ULTRASOUND AND PHOTOACOUSTIC IMAGING

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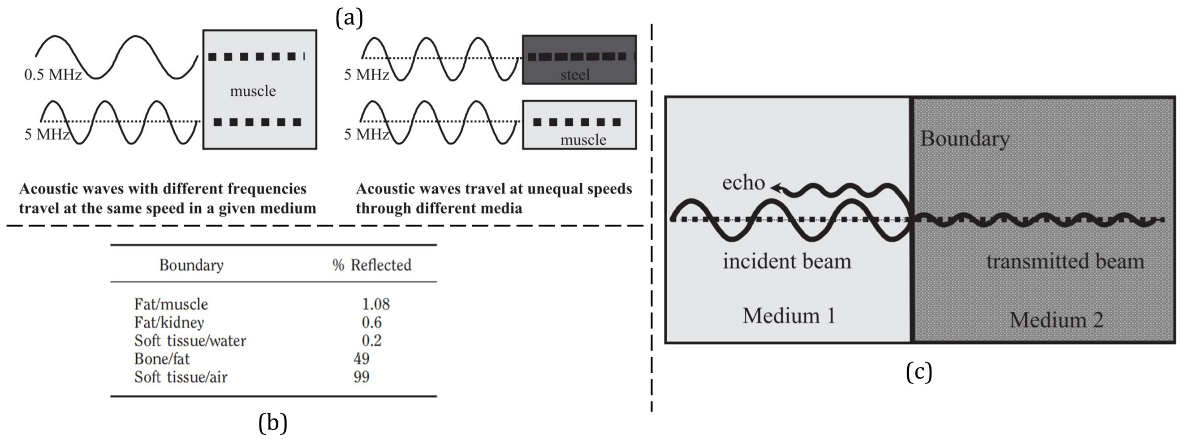

Figure 1: Graphical illustration of the sinusoidal positional embedding. (Images were obtained from [1])

1 Ultrasound Imaging Systems

There are two basic principles regarding how ultrasound is generated and an image is formed. In ultrasound imaging, a transducer generates acoustic waves via the **piezoelectric effect**. When an electric current is applied to the crystals, the crystals deform and vibrate. This deformation produces the ultrasound beam. An ultrasound image is formed via the **pulse-echo principle**. Ultrasound waves are produced in pulses, because the same crystals are used to generate and receive acoustic waves, and they cannot do both at the same time. As the ultrasound pulse enters the patient, it is bounced or reflected back to the transducer. An ultrasound image is formed by measuring the amplitude and the travel time of the reflected sound echoes to reach the transducer. If there is a high acoustic impedance mismatch between two tissue interfaces, then the reflected echo will have a high amplitude resulting in a high intensity pixel appearance in the formed ultrasound image.

1.1 Physics of Ultrasound Imaging

1.1.1 Frequency

The frequency of an ultrasound wave has a meaning of the number of cycles or pressure changes that occur in 1 second. Ultrasound is sound with a frequency $\geq$ 20 kHz (above the frequency of human audible sound). Typical ultrasound frequency used for clinical purpose are in the range of [2 MHZ, 10 MHz].

1.1.2 Propagation speed

The speed at which ultrasound can travel through a medium, which is about *1540* meters/second for soft tissue. The speed depends solely on the density and stiffness of the medium. Images in Fig 1 (a) demonstrate the speed and frequency of ultrasound wave.

1.1.3 Ultrasound Interaction with Tissue

When a beam of ultrasound travels through a medium, a reflection of the beam can happen. This reflection is called an *echo*. The production and detection of the echos form the basis of ultrasound imaging. A reflection occurs at the boundary between two materials, provided that the acoustic impedance of the materials is different. Acoustic impedance is calculated as the product of the tissue density and the speed of the sound wave:

$$Z = \rho \times c, \tag{1}$$

where ρ is the density of the tissue, and c is the speed of the sound. If two materials have the same acoustic impedance, their boundary will not produce a reflection (i.e., echo). If the difference in acoustic impedance is large, a strong echo (i.e., large amplitude) will be produced, and vice versa. If the difference in acoustic impedance is too large, all the ultrasounds will be reflected. As shown in Fig. 1 (b), regions containing bone or air can produce large echos that not enough ultrasound remains to image the tissues underneath. Strong reflection or echos show on the ultrasound image as white and weaker echos as gray.

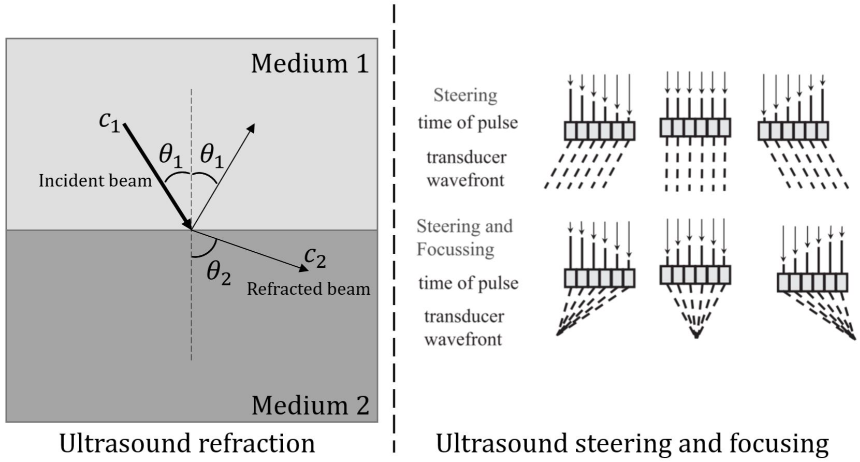

Figure 2: Left: Ultrasound refraction and Snell's law. Right: Ultrasound steering and focusing (image was obtained from [1])

1.1.4 Angle of Incidence

The percentage reflections shown in Fig. 1 (b) are applicable for a normal angle of incidence (90° to the boundary). If a beam of ultrasound strikes a boundary with an angle other than 90°, the echo will return from the boundary at angle equal to the angle of incidence, and the transmitted beam will be deviated from the straight line by an amount that depends on the difference in the velocity of ultrasound. This deviation of the beam traveling direction is known as refraction, and it is governed by the Snell's law (as shown in the left panel Fig. 2):

$$\frac{\sin\theta_1}{c_1} = \frac{\sin\theta_2}{c_2}, \tag{2}$$

where $\sin\theta_1$ and $\sin\theta_2$ are incident and refracted angles of the ultrasound beam, respectively, and c_1 and c_2 denote, respectively, the speed of sound in the first and second medium. Ultrasound machine always assumes that the ultrasound travels in a straight line, therefore, refraction only confuses the machine. However, refraction from rough surfaces produces echoes in many directions, which enables sound to get back to the transducer from oblique surfaces.

1.1.5 Attenuation

The intensity of an ultrasound beam is reduced by attenuation due to reflection, refraction, scattering, and absorption. Reflection and refraction occur at surfaces that are large compared with the wavelength of the ultrasound. For objects that are smaller than the wavelength, the energy of the ultrasound is scattered in many directions and eventually get absorbed as particle vibration and heat generation. Therefore a high-frequency (smaller wavelength) beam will be attenuated more than a lower frequency (larger wavelength). To penetrate deep into the body, one needs to use a lower-frequency transducer (larger wavelength). However, only higher frequencies can enable finer details. Therefore, there is a trade-off between resolution and depth of penetration for different imaging applications.

In ultrasound, relative intensity of the beam is important. Because the change in intensity is so large, the relative change in intensity is measured in decibels (dB):

$$10\log(\frac{\text{transmitted intensity}}{\text{incident intensity}}). \tag{3}$$

A 3-MHz beam will be attenuated 3dB in the first centimeter (decreased to its 50% intensity), then another 3dB in the next centimeter. Therefore, after passing through 20 centimeters of tissue and being reflected back to the transducer, the intensity will be attenuated by a factor of more than a million.

1.2 Transducers

Transducer is the most critical component in any ultrasonic imaging system. Transducers use a piezoelectric material that a characteristic acoustic impedance perfectly matched to that of the (human) body, has high efficiency as a transmitter, and high sensitivity as a receiver. However, due to the 99.9% reflection rate of air/soft-tissue (as shown in

Fig. 1 (b)), transducers must be directly coupled to the patient skin without any air gap. This coupling is accomplished by using gel or oil between transducer and human skin.

1.3 Ultrasound Beams

By using delays in the pulsing of elements, the beam from an array can be steered, i.e., make the beam sweep as in a scanning motion (as shown in the right panel Fig. 2).

1.4 Speckle Noise

Speckle noise comes the acoustic echoes with random phases and amplitudes. The superposition of these echos produces the complicated interference pattern, which is known as speckle noise. Speckle noise is a pseudo-random process, because it generally depends on the structure of the tissue. Speckle noise tends to reduce the image contrast and blur the image details.

2 Photoacoustic Imaging

In photoacoustic (PA) imaging, ultrasound waves are produced by irradiating the tissue with modulated electromagnetic radiation, usually pulsed on a nanosecond timescale. In the case of optical excitation, absorption by specific tissues such as hemoglobin, melanin, or water followed by rapid conversion to heat produces a small temperature rise. This rise of temperature leads to an initial pressure increase, which then subsequently relaxes, resulting in the emission of broadband low-amplitude acoustic waves. The acoustic waves propagate through the tissue to the surface, where they are detected by the ultrasound receiver. By measuring the time of arrival of the acoustic waves and knowing the speed of sound in tissue, a PA image can be reconstructed in the same way that a pulse-echo ultrasound image is formed. The acoustic pressures in PA are several orders of magnitude smaller than that in ultrasound.

2.1 Source of Contrast

In ultrasound, an image represents the acoustic impedance mismatch between different tissues. A PA image, however, is absorption-based. It represents the initial pressure distribution produced by the deposition of the optical energy, which depends on the optical absorption and scattering properties of the tissue [2]. PA imaging can provide greater tissue differentiation and specificity than ultrasound because different tissue types can be much larger in optical absorption than those in acoustic impedance.

Although PA imaging can provide larger tissue differentiation, it comes at a cost, the penetration depth, which is limited due to the strong optical attenuation exhibited by most tissue. To understand the contrast produced by PA imaging, we need to understand how the PA signal is created:

- A pulsed laser light is incident on the tissue surface. Depending on the wavelength, the light penetrates to some depth.
- The light is then scattered and absorbed by the tissue.
- The absorbed laser energy is converted into heat in the tissue, which produces an initial pressure increase of the tissue, and later emits the acoustic waves.
- The acoustic waves propagate back to the surface and are detected by the transducer.

The dominance of optical absorption as the primary source of PA image contrast lends PA imaging to the visualization of anatomical features that contain an abundance of chromophores (i.e., light absorbing molecules) such as haemoglobin, lipids and water.

3 Beamforming in Ultrasound and PA

3.1 Delay and Sum

Delay and Sum (DAS) is the most standard technique in ultrasound and PA beamforming (as shown in Fig. 3). DAS is mathematically defined as [3]:

$$y_{DAS}(k) = \sum_{i}^{M} x_i(k - \Delta_i), \tag{4}$$

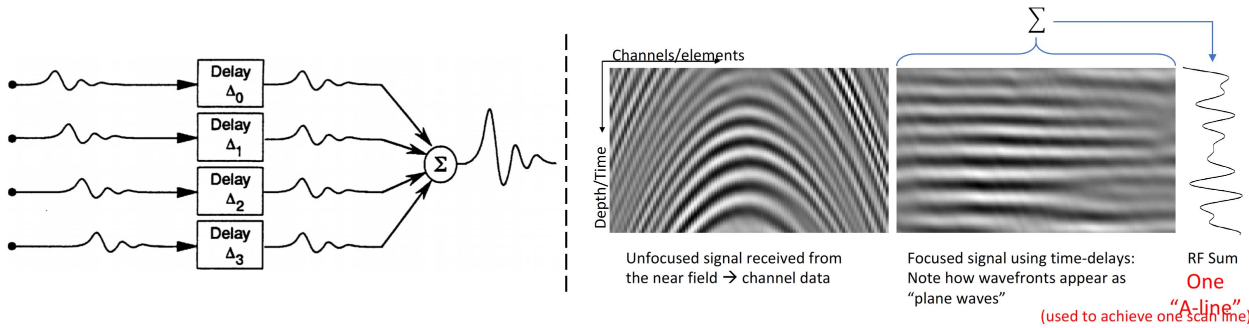

Figure 3: A simple illustration of the delay and sum beamformer.

where $y_{DAS}(k)$ denotes the output of the beamformer of the k^{th} time index (or depth), M denotes the number of the array elements, and $x_i(k)$ and Δ_i are the detected signals and the corresponding time delay for detector i.

3.2 Short-lag Spatial Coherence

Instead of the simple summation, the time delayed signal can be used in a more advanced beamformer, namely the short-lag spatial coherence (SLSC). SLSC beamforming displays the similarity of received signals in the aperture domain, as a function of element separation m. Then, an SLSC image is formed as the integral of the spatial coherence function over the first M lags. It displays spatial coherence between the received echos at different short-lag values (i.e., element separation m), thereby reduces speckle noise and removes clutter artifacts. As shown in Fig. 4, for a receive aperture with N elements, the *time-delayed* signal detected by the i^{th} element of n^{th} time index (or depth) is defined as $s_i(n)$ (note that $s_i(n)$ is a zero-mean signal). The estimated spatial covariance across the receive aperture is defined as [4]:

$$\hat{C}(m) = \frac{1}{N-m} \sum_{i=1}^{N-m} \sum_{n=n_1}^{n_2} s_i(n) s_{i+m}(n), \tag{5}$$

where m is the distance, or lag, which is really the number of elements between two points in the aperture. Notice that signals s_i and s_{i+m} have zero-mean. The covariance, $\hat{C}(m)$, is further normalized by the individual variances of the signals s_i and s_{i+m}, the spatial correlation can be computed by:

$$\hat{R}(m) = \frac{1}{N-m} \sum_{i=1}^{N-m} \frac{\sum_{n=n_1}^{n_2} s_i(n) s_{i+m}(n)}{\sum_{n=n_1}^{n_2} s_i(n) \sum_{n=n_1}^{n_2} s_{i+m}(n)}. \tag{6}$$

It is found that the largest losses in spatial coherence will occur in the regions of low lags [4]. Therefore, the short-lag spatial coherence (SLSC) is computed as the integral of the spatial conherence function over the first M lags:

$$R_{sl} = \int_1^M \hat{R}(m) dm = \sum_{m=1}^{M} \hat{R}(m), \tag{7}$$

where M is a hyperparameter, for which a parameter Q is introduced to represent Q as a percentage of the aperture width N:

$$Q = \frac{M}{N} \times 100\% \tag{8}$$

4 Application of ultrasound and PA in Bone

One of the applications of ultrasound and PA bone imaging is to assist spinal fusion surgeries in real time. Spinal fusion is surgery to permanently connect two or more vertebrae in the spine. It is done by placing screws through the pedicles of vertebrae to connect them with a metal rod and stabilize the spine. It is critical to ensure the correct trajectory during the hole creation process in order to avoid accidental bone breaches and screw misplacement as shown in Fig. 5.

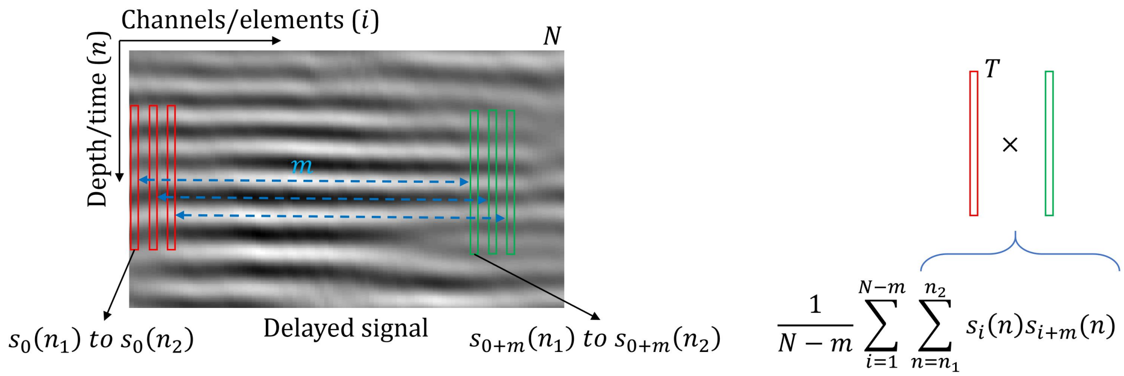

Figure 4: An illustration of the short lag spatial coherence beamforming.

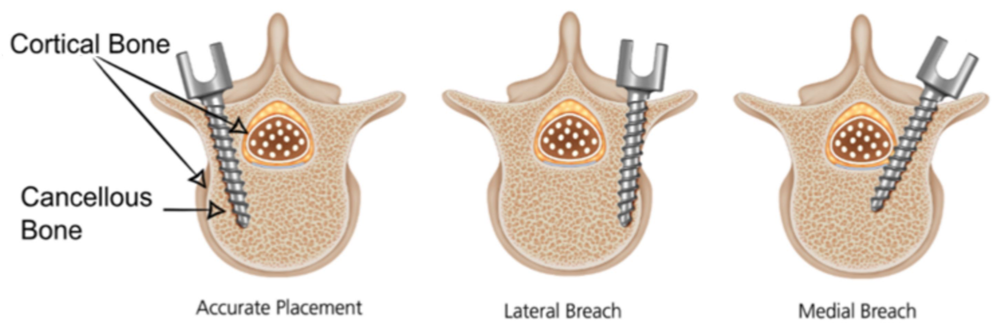

Figure 5: Examples of accurate and inaccurate pedicle screw placement. (Image obtained from [5])

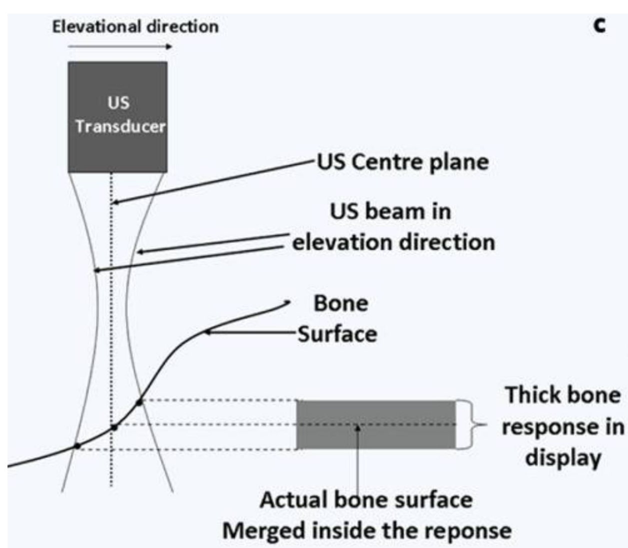

Figure 6: Bone boundary has thickness due to the inclination. (Image obtained from [6])

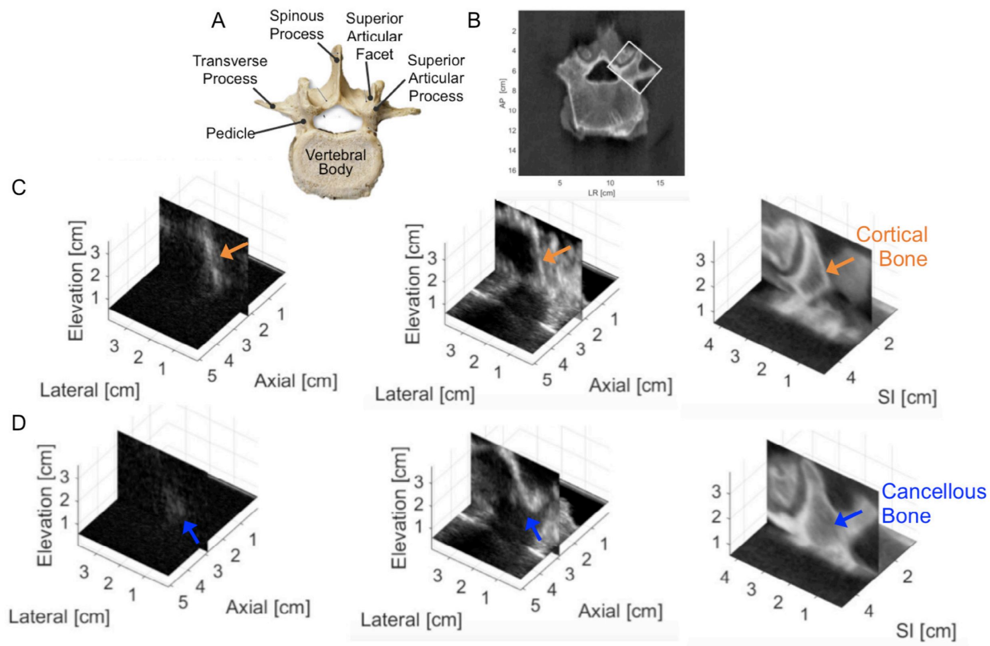

Figure 7: Bone boundary has thickness due to the inclination. (Image obtained from [7])

4.1 Ultrasound of bone

Ultrasound images are formed by measuring the amplitude and the travel time of the reflected sound echoes to reach the US transducer. If there is a high acoustic impedance mismatch between two tissue interfaces the reflected sound echo signal will have a high amplitude resulting in a high intensity pixel appearance in the formed US image. A high-intensity pixel in an US image indicates a strong likelihood of the presence of a boundary, such as soft tissue interface or bone. Since bone tissue has one of highest acoustic impedance values compared to other tissues such as muscle, fat, liver or water, most of the signal is reflected back from the bone interface resulting in a high-intensity feature in the reconstructed US image. This high-intensity bone feature is followed by a region with very low-intensity values appearing right after bone boundary. This low-intensity region, “shadow region”, is one of the typical US imaging artifacts denoting that interior bone surfaces cannot be imaged with US imaging. The high-intensity feature depicting bone boundary response looks like a line with a shape closely resembling the surface. However, compared to bone surface appearance in CT or fluoroscopy imaging, the surface response in US is not a sharp transition region but rather has a thickness which can reach a value of 4 mm in certain cases. The response thickness is affected by the inclination of the image surface with respect to the US transducer. The greater the inclination of the imaged surface, the greater the response thickness (as shown in Fig. 6).

4.2 PA of bone

Photoacoustic imaging has been explored as an option to uncover expected differences between the cortical and cancellous bone for the potential guidance of spinal fusion surgery. An optical fiber that delivers laser light could either be isolated from or attached to the surgical tool. A standard clinical ultrasound probe would then be placed with acoustic coupling gel on the vertebra of interest. The purpose of this ultrasound probe is to receive the acoustic response generated by optical absorption within the blood-rich cancellous core. Although it is hard for ultrasound to travel through cortical bone, it is possible for the acoustic pressure response of cancellous core (resulting from the absorption of the laser pulse) to travel through the 244 μm to 1.75 mm-thick cortical bone layer covering the cancellous core. Therefore, in this application of photoacoustic imaging for surgical guidance, blood-rich regions can be targeted rather than avoided. The difference between the PA images obtained by pointing optical fiber toward the cortical bone and the cancellous core of the pedicle can be observed from Fig. 7. The PA image of the cancellous bone have lower amplitudes and are more diffuse. This difference is achievable because the optical absorption of blood is orders of magnitude higher than that of bone, which permits a photoacoustic response from the bood-rich cancellous core.

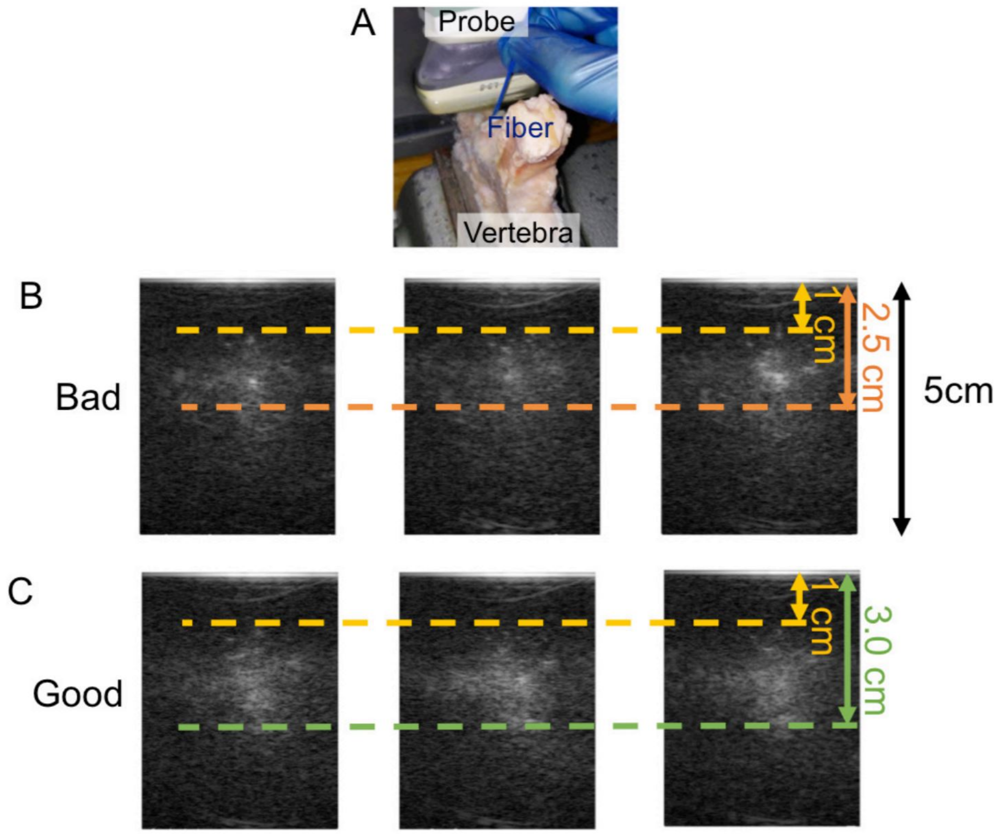

Figure 8: Fiber position affects the appearances of the PA images. (Image obtained from [7])

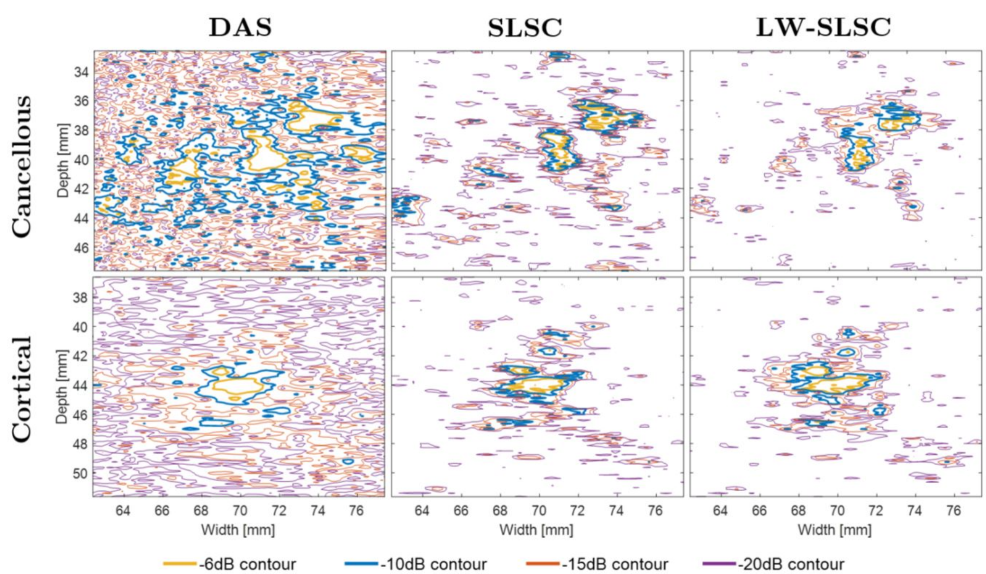

Figure 9: Contour plots of photoacoustic DAS, SLSC, and LW-SLSC images. (Image obtained from [5])

4.2.1 Fiber position matters

The trajectory of fiber position affects the appearance of the resulting PA image (as shown in Fig. 8). Generally, a good fiber position yields deeper signals corresponding to the response from the blood-rich cancellous core.

4.2.2 Distinguish cortical from cancelleous bone

During the spinal fusion surgery, when placing the screw inside the vertebra, it is important to prevent breaching the pedicle wall (which consists of cortical bones). To identify cortical bone from cancellous bone, the laser light pulses can be delivered through the tip of the drill while it is being inserted into the pedicle to create a hole for the screw. The differences in the PA image-appearance of cortical and cancellous bones can be identified. Three beamforming techniques can be used to reconstruct PA images, delay-and-sum (DAS), short-lag spatial coherence (SLSC), and locally weighted short-lag spatial coherence (LW-SLSC). These beamforming methods are described in section 3. As shown in Fig. 9, PA differentiation of cortical and cancellous bones is possible with either of the three deamformers, but using a contour level of -6 dB with DAS is considered to be optimal. The -6 dB contour is optimal because it allows more localized visualization of the PA response from the fiber tip without confusing this response from that of surrounding tissue. On the other hand, coherence-based PA imaging enables the localization of fiber tips for tip tracking applications. The improved localization of the tool tip can be appreciated by estimating the centroid of the -6 dB contour levels for the SLSC and LW-SLSC images.

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