what is element width, curve and pitch of a transducer. Now these transducer elements can be electronically scanned to obtain B-mode images and for linear arrays the pitch should be less than lambda to avoid grating lobes.



Now remember that this is true for linear arrays. When it comes to linear array imaging, we group some elements, and we fire them together and because of which we can generate a focused wave and this focused wave is actually scanned across the elements. So now we can also provide certain delays here to control the point at which this focusing happens. We will discuss more on this in the chapter on beamforming. And therefore, the lateral resolution can be improved by focusing at a particular depth. Now take an example if a 128-element linear array can be fired by grouping 32 elements at a time.

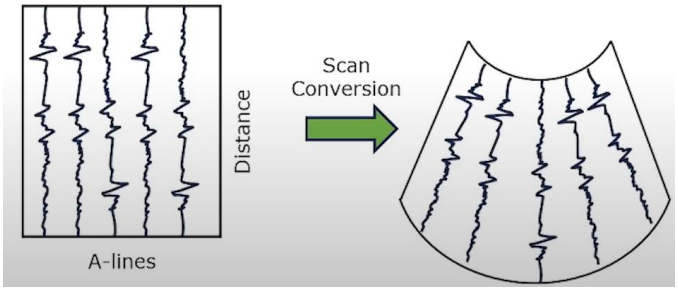
So, then we take 32 elements first. fire, then we shift by one, fire again. So essentially all the time I am firing 32 elements, but the locations from where I am firing keeps dynamically changing. And the number of A-lines required to make an image will be the number of elements in your transducer minus number of grouped elements plus one. So, if it is 128 elements in your array, then the number of firings will be 128 minus 32 plus one if we are firing 32 elements together. Then we discussed phased arrays which are very useful when you have a narrow acoustic window, and you want to take a look at a broader region. And in this you have the possibility of steering the beam by providing these kind of delay patterns.

For phased arrays, you have to have a pitch less than lambda by 2 to avoid these side lobes called grating lobes, which can actually degrade the performance of your imaging. And more on grating lobes in the chapter on beamforming.



Then we discuss sector or curvilinear arrays which are used for imaging abdominal organs where the field of view desired is very wide and the sector data is actually acquired in polar coordinates and then it is converted to Cartesian coordinates in a 2D

grid and therefore some interpolation and coordinate transformation is needed in this case.



So, to summarize, this week we discussed how ultrasound images are formed. We discussed the various imaging modes. We discussed imaging artifacts. We discussed the parts of the transducer, resonant frequency of the ultrasound crystal, etc. And we discussed the basics of ultrasound arrays including linear, phased and curvilinear arrays. So, we will see you in the next week. Thank you.