# AAPM Tutorial

## Single-Element Transducers[1]

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A complete understanding of the numerous types of diagnostic ultrasound (US) transducers starts from a thorough understanding of single-element circular transducers. This class of transducers provides the basis for understanding fundamentally important concepts such as the piezoelectric effect, crystal resonance, thickness mode resonance, backing, matching layers, focusing, and spatial resolution. An understanding of these principles will assist radiologists in using the full potential of the various diagnostic US transducers.

### INTRODUCTION

The clinical demand to provide improved diagnostic ultrasound (US) resolution has resulted in creative transducer developments. Fundamental understanding of the principles of ultrasonic wave phenomena necessitated that the transducer be closer to the structures being imaged, hence, the development of intracavitary transducers. These new classes of ultrasonic transducers have allowed the user to position the transducer much closer, and in some cases immediately adjacent, to the structure of interest.

This article describes the physics behind how these new transducers produce better images through the understanding of single-element transducers, the ultrasound fields they produce, and their basic principles of spatial resolution.

**Abbreviations:** $FWHM_A$ = full width at half maximum axial, $FWHM_L$ = full width at half maximum lateral

**Index terms:** Physics • Ultrasound (US), physics

**RadioGraphics** 1993; 13:947–957

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## SINGLE-ELEMENT PIEZOELECTRIC TRANSDUCER

### Piezoelectric Effect

For all diagnostic US applications, the transducer is the heart of the imaging system. The term ***transducer*** refers to any object that transforms or changes one type of energy into another. For example, a loud speaker converts electrical energy into mechanical energy, an audible pressure wave, and a microphone converts an audible pressure wave into an electrical signal. A US transducer converts mechanical energy, the ultrasound pressure wave, into electrical energy and can also perform the reverse process. The phenomenon that explains this unique transformation is called the *piezoelectric effect.*

In the early 1880s, Pierre and Jacques Curie, better known for their discoveries in radiation physics, discovered that certain crystals develop an electric charge when they are mechanically strained. The term *piezoelectricity* means pressure electricity and derives from the Greek word *piezo,* meaning pressure. When a voltage is applied across a piezoelectric crystal (Fig 1), the thickness of the crystal either expands or compresses, depending on the polarity of the voltage and the molecular configuration of the crystal. Piezoelectric transducers are reciprocal devices in that they operate both by converting electrical energy into acoustic energy (pressure waves) to generate ultrasound waves and by converting the received pressure waves from echoes into electrical signals to be processed into useful diagnostic information.

### Thickness Mode Resonance

The thickness of the crystal in a single-element piezoelectric transducer strongly affects the center frequency of the diagnostic US system by the mathematical expression

$$f_r = c_x/2l_x,$$

where $f_r$ is the preferred or resonant frequency, $c_x$ is the crystal speed of sound and is in the range of about 5,000 m/sec, and $l_x$ is the thickness of the crystal.

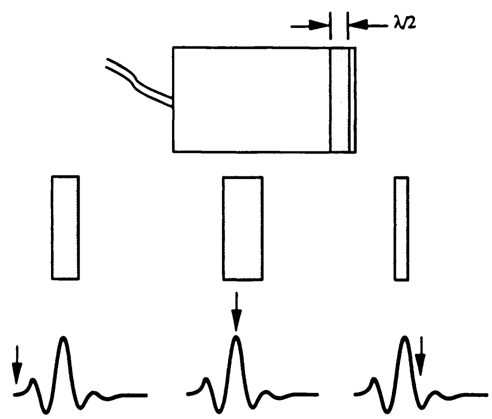


**Figure 1.** Demonstration of the piezoelectric effect. Shaded area represents the piezoelectric crystal in a single-element transducer. When there is no voltage applied across the piezoelectric crystal as indicated by the arrow (left waveform), the thickness of the crystal is in its resting state. When a large positive voltage is applied across the crystal (arrow location for middle waveform), the crystal thickness expands, and when a negative voltage is applied across the crystal (arrow location for right waveform), the thickness contracts. Crystal thickness strongly affects the resonant frequency of the single-element piezoelectric transducer. The preferred frequency is when the thickness of the crystal in a single-element transducer is one-half of the wavelength of the crystal, λ.

The resonant frequency is principally determined by the thickness of the crystals in single-element transducers, hence the terms ***half-wavelength resonance*** and ***thickness mode resonance,*** that is,

$$l_x = \lambda/2,$$

where $\lambda$ is the wavelength of sound in the crystal. In other words, the crystal is at resonance when its thickness is half-wavelength. For example, if the crystal speed of sound is 5,000 m/sec and a resonant frequency of 5 MHz is desired, the half-wavelength thickness of the crystal is

$$l_x = \lambda/2 = \frac{c_x}{2f_r} = \frac{5{,}000 \text{ m/sec}}{2 \times 5 \text{ MHz}} = 0.5 \text{ mm}.$$

The electrical excitation of the crystal can be compared with the striking of a tuning fork, which has a preferred frequency when struck. The electrical excitation from the pulser of the US system imparts energy into the single-element crystal, which in turn causes the crystal to vibrate at a preferred frequency, the resonant frequency. Thus, the manufacturer of single-element diagnostic US transducers can control the resonant frequency of the transducer by carefully determining the thickness of the crystal material.

## Backing Layer

When the crystal vibrates at a preferred frequency, it will continue to vibrate until all of the energy that was imparted into it is dissipated and the crystal returns to its resting state. Consider your best china glass (Fig 2). If you tap it lightly with a piece of metal, you will hear it ring for a relatively long time, since it takes a long time for the energy imparted by the tap to be dissipated as audible sound. On the other hand, if you tap this same glass when it is filled with water, the sound you hear will be of much shorter duration. Some of the energy imparted from the tap into the glass is converted into the sound you hear, and some is converted into the water as heat.

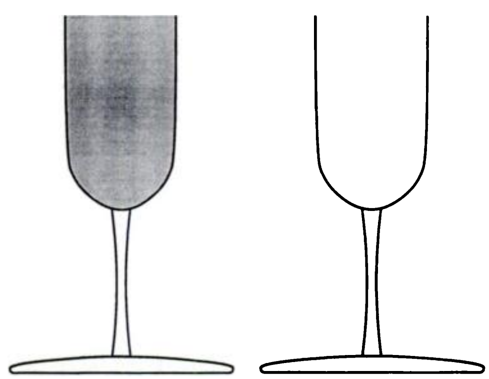

**Figure 2.** Effect of backing, illustrated with a fine china glass. Tap the glass lightly with a piece of metal and listen for how long it takes for the audible sound to dissipate. Repeat the experiment when the glass is filled with water and listen again. This demonstrates the purpose of backing in single-element US transducers, with water being the model for the transducer backing. Backing affects how fast the vibrating crystal element returns to its resting state. The more efficient the backing, the faster the vibrations of the crystal dissipate. The trade-off is dividing the vibrating energy between the propagated pressure wave and the backing.

The backing of the crystal has essentially the same effect. The electrical excitation to the crystal imparts energy into the crystal. Some of that energy from the crystal is converted into ultrasound pressure waves, and some is converted into the backing material in the form of heat. The effect is to cause the energy to be dissipated from the crystal at a much faster rate than would occur without the backing.

There are, of course, trade-offs. As more of the energy from the vibrating single-element crystal from the electrical excitation goes into the backing, there is less energy to go into the propagated pressure wave (Fig 3). Because US transducers operate in both transmit and receive modes, some energy is lost into the backing not only at transmission but also at receipt. This affects the overall sensitivity of the US transducer. Also, to achieve good axial resolution, the crystal must dissipate its energy quickly. So, the overall objective is to have the energy in the crystal removed quickly without affecting sensitivity. This is accomplished with matching layers.

## Matching Layer

When there is a large difference in ***characteristic acoustic impedance*** ($Z = \rho c$, where $\rho$ is density and $c$ is speed) between two objects, such an impedance mismatch results in a large reflection of the incident ultrasound wave. Such a large mismatch exists between the impedances of the crystal of a transducer ($Z_{crystal} > 10$ Mrayls) and tissue ($Z_{tissue} \approx 1.5$ Mrayls). In a US transducer, this results in a substantial amount of the energy in the crystal being reflected at the tissue surface back into the transducer.

To decrease this mismatch between the crystal and tissue, a sophisticated layer of material is put on the surface of the crystal, called a ***matching layer*** (Fig 4). One type of matching layer is called a ***quarter-wavelength matching layer.*** Its impedance is $\sqrt{Z_{crystal} Z_{tissue}}$, and its thickness is a quarter of a wavelength of the matching material. Under these conditions, the quarter-wavelength matching layer propagates 100% of the energy from the crystal into the tissue, but there is a problem. Very

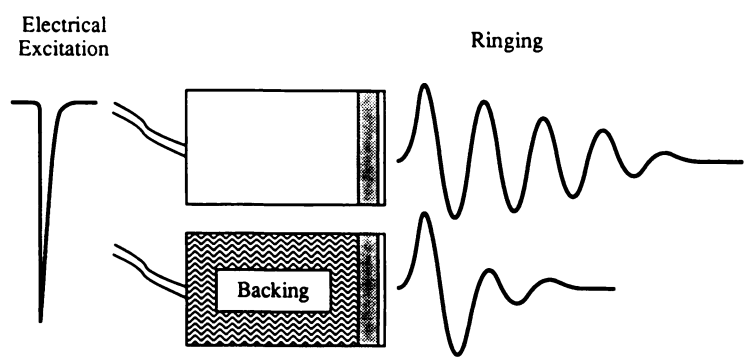


**Figure 3.** Demonstration of the effect of backing of the transducer crystal. The top diagram shows that without backing, it takes much longer for the energy to be dissipated from the crystal into pressure waves. With backing, it takes a much shorter time for the energy to be dissipated from the crystal into pressure waves.

short diagnostic ultrasound pulses do not contain a single frequency, and the quarter-wavelength matching layer works best at a specific single frequency. So, to approach a match that will transmit 100% of the energy, a *tapered matching layer* is used with diagnostic US transducers. A tapered matching layer contains a number of layers of materials that basically taper the impedance from that of the crystal to that of tissue.

Tapered matching layers are so good today that modern transducers do not need backing. Remember, backing is intended to remove the energy from the crystal as quickly as possible, and, with modern matching layers, this is accomplished so that the energy in the crystal is converted very quickly into the transmitted pressure wave. The effect is to improve the sensitivity of the transducer (no backing layer required), while maintaining the very short duration pulses required for good axial resolution.

## BANDWIDTH AND QUALITY FACTOR

In addition to describing ultrasound waveforms (or signals) in terms of their temporal representation, we can also describe them in terms of their frequency representation (frequency spectrum). In the extreme, a continuous-wave temporal waveform contains only one frequency. As the temporal duration of the waveform is decreased, more and more frequencies are contained within that signal. In the frequency spectrum, the range of ultrasound frequencies is termed the *bandwidth*. A relatively long-duration waveform (Fig 5) contains a relatively limited number of frequencies, that is, a narrow band of frequencies, and this is referred to as a *narrow-band* signal. A relatively short-duration waveform is termed a *broad-band* signal because it contains a broad distribution of frequencies.

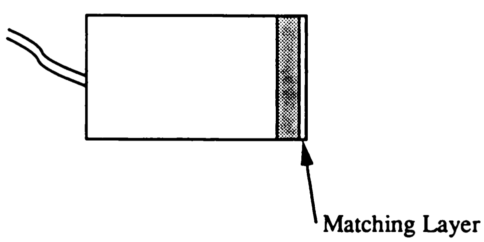


**Figure 4.** The purpose of a matching layer is to maximize the transfer of energy in the transducer crystal into the propagated pressure wave as quickly as possible. With modern matching layers, transducers no longer need a backing layer.

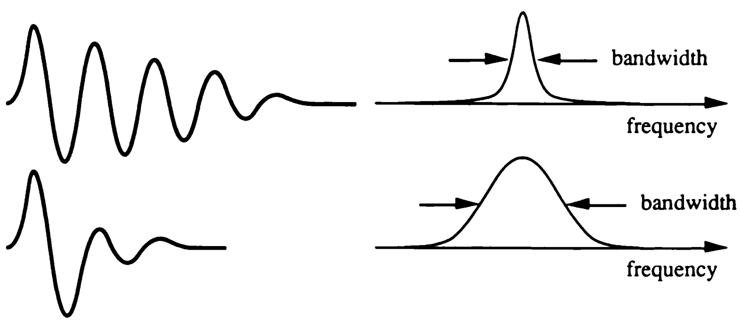


**Figure 5.** Relationship between the temporal waveform of two pulses and their frequency spectra. The longer the temporal duration, the narrower the frequency spread. Long-duration waveforms are narrow-band signals because they contain a narrow band of frequencies, and short-duration waveforms are broad-band signals because they contain a broad distribution of frequencies.

The quality factor $Q$ refers to the damping (rate of decay) of an ultrasound signal and represents how fast the transducer loses energy. The value of $Q$ for a weakly damped transducer (relatively long-duration signal) is a relatively high number, whereas that for a heavily damped transducer (relatively short-duration signal) is a relatively low number. Mathematically, $Q$ is proportional to the energy stored in the transducer and inversely proportional to the energy lost from the transducer, that is,

$$Q \propto \frac{\text{energy stored in transducer}}{\text{energy lost from transducer}}.$$

There is a mathematical relationship between the three quantities of $Q$, bandwidth ($\Delta f$), and resonant frequency ($f_r$), which is

$$Q = fr/\Delta f.$$

By knowing any two of these quantities, the third can be computed. A typical value of $Q$ for an imaging-quality single-element transducer is about 2. If the operating frequency of a diagnostic transducer is 7.5 MHz, the bandwidth is calculated to be 3.75 MHz, of 50% of the resonant frequency (we might refer to this transducer as having a 50% bandwidth).

In summary, for a single-element transducer that produces good-quality diagnostic US pulses in which more energy is dissipated in the backing (if backed) or in which the matching layer is very efficient, the temporal duration of the signal is relatively short, the value of $Q$ is low (termed *low* $Q$), and the signal is broad band.

In general, for a diagnostic US system, $Q$ is not used to refer to the transducer (although that is not incorrect if it is known), but rather it is used to refer to the overall system performance and is thus termed *system* $Q$. It has the same meaning as above and incorporates not only the transducer performance but also that of the rest of the system.

## FOCUSING PRINCIPLES AND SPATIAL RESOLUTION

The classic engineering trade-off of diagnostic US instrumentation is that between resolution and the depth of the image (depth of penetration). Both are directly affected by the ultrasonic frequency. As frequency is increased, resolution improves and penetration decreases. Resolution improves because the ultrasonic wavelength in tissue decreases (becomes a smaller number). Wavelength is inversely related to frequency; increase one and the other decreases, that is,

$$\lambda = c/f.$$

As frequency increases, the ultrasonic attenuation also increases. Penetration is directly affected by the tissue attenuation coefficient, which, in turn, is directly related to frequency. At an ultrasonic frequency of 1 MHz, an "average" attenuation coefficient for soft tissue is approximately 0.7 dB/cm, whereas at 2 MHz, it is 1.4

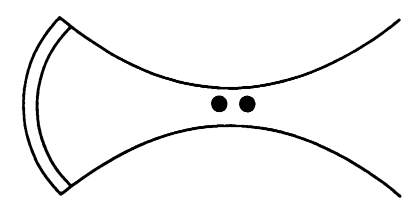

**Figure 6.** Pictorial example of axial resolution for a focused circular single-element transducer. Axial resolution is the ability to distinguish (ie, image) the two axially positioned targets (circles).

dB/cm. Thus, the attenuation coefficient is directly related to frequency; increase one and the other increases. Thus, the attenuation coefficient can be normalized to frequency as 0.7 dB/cm-MHz.

Resolution is the ability to image or resolve discrete structures. Resolution is determined by many components and properties of the instrumentation and patient, including transducer type, beam geometry, frequency, and bandwidth; receiving and processing electronics; video monitor; and tissue attenuation and sound speed. For simplicity, it is easier to understand resolution by considering two types of resolution, axial and lateral.

## Axial Resolution

Axial resolution (also known as range resolution or depth resolution) is the ability to resolve discrete structures along the beam axis (Fig 6). Quantitatively, it is represented as the minimum distance between two structures at different ranges at which both can just be discretely identified as two separate structures. The best axial resolution, as a first-order approximation, is represented by the expression

$$\text{best axial resolution} = SPL/2 = N\lambda/2,$$

where *SPL* is the spatial pulse length (the space occupied by the pulse) and $N$ is the number of cycles of the pulse. The wavelength $\lambda$ is the space occupied by one cycle. The transducer design affects the minimum number of cycles. More highly damped transducers (eg, low-$Q$ transducers) produce very few cycles of ultrasound when excited by the pulser voltage. If $N = 3$, at ultrasonic frequencies of 3.5 MHz ($\lambda = 0.44$ mm) and 7.5 MHz ($\lambda = 0.21$ mm), the best axial resolutions are 0.67 mm and 0.32 mm, respectively. As the frequency increases, and other quantities remain constant, axial resolution improves.

The term ***best axial resolution*** has been employed because, in practice, the receiving and processing electronics affect axial resolution, as does the quality of the video monitor. The electronics and monitor are often lumped into the system $Q$. Low-valued system $Q$s provide better axial resolution than do high-valued ones. As a rule of thumb, the relation between the number of cycles and the system $Q$ is $N = 1.9\,Q$ such that the best axial resolution can be expressed as

$$\text{best axial resolution} = 0.95\,Q\lambda.$$

## Lateral Resolution

Lateral resolution is the ability to resolve discrete structures perpendicular, or lateral, to the beam axis (Fig 7). Quantitatively, lateral resolution is represented as the minimum distance between two side-by-side structures at the same range at which both can just be discretely identified as two separate structures. The best lateral resolution, as a first-order approximation, is represented by the expression

$$\text{best lateral resolution} = \text{minimum beam width}.$$

The term ***best lateral resolution*** is employed here for the same reasons that best axial resolution was used.

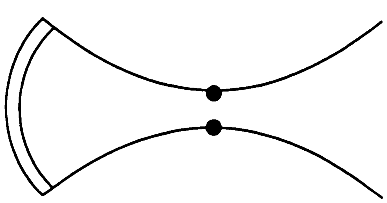

**Figure 7.** Pictorial example of lateral resolution for a focused circular single-element transducer. Lateral resolution is the ability to distinguish (ie, image) the two laterally positioned targets (circles).

Figure 8 shows the beam width of two, unfocused, circular source transducers (operating at the same frequency) of different radii, $a_1$ and $a_2$, where $a_2$ is greater than $a_1$. For the $a_2$ transducer, the distance of the near field is longer, but the beam width is also wider. In other words, the best lateral resolution of the $a_2$ transducer is worse than that of the $a_1$ transducer in the near field. However, in the far field, at a sufficient range, the best lateral resolution is worse for the $a_1$ transducer, demonstrating that lateral resolution is a function of imaging depth for unfocused, circular source transducers.

The range of the *near field* (also known as the *Fresnel zone*) is a function of transducer dimensions and wavelength λ by the expression

$$\text{near field range} = a^2/\lambda,$$

where $a$ is the transducer radius (Table 1). For an unfocused, circular source transducer, the best lateral resolution in the near field is affected mainly by transducer radius. Wavelength also affects lateral resolution in terms of maintaining the same lateral resolution over the near field range.

When an ultrasonic field is focused, the focal range occurs in the near field of the transducer. Figure 9 shows the beam width from the same transducer operating at the same frequency but for two different focal lengths (focal length is the distance along the beam axis from the transducer to the focus). For the longer focus length,

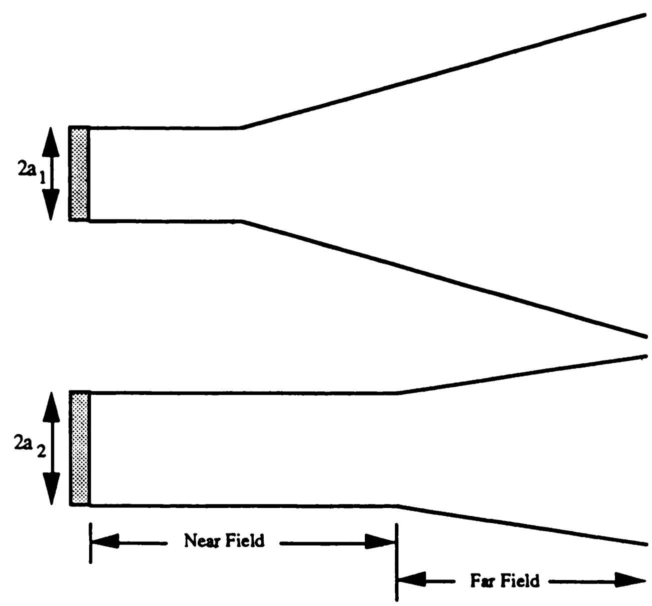


**Figure 8.** Diagram of two unfocused circular single-element transducers of different radii, $a_1$ and $a_2$, where $a_2$ is greater than $a_1$, each operating at the same frequency. Note that the near field (Fresnel zone) is longer for the lower transducer.

**Table 1**
**Comparison of Frequency, Wavelength, and Near Field Length for an Unfocused, Circular Transducer**

| Frequency (MHz) | Wavelength (mm) | Near Field Length (cm) |
|---|---|---|
| 2.5 | 0.62 | 14.7 |
| 5.0 | 0.31 | 29.3 |
| 7.5 | 0.21 | 44.0 |

Note.—Propagation speed was 1,540 m/sec; transducer diameter and radius were 19 mm and 9.5 mm, respectively.

the minimum beam width is greater than that for the short-focus case. The best lateral beam width at focus ($BW$) is directly proportional to wavelength ($\lambda$) and focal length ($ROC$, which stands for radius of curvature) and is inversely proportional to the transducer diameter ($D$), that is,

$$BW = (1.4\ \lambda\ ROC)/D.$$

In imaging terminology, f-number ($f^{\#}$) is often used to quantitate focusing, where the lower the f-number value, the better is the focusing. The best lateral beam width at focus is related to the f-number by

$$BW = 1.4\ \lambda\ (ROC/D) = 1.4\ \lambda f^{\#},$$

where, at the same frequency, beam width at focus can be improved by decreasing the f-number (the ratio of focal length to transducer diameter). Another way to improve lateral resolution at the focus is to decrease the wavelength (increase frequency) (Tables 2, 3).

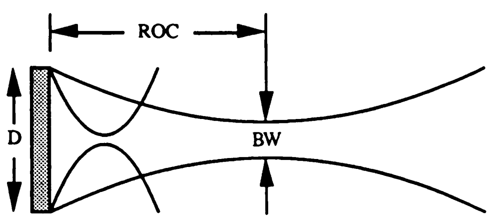


**Figure 9.** Diagram of a focused circular transducer of diameter $D$ operating at the same frequency but for two different focal lengths (focal length is the distance along the beam axis from the transducer to the focus). $ROC$ = radius of curvature of the transducer surface, $BW$ = beam width at the focus.

**Table 2**
**Comparison of Frequency, Wavelength, Focal Length, and Beam Width for a Focused, Circular Transducer**

| Frequency (MHz) | Wavelength (mm) | Focal Length (mm) | Beam Width (mm) |
|---|---|---|---|
| 2.5 | 0.62 | 38 | 1.72 |
| 5.0 | 0.31 | 38 | 0.86 |
| 7.5 | 0.21 | 38 | 0.57 |

Note.—The f-number value was held constant at 2, which results in a constant focal length of 38 mm for a transducer diameter of 19 mm. Propagation speed was 1,540 m/sec.

**Table 3**
**Comparison of Frequency, Wavelength, Focal Length, and Beam Width for a Focused, Circular Transducer for Three f-Number Values of 1, 2, and 4**

| Frequency (MHz) | Wavelength (mm) | Focal Length (mm) | Beam Width (mm) |
|---|---|---|---|
| 5.0 | 0.31 | 19 | 0.43 |
| 5.0 | 0.31 | 38 | 0.86 |
| 5.0 | 0.31 | 76 | 1.73 |

Note.—The frequency was held constant at 5 MHz, which results in a constant wavelength of 0.31 mm for a transducer diameter of 19 mm. Propagation speed was 1,540 m/sec.

A more accurate view of axial and lateral resolution takes into consideration the fact that diagnostic US B-scan images are speckle images. This approach considers resolution from a statistical point of view. The axial resolution depends on the overall system bandwidth $\Delta f$ and is represented in terms of the full width at half maximum axial ($FWHM_A$) quantity, and the lateral resolution depends on the aperture diameter ($D$), center or resonant frequency ($f_r$) and focal length ($FL$) in terms of the full width at half maximum lateral ($FWHM_L$) quantity as

$$FWHM_A = 1.37/\Delta f$$

$$FWHM_L = c\,FL/f_r D,$$

for which numerical examples are shown in Tables 4 and 5.

In summary, axial resolution is affected by the wavelength, number of cycles per pulse, and system $Q$. As axial resolution improves, the wavelength decreases (frequency increases), the number of cycles per pulse decreases, and the system $Q$ decreases. Lateral resolution is affected by the wavelength, transducer size and geometry (focusing), and focal range. As lateral resolution improves, the wavelength decreases (frequency increases), transducer size increases, and the focal range decreases. Away from the focal region axially, the lateral resolution quickly deteriorates.

There are many trade-offs to consider for the best resolution, and a number of these trade-offs take into account the specific geometric requirements of the patient. That is why there are so many transducer designs: Each is designed to optimize image quality for a specific clinical requirement.

**Table 4**
**Axial Resolution in Terms of $FWHM_A$**

| Frequency (MHz) | Wavelength (mm) | Bandwidth (%) | $FWHM_A$ (mm) |
|---|---|---|---|
| 2.5 | 0.62 | 40 | 1.37 |
| 2.5 | 0.62 | 60 | 0.92 |
| 5.0 | 0.31 | 40 | 0.69 |
| 5.0 | 0.31 | 60 | 0.46 |
| 7.5 | 0.21 | 40 | 0.46 |
| 7.5 | 0.21 | 60 | 0.31 |

Note.—Assumes overall system bandwidths $\Delta f$ of 40% and 60%.

**Table 5**
**Lateral Resolution in Terms of $FWHM_L$**

| Frequency (MHz) | Wavelength (mm) | Focal Length (cm) | $FWHM_L$ (mm) |
|---|---|---|---|
| 2.5 | 0.62 | 5 | 1.54 |
| 2.5 | 0.62 | 10 | 3.08 |
| 5.0 | 0.31 | 5 | 0.77 |
| 5.0 | 0.31 | 10 | 1.54 |
| 7.5 | 0.21 | 5 | 0.51 |
| 7.5 | 0.21 | 10 | 1.03 |

Note.—Propagation speed was 1,540 m/sec; aperture diameter was 2 cm.

## SUGGESTED READINGS


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