**Biomedical Ultrasound Fundamentals of Imaging and Micromachined Transducers**

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**Lecture - 02**

Hello, and welcome to Biomedical Ultrasound: Fundamentals of Imaging and Micromachine Transducers. I'm Professor Carla Mercado-Shekhar from the Department of Biological Sciences and Engineering at IIT Gandhinagar. Today, I will provide an overview and introduction to ultrasound imaging.

Let's begin with a brief history of ultrasound and how it has evolved into the technology we know today. In the early 1800s, Jean Colladon determined the speed of sound in water using an underwater bell. Following this, Lord Rayleigh published a foundational work on the theory of sound, which explored the mathematical principles behind acoustic waves. The late 1800s saw the discovery of the piezoelectric effect by the Curie brothers, laying the groundwork for the development of ultrasound transducers.

During World War I, sonar technology emerged, which stands for sound navigation and ranging. This system uses hydrophones to detect sound underwater, originally developed to locate submarines. In the 1940s, World War II led to advancements such as post-echo radar, many principles of which are still utilized in ultrasound imaging today.

In the same decade, Dr. Carl Dusick became the first physician to demonstrate the clinical applications of ultrasound by imaging brain tumors, marking a significant milestone. Researchers like George Ludwig further characterized the speeds of sound waves in various soft tissues, particularly through animal studies.

By the 1980s, ultrasound became more widely used, with advancements in various imaging modes to visualize tissues effectively. I encourage you to explore several references for a deeper understanding of the history of ultrasound.

Now, let's discuss the piezoelectric effect and its significance in ultrasound imaging. This effect is crucial for generating ultrasound, as it utilizes crystalline materials like quartz to produce a potential difference when subjected to mechanical strain.

Subsequently, these piezoelectric materials can generate strain in response to an applied voltage. In a typical ultrasound system, electronic instrumentation sends an ultrasound signal to a piezoelectric element, causing it to vibrate and produce ultrasound waves that propagate into the surrounding medium. This piezoelectric element can function both as a transmitter and a receiver of sound waves, a principle known as reciprocity.

So, what is ultrasound? Ultrasound refers to a wave—specifically, a disturbance or variation that travels through a medium. In the context of ultrasound imaging, the medium of interest is often fluid. Ultrasound is classified as a mechanical or acoustic pressure wave with frequencies exceeding the human range of hearing, typically above 20 kilohertz.

A common imaging mode in ultrasound is brightness mode imaging, or B-mode imaging. This mode produces images in grayscale pixels, representing the backscattered ultrasound waves that indicate tissue reflectivity, essentially measuring how much ultrasound returns to the transducer detector. In diagnostic imaging, frequencies typically range from 2 to 15 megahertz, although diagnostic ultrasound can also operate in the hundreds of kilohertz range for therapeutic applications.

For instance, frequencies around 200 kilohertz to 1 megahertz are used for therapeutic effects like inducing cavitation or thermal ablation. As frequencies increase beyond 15 megahertz, ultrasound can be utilized to image smaller structures, making it suitable for intravascular imaging or small animal studies in preclinical research.

Here are some examples of ultrasound imaging systems currently in use. A typical ultrasound procedure involves a patient lying down while a sonographer holds a transducer to image the internal structures of the body. The ultrasound machine features a monitor, keyboard, and knobs that allow the sonographer to adjust imaging parameters as needed. In the top right, you can see various ultrasound transducers, including a linear array probe and two curved array probes. The choice of probe depends on the specific application for ultrasound imaging.

We also have mobile scanners, showcasing ultrasound's portability. Modern ultrasound technology includes handheld devices that can be connected to a smartphone, allowing for convenient visualization of internal body structures on the go. In the bottom right, there’s an example of a brightness mode (B-mode) ultrasound image displaying the liver and right kidney.

Ultrasound imaging has gained widespread use in clinical settings for several reasons. Firstly, it is exceptionally fast, capable of capturing over 50 frames per second, with advanced systems achieving thousands of frames per second. Additionally, ultrasound is relatively inexpensive compared to other imaging modalities. Importantly, it is safe, as it does not involve ionizing radiation and is non-invasive, meaning there’s no need for surgical procedures to obtain images.

Ultrasound is particularly effective at depicting soft tissue structures, especially dynamic ones. For instance, in Doppler ultrasound imaging, this modality can detect the flow of red blood cells, allowing for real-time visualization of moving structures. The frequencies used in ultrasound also enable excellent submillimeter spatial resolution, which can increase with frequency—allowing imaging down to hundreds of microns within tissue.

However, there are trade-offs, particularly concerning penetration depth, which we will explore later. One of the advantages of ultrasound is its real-time imaging capability, resulting in short acquisition times; a scan can take as little as two minutes, though it may extend to several minutes depending on the specific imaging requirements. This efficiency makes ultrasound convenient and accessible, particularly in rural areas, due to its compact and portable design.

Despite its many benefits, ultrasound imaging has limitations. For example, the images can exhibit low contrast, making it challenging to distinguish between different structures within the tissue. Additionally, the relatively small size of the transducer compared to modalities like CT or MRI leads to a limited field of view, necessitating movement of the transducer across different areas to achieve comprehensive imaging of the target region.

Interpreting structures in ultrasound imaging can sometimes be challenging, as different tissues may appear similar, complicating diagnosis. Additionally, ultrasound is operator-dependent; the way a sonographer positions the transducer and the areas they choose to focus on can significantly influence the quality of the images. This variability means that probe placement is crucial for optimal imaging. To address these operator dependencies, alternative approaches in ultrasound are being developed.

Moreover, ultrasound has limitations in imaging through air, such as in unfilled bladders or lungs, as well as through bone. This limitation relates to the concept of acoustic impedance, which we will cover in later discussions.

Now, let's explore the types of ultrasound waves commonly used in imaging. The first and more prevalent type is the longitudinal wave. In this wave, energy travels in one direction, illustrated by a wave moving from left to right. Each black dot represents particles or small tissue structures that the ultrasound interacts with. As the wave propagates, these particles move in the same direction as the wave, demonstrating the characteristic behavior of longitudinal waves.

Within this wave, there are regions where particles are compressed closely together—known as compression regions—and areas where they are spaced further apart—called rarefaction regions. Another type of wave utilized in ultrasound is in elastography, which we will discuss in a future lecture.

The difference between longitudinal waves and shear waves lies primarily in the movement of the particles. In longitudinal waves, particles move in the same direction as the wave propagation, while in shear waves, also known as transverse waves, particles move perpendicularly to the direction of the wave motion. We will revisit shear waves in a future lecture when we discuss elastography.

Returning to longitudinal waves, these are typically generated by creating vibrations within the tissue. A detector, often made of piezoelectric material, is positioned on the opposite side to capture the ultrasound signals. The resulting waveform, which is usually sinusoidal, is plotted as a function of time. The amplitude of the ultrasound is defined by its pressure, with regions of high pressure where particles are closely packed referred to as compression, and regions of low pressure where particles are spaced apart referred to as rarefaction.

For instance, a piston transducer can excite the medium with a single pulse of ultrasound. The signal detected from this transducer resembles the waveform pattern we discussed earlier.

In ultrasound imaging, short pulses are typically sent, usually consisting of one or two cycles. Here, we can observe a single pulse, characterized by several parameters, including its period. As mentioned earlier, this pulse often takes the form of a sinusoidal wave, with one period corresponding to one complete cycle, denoted as $T$. Given that ultrasound operates at megahertz frequencies, the duration of a single pulse is usually in the microsecond range, often as short as a few microseconds.

When examining two cycles of an ultrasound pulse, the total duration is referred to as the pulse duration. The frequency, defined as the inverse of the pulse period, is measured in hertz. For context, one hertz equals one cycle per second, while one kilohertz equals 1,000 cycles per second. In ultrasound imaging, we typically use frequencies greater than one megahertz, equating to about a million cycles per second.

In clinical applications, such as transcutaneous ultrasound—where images are captured through the skin—frequencies range from 1 to 15 megahertz. For echocardiography, a widely used technique for imaging the heart, frequencies between 1 and 7 megahertz are common. When imaging smaller structures, such as in ophthalmic ultrasound, higher frequencies are employed to achieve better spatial resolution.

In ultrasound imaging, higher frequencies—typically above 10 megahertz—are used for more superficial structures. Later, we will explore how frequency influences the depth of penetration, spatial resolution, and the selection of ultrasound applications within the body. It’s important to note that ultrasound imaging involves not just a single pulse, but multiple pulses. For example, we can illustrate this with a second and third pulse added to our discussion.

The pulsing of ultrasound is characterized by the pulse repetition period, which is the time interval between consecutive pulses, typically measured in microseconds. We also consider the wavelength of ultrasound, which is defined as one cycle of the wave. This wavelength is calculated by dividing the speed of sound—specifically the longitudinal sound speed—by the frequency. The relationship between frequency and time period can also be expressed as the speed of sound multiplied by the time period, yielding the wavelength.

In the literature, you might encounter the term Pulse Repetition Frequency (PRF), which is simply the reciprocal of the pulse repetition period. These parameters are essential for understanding ultrasound imaging. Another important parameter is the duty cycle, which refers to the duration that the pulse is "on" or transmitting. Since ultrasound imaging typically employs a pulse-echo technique, it's crucial to account for the time needed for the echo to return after interacting with the tissue. This means that while one pulse is traveling to the tissue, it must not interfere with the returning echoes from previous pulses.

During the off time, the transducer is not transmitting, allowing it to receive echoes from the ultrasound signals. The duty cycle is defined as the ratio of the on time of the transducer to the total time (on time plus off time), expressed as a percentage by multiplying the ratio by 100. For typical imaging applications, the duty cycle is usually kept below 1%. However, therapeutic ultrasound applications may use duty cycles exceeding 1% to induce specific bioeffects in tissues.

There is an important relationship between time and space in ultrasound imaging, governed by the speed of longitudinal waves. This speed allows for the conversion between time and distance. While pulse duration describes the time component of a wave, spatial pulse length describes the corresponding distance component.

In visual representation, the y-axis can represent time, defined by periods of the wave, while the x-axis represents distance. At different time instances, the wave propagates through space. For example, after one period has passed, the wave moves a distance equivalent to one wavelength. This illustrates how ultrasound waves have both temporal and spatial components, enhancing our understanding of their behavior.

It's important to note that the speed of sound in a medium is also influenced by its compressibility, represented by the bulk modulus, which is a key elastic parameter. Different tissues exhibit various longitudinal sound speeds; for instance, fresh water at 20 degrees Celsius has a sound speed of approximately 1482 meters per second. Given that much of the human body is primarily composed of water, the sound speeds of various tissues fall within a range centered around this value.

Fat, for example, has a lower sound speed than water, while denser tissues like bone can reach sound speeds as high as 4000 meters per second. For soft tissues, diagnostic scanners typically use an average longitudinal sound speed of about 1540 meters per second.

So how is longitudinal sound speed measured? Most of the values reported for tissue sound speeds are empirically derived through experiments. One common method is called through transmission. This setup involves conducting ultrasound experiments in a water tank, which mimics the soft tissue environment, as water makes up a significant portion of the body. The water tank can be adjusted to different temperatures, as sound speed can vary with temperature. A pulse generator is used to excite a piezoelectric transducer, which is an integral part of this measurement process.

In this setup, the ultrasound signal is transmitted through water, with the first piezoelectric transducer converting electrical energy into acoustic energy. The ultrasound wave then propagates through the water, and a second piezoelectric transducer acts as a receiver, capturing the signal. Since ultrasound signals typically have a low amplitude, a preamplifier is used to boost the voltage of the received signal. This amplified signal is then sent to a digital oscilloscope, where it is digitized and analyzed on a computer.

Next, we can introduce a tissue sample into the setup. In this case, the ultrasound wave is transmitted through the tissue, which generally has a different sound speed compared to water. Similar to the previous experiment, the ultrasound wave passes through the tissue, but there may be a reduction in the amplitude of the signal due to attenuation, a topic we will cover in a later lecture. The received signal from the tissue is again captured by the second piezoelectric transducer, digitized, and sent for analysis on the computer.

This configuration is known as a through transmission setup, as it involves transmitting an ultrasound pulse through the tissue and receiving it with another transducer. Examples of the signals recorded on the oscilloscope can be seen here.

In this setup, you can observe the amplitude of the received signal, which can later be converted to pressure. The graph will show a pulse that has traveled through the water tank without the tissue sample. When you plot the signal received from the experiment with the sample, you'll notice that it appears time-shifted by a duration of $\Delta T$. Additionally, the amplitude of this signal will decrease slightly due to acoustic attenuation as it passes through the tissue, a topic we will explore in more detail in a subsequent lecture.

To estimate the longitudinal sound speed of a specific sample, you can use an equation that incorporates the longitudinal sound speed of water ($C_w$ ). This equation references a valuable resource from a journal published in the 1950s, which provides tabulated data on the sound speed of fresh water at various temperatures. This resource can assist you in determining the sound speed of water at a specific temperature.

The equation also considers the time difference, which is the difference in arrival times of the signals with and without the sample. This time difference is divided by the thickness of the sample, which should be measured using a ruler or caliper, as this measurement must be done empirically.

In this lecture, we covered the history of ultrasound, the concept of piezoelectricity, and how different piezoelectric materials generate ultrasound. We explored the advantages and limitations of ultrasound imaging, as well as the types of waves commonly assessed, such as longitudinal and shear waves, which we will discuss further in later lectures. We also examined key pulse parameters that characterize an ultrasound pulse, including period, frequency, and pulse duration.

Lastly, we looked at how to measure the longitudinal sound speed of a material through a practical experimental setup. This concludes today's lecture. In our next session, we will discuss the wave equation.