SECTION 4
Shock-Wave Lithotripsy
CHAPTER 49
Physics of Shock-Wave Lithotripsy

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Introduction

Shock-wave lithotripsy (SWL) was introduced in the 1980s for the treatment of urinary stones and earned near-instantaneous acceptance as a first-line treatment option [1]. At that time, SWL revolutionized treatment in nephrolithiasis worldwide and within one decade it was reported that more than 85% of patients in Europe and the USA were treated with lithotripsy [2, 3]. Since then, the use of SWL has been in decline and it has been estimated that around 70% of kidney stones were treated using SWL in 2002 [4] but less than 50% in 2005 [5]. Over the years lithotripsy has undergone several waves of technologic advance, but with little change in the fundamentals of shock-wave generation and delivery. That is, lithotripters have changed in form and mode of operation from a user perspective, and in certain respects the changes have been dramatic, but the lithotripter pressure pulse is still essentially the same. Lithotripters produce a signature waveform, an acoustic shock wave. This pressure pulse, or shock wave, is responsible for breaking stones. However, it is also responsible for collateral tissue damage that in some cases can be significant [6–11].

Lithotripters produce a powerful acoustic field that results in two mechanical forces on stones and tissue: (1) direct stress associated with the high amplitude shock wave; and (2) stresses and microjets associated with the growth and violent collapse of cavitation bubbles. Recent research has made significant advances in determining the mechanisms of shock-wave action, but the story is by no means complete. What fuels this effort is the realization that a totally safe, yet effective lithotripter has yet to be developed. For example, one trend in the evolution of SWL has been the development of lithotripters that produce very high amplitude and tightly focused shock waves, motivated by the desire to deliver more energy directly onto the stone. However, clinical data suggest that the use of tightly focused shock waves results in increased adverse effects and higher retreatment rates [4, 12–16], and basic research suggests that many shock waves miss the stone [17] and those that do interact with the stone do not result in effective stone fragmentation [18, 19]. In the past few years, manufacturers have started to widen the focal zone of their lithotripters, a move that takes advantage of new findings on the mechanisms of stone breakage with the potential to improve performance (see Focal zone of the lithotripter, below).

A major objective within the lithotripsy community is to find ways to make SWL safer and more efficacious. The perfect lithotripter may not exist, so urologists are left to determine how best to use the machines at hand. One step toward improving outcomes in SWL is to have a better understanding of how current machines work.

This chapter introduces the basic physical concepts that underlie the mechanisms of shock wave action in SWL. Our aim is to give the background necessary to appreciate how the design features of a lithotripter can affect its function. We also present a synopsis of current theories of shock-wave action in stone breakage and describe recent developments in lithotripter technology.
Characteristics of a lithotripter shock wave

A typical shock wave measured at the focus of a lithotripter is shown in Figure 49.1A. The wave is a short pulse of about 5 μs duration [1 microsecond (μs) = 1 millionth of a second]. In this example the wave begins with a near instantaneous jump to a peak positive pressure of about 40 MPa [1 megaPascal (MPa) is about 10 atmospheres of pressure]. This fast transition in the waveform is referred to as a “shock.” The transition is faster than can be measured and is less than 5 ns in duration [1 nanosecond (ns) is 1 billionth of a second]. The pressure then falls to zero about 1 μs later. There is then a region of negative pressure that lasts around 3 μs and has a peak negative pressure of around −10 MPa. The amplitude of the negative pressure is always much less than the peak positive pressure and the negative phase of the waveform generally does not have a shock in it, i.e. there is no abrupt transition. The entire 5-μs pulse is generally referred to as a shock wave, shock pulse or pressure pulse; however, technically it is only the sharp leading transition that is formally a shock.

Figure 49.1B shows the amplitude spectrum of the shock pulse, i.e. the different frequency components in the pulse. It can be seen that a lithotripter shock wave does not have a dominant frequency or tone, but rather its energy is spread over a very large frequency range; this is a characteristic feature of a short pulse. It can be seen that most of the energy in the shock wave is between 100 kHz and 1 MHz. This means the analogy of an opera singer shattering a crystal glass by singing at the right pitch (single frequency) is not appropriate for lithotripsy due to the multitude of frequencies present.

The waveform shown in Figure 49.1A was measured in an electrohydraulic lithotripter (a description of different types of shock wave generators is given below). Most lithotripters produce a similar shaped shock wave and in Figure 49.2A waveforms measured in an electrohydraulic lithotripter and an electromagnetic lithotripter at a high power setting are compared. It can be seen that both waveforms have a form that is common to all shock waves used in lithotripsy: a high amplitude compressive phase of extremely rapid transition (<1 ns) and short duration (∼1 μs) followed by a trailing tensile phase (∼3 μs). In this case the amplitude of the peak positive pressure of the electromagnetic lithotripter is 2.5 times that of the electrohydraulic lithotripter. In general, pressure amplitudes are machine and setting specific, and the peak positive pressure can vary from 20 to 110 MPa and the negative pressure from −5 to −15 MPa. For example, Figure 49.2B shows waveforms measured at lower power settings of both machines, and again the waveforms are similar but the peak amplitudes differ by less than 50%. Further, the spatial dependence of the pressure field is also machine and setting specific and this will be discussed below.

Acoustics primer for shock-wave lithotripsy

What is an acoustic wave?

An acoustic wave, or sound wave, is created whenever an object moves within a fluid (a fluid can be either a gas or liquid). In Figure 49.3 it is shown that as an object moves, it locally compresses the fluid that surrounds it, i.e. the molecules are forced closer together. The region of compressed molecules in turn pushes against the molecules next to it. This relieves the compression in the first region but leads to a new compressed region. The molecules in the second region then start to compress the next on a log scale. The peak of the amplitude response is around 300 kHz, which corresponds to a duration of 4 μs. It can be seen that this persists until frequencies of about 20 MHz.

![Figure 49.1](image-url) (A) A pressure waveform measured at the focus of an electrohydraulic lithotripter (Dornier HM3). (B) Fourier transform of the waveform in A showing how the energy is distributed as a function of frequency. Both axes are shown
pressed. For the case where the object moves away from the fluid, there is a resulting rarefaction of the molecules, i.e. the moving object leaves a partial vacuum. In this case, the neighboring molecules will move to fill the void, leaving a new region of rarefraction. This continues one region to the next and the rarefractional disturbance propagates through the medium as a tensile acoustic wave. In most cases a tensile wave propagates just like a compressive wave and with the same sound speed.

Typical acoustic sources, such as audio speakers, vibrate backwards and forwards. This produces alternating compression and rarefraction waves that are referred to as the compressive phase and tensile phase of the acoustic wave. Often the waveform is sinusoidal in nature. Note, however, that the majority of acoustic waves, including the acoustic pulses generated in lithotripsy, are not sinusoidal in form. For small-amplitude waves (linear acoustics) every point of the waveform moves at the same speed, the sound speed $c_0$. This is a material property and for water and tissue it is about 1500 m/s. It will be shown later that for large amplitude (nonlinear) acoustic waves, such as shock waves, the adjacent region and so on; it is thus that a “wave” of compression travels through the fluid. This is an “acoustic wave” and the speed of wave propagation (called the sound speed) is a material property of the medium; for air it is about 340 m/s, and in water and most soft tissues in the body it is about 1500 m/s. Note that individual molecules do not travel with the acoustic wave, rather they just jostle their adjacent neighbors. Therefore, for an acoustic wave to propagate, there must be a medium present that can support the vibrations. This is an important physical difference between classical waves (e.g. acoustic waves, seismic waves, water waves) and electromagnetic waves (e.g. light, radio waves, X-rays). For electromagnetic waves energy is carried by photons, which may be thought of as particles, which physically travel through space; thus, a medium is not needed for the signal to be transferred. Therefore, light can travel through a vacuum but sound cannot.

Sound waves have compressive and tensile phases

The explanation above describes the compressive phase of a sound wave, i.e. where the molecules are compressed. For the case where the object moves away from the fluid, there is a resulting rarefaction of the molecules, i.e. the moving object leaves a partial vacuum. In this case, the neighboring molecules will move to fill the void, leaving a new region of rarefraction. This continues one region to the next and the rarefractional disturbance propagates through the medium as a tensile acoustic wave. In most cases a tensile wave propagates just like a compressive wave and with the same sound speed.

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![Figure 49.2](image_url) (A) Focal waveforms measured in the Dornier HM3 at 24kV and the Storz SLX at Energy level 9. (B) Comparison at lower settings: the amplitudes are about the same but the SLX waveform has not formed a shock.

![Figure 49.3](image_url) Molecular view point of a sound wave. (A) Medium is at rest. (B) A piston pushes all the molecules out of the left side, resulting in a localized region of compression at the face of the piston (dark region). (C) The neighboring molecules are compressed and the compression region moves away from the piston. (D) The wave continues to move away from the piston while the molecules at the piston return to their initial condition.
sound speed is slightly changed by the presence of the wave.

The waveform shown in Figure 49.1 displays the pressure pulse as a function of time at a given point in space. This is typically how acoustic waves are measured, e.g. a microphone will record how pressure varies in time at one point in space. However, acoustic waves also vary in space and it is often useful to think of the wave in terms of its spatial extent. The relationship between the temporal separation of points on an acoustic wave (Δt) and the spatial separation of the points (Δx) is given by:

\[ \Delta x = \Delta t \, c_0 \]  

(Eq 49.1)

Recall that in water or tissue the \( c_0 \) is 1500 m/s (1.5 mm/μs) and therefore, for the shock wave shown in Figure 49.1, the positive part of the wave (a portion 1 μs long in time) will have a spatial extent in water of 1.5 mm. For a sinusoidal wave, the spatial extent of one cycle of the wave is called the wavelength.

**Sound waves are not just pressure waves**

When a sound wave propagates, it affects the density, pressure, and particle velocity of the fluid particles. The impact on the density occurs because as molecules are compressed together, the local density (\( ρ \)) will increase and in regions of rarefaction the density will decrease. For an acoustic wave it is convenient to write the total density as:

\[ ρ = ρ_0 + ρ_a \]  

(Eq 49.2)

where \( ρ_0 \) is the ambient density of the medium (in the absence of sound) and \( ρ_a \) is the variation in the density due to the acoustic wave.

The pressure in the fluid can similarly be written as the sum of two terms:

\[ p = p_0 + p_a \]  

(Eq 49.3)

where \( p_0 \) is the ambient pressure (in the absence of sound) and \( p_a \), the *acoustic pressure*, is the fluctuation due to the sound wave. For most fluids, acoustic pressure and density are directly related by an “equation of state” which takes the form:

\[ p_a = \rho_a c_0^2 \]  

(Eq 49.4)

That is, where the wave is compressed the pressure will be positive and where the fluid is rarefied the pressure will be negative. Physically, pressure represents a force per unit area and has units of Pascals (Pa). One Pascal is quite a small pressure and atmospheric pressure at sea level is approximately 100 000 Pa. In biomedical ultrasound, acoustic pressure is normally measured in megaPascals (MPa).

By way of example, the amplitude of the pressure from a diagnostic ultrasound scanner is about 2 MPa at the focus. Typically, values for \( ρ_0 \) and \( c_0 \) in tissue are 1000 kg/m³ and 1540 m/s, respectively, and so this corresponds to a relative density perturbation of \( ρ_a/ρ_0 = 0.0009 \). For lithotripsy, peak pressures can be upwards of 100 MPa, which results in \( ρ_a/ρ_0 = 0.04 \). Therefore, the density disturbances associated with acoustic waves in medical devices, even the very strong waves that are produced in lithotripsy, actually result in very weak (<5%) compression of the fluid.

**Progressive waves and particle velocity**

The case shown in Figure 49.3 where the compression wave moves in one direction is referred to as a *progressive wave*. In contrast, when there are sound waves traveling in different directions, this is referred to as a *compound wave*, which will not be considered here. For a progressive wave the molecules in the compressed region also have a small net velocity away from the source. The net velocity of the molecules in a region of space is referred to as the *particle velocity* \( (u) \) and for a progressive acoustic wave it can be expressed as:

\[ u_a = p_a / ρ_0 \, c_0 \]  

(Eq 49.5)

Using the example of a 100-MPa shock wave, the instantaneous particle velocity at the peak is about 67 m/s. It will be shown below that the particle velocity is needed in order to determine the energy in an acoustic wave.

If the velocity is integrated in time, the displacement of tissue can be calculated. For the 100-MPa shock wave described above, the tissue will move about 30 μm during the 1 μs compressive phase and then slowly return to its original position during the tensile phase. It has also been suggested that the particle velocity within a biologic target may produce sufficient strain to damage the cells. Below it will be discussed how the deformation can also build up over many pulses and in certain cases may damage tissue.

**Acoustic impedance**

The density and sound speed of a material (Eq 49.4) determine its *specific acoustic impedance* \( (Z_0 = ρ_0 \, c_0) \). This term is often shortened to *acoustic impedance* or just *impedance*. The impedance of tissue and water is about \( 1.5 \times 10^6 \) kg/m²/s. The units are often referred to as Rayls, after the eminent 19th century acoustician Lord Rayleigh, although the Rayl is not a standard SI unit.

Therefore, for a progressive acoustic wave, the pressure, density and particle velocity are not independent but are linearly related to each other.
where the coefficients are material properties. It follows that regions of high pressure are also compressed and have a high particle velocity (away from source), and regions of low pressure are rarefied and have a negative particle velocity (towards the source). As the acoustic wave travels, the fluctuations in density, pressure, and particle velocity all move together (i.e. “in phase”). Therefore, in a fluid with known material properties, if one property of an acoustic wave, such as the acoustic pressure, is measured, then Eq 49.5 can be used to determine the other acoustic properties.

### Wave intensity or energy

A propagating acoustic wave carries energy. The amount of acoustic energy per unit area is called the energy flux, energy density, energy flux density, or the pulse intensity integral. An IEC standard, which describes how pressure measurements should be taken on a lithotripter to ensure accurate results and fair comparisons across devices, defines the energy per unit area the “pulse intensity integral” (PII) [20]. It can be calculated from the following integral:

\[
\text{PII} = \int p_a u_a \, dt \tag{Eq 49.7}
\]

where the integration is done over the duration of the pulse. This is the acoustic equivalent to the expression from physics “work equals force times distance,” where acoustic pressure is the force per unit area and the time integral of the velocity gives the distance. The units for PII are joules per square meter (J/m²). For a progressive wave it is known that the particle velocity is related to the acoustic pressure \( p_a = \frac{p_a}{Z_0} \) and therefore:

\[
\text{PII} = \int \frac{p_a^2}{Z_0} \, dt \tag{Eq 49.8}
\]

in which case the pressure of the wave needs to be measured to determine PII. Note that to calculate the integral, the entire pressure versus time waveform needs to be accurately measured so that the integration can be done.

To determine the energy in an acoustic wave, a specific area \( A \) has to be chosen and the energy that passes through that area can then be calculated as:

\[
E = \iint \text{PII} dA \tag{Eq 49.9}
\]

where the double integral indicates a surface integral over the area \( A \). The units for energy are Joules (J). The energy \( E \) will depend on both the size of the area \( A \) and how the intensity varies across the area. The focal acoustic pulse energy is calculated using the area in the focal plane where the pressure is greater than half the maximum pressure (this is equivalent to the focal zone, see below). Various other conventions have been employed to calculate energy in SWL, e.g. the projected area of a stone or the area where the peak pressure is above 5 MPa [21].

Another acoustic property is the power per unit area or the intensity \( I \). Power is energy per unit time and so the intensity is the energy density divided by the time over which the integration was done (Eq 49.8), which is normally the pulse length \( T_p \):

\[
I = \frac{\text{PII}}{T_p} \tag{Eq 49.10}
\]

Intensity has units of watts per square meter (W/m²) but it is more common in biomedical ultrasound to use W/cm². For a sinusoidal pressure wave the integral (Eq 49.8) can be calculated exactly and the intensity is:

\[
I = \frac{\hat{p}^2}{2Z_0} \tag{Eq 49.11}
\]

where \( \hat{p} \) is the peak pressure of the sinusoidal wave. If the impedance for water or tissue \( Z_0 = 1.5 \) MRayls is substituted, the relationship can be expressed as \( \hat{p} I = 3 \) where \( \hat{p} \) is in atmospheres of pressure and \( I \) is in W/cm². For pulsed pressure waves, such as in lithotripsy, a simple expression does not exist for the intensity, as even small changes in the pulse shape can have a significant effect on the integration used to calculate PII.

### Reflection and transmission of sound waves

When an acoustic wave encounters a medium with a different impedance, then part of the wave will continue to propagate into the new medium (the transmitted wave) and part of the wave will be reflected back into the original medium (the reflected wave). In the case of normal incidence, where the propagation direction of the shock wave is perpendicular to the surface, the amplitude of the transmitted and reflected waves depends only on the change in impedance between the two media, what is referred to as the impedance mismatch. In terms of acoustic pressure the transmission and reflection coefficients are:

\[
R_p = \frac{Z_2 - Z_1}{Z_2 + Z_1} \tag{Eq 49.12}
\]

\[
T_p = \frac{2Z_2}{Z_2 + Z_1} \tag{Eq 49.13}
\]

There are a different set of coefficients for the intensity or energy, which are called the intensity transmission and reflection coefficients:
great care must be taken to eliminate air pockets between the shock head and the body [22, 23]. This is also one reason why stones are not targeted for treatment through lung or segments of gas-filled bowel. Indeed, the best acoustic window, which allows the shock wave a pure tissue path to the kidney, is on the flank of the patient (delineated by the ribs, spine, and pelvic bone).

**Focusing and diffraction of sound**

In lithotripsy, focusing of the shock waves is used to concentrate the acoustic energy on to the stone, while reducing the impact on the surrounding tissue as much as possible. Lithotripters achieve focusing by various means, including the use of reflectors, acoustic lenses, and spherically curved sources. Regardless of the method used, the physics that describes the focusing of the waves is similar for all these cases. An ideal focus would be the case where all energy is localized to an infinitesimally small region in space. However, the physics of wave propagation does not allow the energy to be focused to an arbitrarily small volume due to a process called diffraction. This means that even though the acoustic pressure may be greatest at one point in space, there is a finite region or volume of surrounding space that is also at high amplitude. This called the focal zone. For a theoretically optimal focusing arrangement, where sound can come in from all angles, diffraction puts a limit on the size of the focal zone of about one wavelength. For the realistic focusing arrangements used in lithotripsy, where the sound only comes from one direction, the focal zone can be from a few millimeters to tens of millimeters in size.

**Focal zone**

The focal zone of a lithotripter (equivalent terms include focal region, hot spot, focal spot, focal volume, zone of high pressure) of a lithotripter is normally ellipsoidal in shape with its longest dimension along the axis of the shock wave. To demonstrate this, Figure 49.5 shows the predicted peak pressure of the focal zone in an unmodified Dornier HM3 lithotripter [24]. The length and diameter

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**Equations**

\[ R_i = \frac{(Z_2 - Z_1)^2}{(Z_2 + Z_1)^2} = R_i^2 \]  
(Eq 49.14)

\[ T_i = \frac{4Z_1Z_2}{(Z_2 + Z_1)^2} = 1 - R_i \]  
(Eq 49.15)
of the focal zone depends on the diameter of the source, the focal length of the source, and the frequency content of the waveform. The dimension of the focal zone is, thus, one characteristic of any given lithotripter that is determined by design features and, as will be discussed below, focal zones can vary greatly between lithotripters.

For a focused acoustic source that generates a sinusoidal waveform, such as an ultrasound transducer, there are analytical expressions for the size of the focal zone. The critical parameters are the wavelength of the sound wave (\(\lambda\)) and the half angle of the aperture:

\[
\alpha = \arcsin(D/2F). \quad \text{(Eq 49.16)}
\]

where \(D\) is the diameter of the source and \(F\) the focal length. The formulae for the length (\(L_{FZ}\)) and the diameter (\(D_{FZ}\)) of the focal zone are:

\[
L_{FZ} = \frac{0.6\lambda}{\sin^2(\alpha/2)} \quad \text{(Eq 49.17)}
\]

\[
D_{FZ} = \frac{0.7\lambda}{\sin\alpha} \quad \text{(Eq 49.18)}
\]

Note that the focal length (\(F\)) is the distance from the mouth of the therapy head to the focus (where the stone should be placed). The focal length should not be confused with the length of the focal zone (\(L_{FZ}\)), which is the region around the focus where the pressure is high.

For a pulsed waveform, as is generated in lithotripsy, there are no explicit formulae for the size of the focal volume as the size depends on the waveform shape. However, the focal region of a lithotripter can be estimated using the formulae for the focal region of a sine wave. Figure 49.6A shows how the focal zone gets shorter and narrower as the diameter of the source aperture is increased. Figure 49.6B shows how the focal zone gets shorter and narrower as the focal length of the source (source–target distance) is decreased. Therefore, to make a small focal zone, a shock source with a large diameter aperture and short focal length would be desired. However, the size of the acoustic window in the flank and the need to be able to target stones deep in the body mean that for most lithotripters both the focal length and the diameter of the aperture are around 15 cm.

**Nonlinear acoustics**

When an acoustic wave has very large amplitude, e.g. a lithotripter shock wave, the speed of the wave is no longer constant but depends on the local compression of the fluid. For “weak” shock waves (recall even at the focus of the highest power lithotripter, the water is compressed by \(< 5\%\)), the speed of propagation (“phase speed”) of an acoustic wave is:

\[
c_{\text{phase}} = c_0 + \frac{\beta p}{\rho_0 c_0} \quad \text{(Eq 49.19)}
\]

where \(\beta\) is the coefficient of nonlinearity of the fluid and is a material property of the medium. For water \(\beta\) is about 3.5 and for tissue it varies from about 4 to 9. Normally, tissue of more complex structure has a greater coefficient of nonlinearity. A reasonable value for healthy soft tissue is 5.

Nonlinearity arises because of two physical processes; first, in regions of high pressure the local sound speed is increased above the usual value, and second the molecules in regions of high pressure have a higher particle velocity and are convected in the direction of acoustic propagation. For sound traveling through tissue, it is the first process that dominates the nonlinearity.

The difference between a nonlinear wave and a linear wave is that for a nonlinear wave, different parts of the wave travel at different speeds, as described by Eq 49.19. Figure 49.7 shows what happens to a sinusoidal wave as it propagates with nonlinearity present. The waveform becomes distorted in shape. In the absence of absorption, the wave obtains an infinite slope and then folds over and becomes multivalued. Ocean waves fanning up the beach are such waves but this waveform is not physically realizable in acoustics, i.e. it is not possible to have more than one pressure at any one point in space.

**Figure 49.6** Predicted focal zone size as a function of the diameter of the source and the focal length of the source for a 500 kHz source. (A) Contours show size of the focal zone for a source that has a focal length of 14 cm and an aperture of 25 cm (red), 15 cm (green), and 10 cm (blue). The focal zone gets longer and wider as the aperture size decreases. (B) Contours show the size of the focal zone for a source with fixed aperture diameter (15 cm) and varying focal length: 8 cm (red), 14 cm (green), and 20 cm (blue). The focal zone gets broader and longer as the focal length increases. For reference the Dornier HM3 hemi-ellipsoidal reflector has a focal length of 13 cm and an aperture diameter of 15 cm.
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Shock formation distance is so long that the wave has been absorbed before it can form a shock.

Rise time

In acoustics, waveforms are prevented from folding over (or breaking) by the presence of acoustic loss mechanisms. All acoustic waves will leave behind a small fraction of their energy as they propagate through a fluid; this loss of energy is referred to as absorption and will be discussed in the next section. The absorption of sound is greater from waveforms with steep gradients. In the case of a shock wave where the slope tends towards infinity, the absorption will also tend towards infinity. The shock will, therefore, never attain an infinite slope, but instead a balance between nonlinear distortion and absorption will result in a shock front where the pressure jumps in a very short time. This time is referred to as the rise time of the shock or the Taylor shock thickness. For a shock wave in water the expression for the Taylor shock thickness is:

\[ T_r = \frac{5}{\Delta P} \text{ ns} \cdot \text{MPa} \] (Eq 49.21)

where \( \Delta P \) is the pressure jump in MPa and the rise time (in ns) is defined as the time to go from 10\% to 90\% of \( \Delta P \). From this expression it is found that a 1-MPa shock should have a rise time of 5 ns and a 10-MPa shock a rise time of 0.5 ns. As a shock becomes stronger, the rise time shortens. Using Eq 49.1, the corresponding spatial extent of the rise time of the 1- and 10-MPa shock waves is 7.5 \( \mu \text{m} \) and 0.75 \( \mu \text{m} \), respectively.

Nonlinear acoustics phenomena are also important in other areas of biomedical ultrasound. In diagnostic ultrasound, nonlinear effects can create problems such as excess heating of tissue [25, 26], but can also be beneficial by enhancing image quality in tissue harmonic imaging [27–29]. Nonlinear effects are also important in high intensity focused ultrasound surgery (HIFU or FUS), where ultrasonic heating of tissue is exploited to destroy specific regions of tissue or to coagulate blood [30–33].

Absorption of sound by tissue

As mentioned above, when a sound wave passes through a medium, most of the energy remains in the sound wave but a small amount of it is absorbed by the medium. The amplitude of an acoustic wave will therefore slowly decay or attenuate as it propagates through a medium. The absorption in water is very low and, aside from controlling the rise time of the shock front, has little effect on lithotripter waveforms. The absorption in tissue, however, is about 1000 times larger than that in water and has a measurable effect on lithotripter waveforms.
The first reliable measurements of lithotripsy shock waves were performed with a hydrophone made of polyvinyl difluoride (PVDF), a piezoelectric plastic [34]. PVDF has very wide bandwidth, is capable of measuring high amplitude acoustic pressures, and can be manufactured so that only a small region is active. Both membrane hydrophones and needle hydrophones have been used, however membrane hydrophones are considered to yield the best measurement of the shock wave [35]. One problem with PVDF is that its adhesion with water is not strong and the tensile phase of the lithotripsy pulse can result in cavitation at the surface of the PVDF. This is a significant limitation that has two main consequences. First, it limits the ability of the hydrophone to measure the tensile phase of the shock wave as once the bubble forms the negative pressure is relieved and the hydrophone registers a pressure close to zero. Second, when the cavitation bubbles collapse, they can irreversibly damage the hydrophone.

The fiberoptic probe hydrophone (FOPH) [36] is considered to be the state of the art for measuring lithotripsy shock waves, and is the recommended device in the IEC standard [20]. The FOPH consists of a laser that injects light into one end of an optical fiber; the other end of the fiber is placed in the lithotripter field. The FOPH measures the light that is reflected from the end of the fiber and exploits the fact that the amplitude of the reflection depends on the pressure in the fluid. Several features make the FOPH superior to the PVDF membrane. Similar to PVDF, the FOPH has a wide bandwidth and is capable of measuring very high pressure amplitudes. The diameter of the active area of the FOPH (100 μm) is smaller than most PVDF hydrophones (500 μm). Also, the FOPH is made of an optical fiber (silica) and the adhesion between water and silica is very high. This means that cavitation is much less likely to occur at the surface of the FOPH and therefore it can more accurately capture the tensile phase of the shock wave. This also means that the FOPH is less susceptible to damage from cavitation. The main drawback with the FOPH is that the signal it generates is weak and therefore it is not good for measuring low pressures (~ 2 MPa and less).

Figure 49.8  Attenuation of sound as a function of frequency for muscle, fat, and kidney tissue (listed in decreasing order of loss). Also shown is the attenuation in water, which is much less (1000 times at 1 MHz) than the attenuation of tissue.

**How shock waves are measured**

**Hydrophones**

The main physical property of a lithotripter is the spatial and temporal distribution of its acoustic pressure field. The acoustic field is typically measured in water using a hydrophone, which converts pressure into an electrical signal. Lithotripters generate short (wide frequency band), high amplitude acoustic pulses, which are focused to a small volume in space. These physical parameters require that the hydrophone needs to be: (1) of a very wide bandwidth (60 kHz to in excess of 20 MHz); (2) robust enough to withstand the high pressures of the shock waves; and (3) possess a small active area (~ 0.5 mm).
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Typically, cavitation is initiated at micron size motes in the fluid or at sites of small gas pockets trapped on rough surfaces [45]. There are a number of different theories [46, 47] that can describe how acoustic cavitation proceeds once the cavity has been formed (at this point the cavity is normally referred to as a bubble). In Figure 49.10 the predicted radius is shown of a spherical bubble subjected to a lithotripter pulse (as in Figure 49.1). In this time scale the shock wave arrives at the focus at 250μs and the bubble is initially crushed by the leading compressive phase. The bubble grows to a millimeter at about 450μs. It then starts to collapse with a violent collapse occurring at 650μs. The time between the two collapse signals is the characteristic time (t_c). (B) Predicted acoustic emissions that the bubble calculated in A will radiate. There are two emissions due to the two collapses. (C) Measured acoustic emissions in a Dornier HM3 using a passive cavitation detection (PCD) (see Figure 49.11). The measured emissions agree with the calculations.

Acoustic cavitation

What is acoustic cavitation?

A second mechanical force generated by lithotripter shock waves is acoustic cavitation. This refers to the generation of cavities in a fluid (i.e. bubbles) when the tensile phase (negative pressure) of the acoustic wave is sufficiently strong to rip the fluid apart. In lithotripsy the tensile phase of the shock wave is large enough (~10MPa) to generate violent cavitation events. Cavitation is believed to play a significant role in tissue damage during SWL and to contribute to stone comminution [38–44].

Figure 49.9 (A) Waveform measured in water with a miniature polyvinyl difluoride (PVDF) hydrophone in a Dornier HM3 at 18kV. (B) Waveform measured in vivo in a pig for the same settings. The peak amplitude in vivo was about 30% less that in water, and the rise time in vivo (78ns) was much longer than that measured in water (26ns).

Basic shape to the in vitro waveform. The main difference is that the in vivo waveform has a 30% decrease in peak positive pressure and a greatly increased shock rise time (~70ns). Both of these effects are consistent with the higher attenuation associated with tissue and the heterogeneity of the tissue in the path to the kidney.

Figure 49.10 (A) Calculated radius versus time curve of a spherical bubble subjected to a lithotripter pulse (as in Figure 49.1). In this time scale the shock wave arrives at the focus at 250μs and the bubble is initially crushed by the leading compressive phase. The bubble grows to a millimeter at about 450μs. It then starts to collapse with a violent collapse occurring at 650μs. The time between the two collapse signals is the characteristic time (t_c). (B) Predicted acoustic emissions that the bubble calculated in A will radiate. There are two emissions due to the two collapses. (C) Measured acoustic emissions in a Dornier HM3 using a passive cavitation detection (PCD) (see Figure 49.11). The measured emissions agree with the calculations.
referred to as inertial cavitation, as the dynamics of the bubble are no longer driven by acoustics but instead by the inertia of the fluid surrounding the bubble. While the bubble is large, some amount of gas and vapor from the fluid will diffuse into the bubble. The bubble will then collapse by virtue of the near vacuum inside the bubble and the roughly 1 atm of ambient pressure in the surrounding fluid. It takes a further 150 μs for the bubble to collapse. The collapse is very violent and the gas that diffused inside is heated and compressed to such an extent that it can produce light [48]. The main collapse is followed by rebounds after which the gas that had diffused into the bubble will slowly diffuse back out into the fluid. Also shown in Figure 49.10 is the acoustic emission radiated by the bubble; a lithotripsy-induced cavitation bubble generates two acoustic emissions, one when it is hit by the compressive wave and one when it collapses hundreds of microseconds later. This unique “double-bang” signature can be used to detect the cavitation events [48].

**Measuring cavitation**

There are numerous techniques by which cavitation can be measured:

**High-speed photography**

Bubble behavior can be observed using a high-speed camera in an in vitro setting [49–52]. In principle this allows the entire dynamics of a bubble to be tracked from genesis to extinction, but in practice this is not feasible. During the growth phase the bubble needs to be imaged at millimeter length scales and tens of microsecond time scales. At the nadir of the collapse, the bubble radius is less than 1 μm and the dynamics of the collapse is at nanosecond time scales. The remnant bubble left after the rebounds is of the order of 10 μm and slowly dissolves over hundreds of milliseconds. Thus, the range of temporal and spatial scales makes it virtually impossible to capture all the bubble dynamics photographically. Therefore, investigators have found it necessary to study cavitation in segments. A further limitation of imaging is that cameras have a limited depth of field and cannot give an adequate record of bubble dynamics throughout the substantial volume of the cavitation field.

**Laser scattering of single bubbles**

The dynamics of a single spherical bubble can be measured very precisely by laser scattering [53]. In this case, a laser beam is used to illuminate a bubble and a photodetector is used to collect the light scattered by the bubble. For a spherical bubble, the amplitude of the scattered light varies in a known way as the bubble radius changes. This method is able to capture most of the temporal and spatial scales associated with the dynamics of a lithotripsy-excited cavitation bubble. However, laser scattering has several restrictions; the sample volume is very small, the method requires unrestricted visual access at high magnification, and the theory that is used to recover the actual bubble size is based on a single spherical bubble. This means the technique will only yield qualitative information about bubble clouds or nonspherical bubbles, both of which are very common in lithotripsy-induced cavitation.

**Acoustic detection in vivo**

Acoustic detection of bubbles is very powerful, in part because it can be used to characterize bubble dynamics within living subjects. Acoustic detection normally works in one of two modes: active cavitation detection (ACD) and passive cavitation detection (PCD) [54–56]. In ACD one transducer is used to send an acoustic wave toward the cavitation field, while a second transducer picks up sound reflections from the bubbles; this is the acoustic analog to laser scattering. In PCD, one or more receiving transducers listen for the “double-bang” acoustic emissions from cavitation bubbles (Figure 49.11). In the case where two receiving transducers (dual PCD) are used, it is possible to take advantage of coincidence detection to sample a small, discrete volume of the cavitation field where the transducers intersect [57]. The timing and amplitude of the two emissions is influenced by various factors such as the size of the initial bubble and the amplitude of the lithotripter pulse.
Thus, although acoustic detection does not image bubbles (i.e. it cannot provide information on bubble number and size), it gives valuable data that can be used to help characterize the acoustic output of a lithotripter and assess the environment and dynamics of the cavitation field [51, 57, 58].

Other techniques have also been developed for measuring cavitation. It has been observed that cavitation leads to pitting on metal foils and the number and depth of pits can be used to assess the violence of cavitation [38, 59, 60]. An electromagnetic probe device has been used to measure the mechanical force exerted on a steel ball by both the incident shock wave and the cavitation activity [61]. The high pressures and temperatures in the interior of the bubble provide an environment which can produce light emissions (sonoluminescence) and also result in enhanced chemical reaction rates (sonochemistry). Production of light and by-products from chemical reactions have both been used to quantify cavitation activity [53, 58]. These are secondary measurements of the cavitation field and interpreting the results in terms of physical processes can be complicated.

**Physics of clinical lithotripsy**

**Shock generation and focusing**

Three shock-wave generating principles have been used in clinical lithotripters.

**Electrohydraulic lithotripters**

The electrohydraulic lithotripter (EHL) has a spark source which generates a shock wave that is focused by an ellipsoidal reflector (Figure 49.12). In an EHL the pressure pulse originates as a shock wave and remains a shock wave at all times during its propagation from the spark source to the reflector, and then as it focuses to the target. As will be shown below, this is not the case for other types of shock-wave sources. In an EHL, focusing of the shock wave is critically dependent on the placement of the spark at the first focus of the ellipse. Misalignment by just a few millimeters can lead to a significant loss in focusing, and a lengthening and broadening of the focal zone. Thus, EHLs are designed so that the alignment of the electrode is consistent. Still, there is variability in the precise location of the spark discharge across the spark gap that is not easy to control. Therefore, from shot-to-shot there can be significant variation (upwards of 50%) in the amplitude of the shock wave and there can be some shift in the position of the focal zone at the target. A unique “feature” of EHLs is that the target is insonified by two pulses. The main focused pulse is preceded by the so-called “direct wave,” which travels directly from the spark to the target without bouncing off the reflector. The direct wave arrives about 30μs earlier and, because it undergoes spherical spreading, it is low in amplitude. However, it has been shown that this direct wave can influence the cavitation generated by the focused wave [62].

In EHL the electrodes wear out and must be replaced. Some lithotripter manufacturers have found ways to enhance the lifetime of their electrodes, such as by encapsulating them and filling the casing with an appropriate electrolyte [63]. Still, electrodes eventually show wear and this can affect their acoustic output.

**Electromagnetic lithotripters**

The electromagnetic lithotripter (EML) uses an electrical coil in close proximity to a metal plate as an acoustic source. When the coil is excited by a short electrical pulse, the plate experiences a repulsive force and this is used to generate an acoustic wave. If the metal plate is flat, the resulting acoustic wave is a plane wave that must then be focused by an acoustic lens (Figure 49.13A). If the plate is in the shape of a tube, the resulting cylindrical wave can be focused by a parabolic reflector (Figure 49.13B). In both cases, focusing is very reproducible and variation in measured pressure waves is less than 10%. Thus, the shock waves generated by EMLs are
applied to a piezoelectric crystal, it deforms and creates an acoustic wave. The crystals are placed on the inside of a spherical cap and the acoustic wave focuses at the centre of curvature of the sphere (Figure 49.14A). This focus is highly reproducible and very small variations in the focal waveforms are reported. Similar to the EML, the acoustic waveform in PEL starts as an acoustic pulse and a shock wave is created by nonlinear propagation distortion. For most clinical settings a shock is produced before the wave reaches the focus. Figure 49.14B shows a representative waveform measured at the focus of a piezoelectric lithotripter. Note the long ringdown for time greater than 3μs.

inherently more consistent than in EHLs. An additional advantage is that there are no electrodes to replace.

One difference between the acoustics of an EML and an EHL is that the acoustic pulse generated by an EML does not start as a shock wave; the displacement of the plate generates a high-intensity ultrasonic wave, which has a smooth waveform with no discontinuities. The amplitude of the wave at clinically relevant power settings is normally high enough that nonlinear distortion occurs during propagation, and a shock is produced before the wave reaches the focus. One exception is a broad focal zone EML which does not produce a shock at the focus [64]. A second difference is that the EML waveforms have a relatively small trailing positive pressure after the negative phase (this can be observed at about 6μs in Figure 49.2). This peak likely has little impact on the stress inside the stone but may affect the cavitation dynamics.

Piezoelectric lithotripter

The piezoelectric lithotripter (PEL) uses piezoelectric crystals to form an ultrasonic wave. When a voltage is applied to a piezoelectric crystal, it deforms and creates an acoustic wave. The crystals are placed on the inside of a spherical cap and the acoustic wave focuses at the centre of curvature of the sphere (Figure 49.14A). This focus is highly reproducible and very small variations in the focal waveforms are reported. Similar to the EML, the acoustic waveform in PEL starts as an acoustic pulse and a shock wave is created by nonlinear propagation distortion. For most clinical settings a shock is produced before the wave reaches the focus. Figure 49.14B shows a representative waveform from a PEL [65]. One significant difference from an EHL or EML waveform is the presence of a tail or coda at the end of the pulse. This is because the piezoelectric crystals “ring” for a couple of cycles after they are excited, a phenomenon not present in an EHL or EML. The coda at the end of the PEL pulse probably has a minimal effect on the stress induced in a kidney stone but may affect the cavitation dynamics.

Coupling of the shock source to the body

Efficient transfer of acoustic energy from one medium to another only occurs when the acoustic impedances
are very close. A water–tissue interface results in very good coupling and theoretically it should be possible to transfer more than 99% of the energy of the shock wave into the body. However, the presence of even a small pocket of air at the skin surface will result in a dramatic reduction in energy transfer to the patient (see Figure 49.4). Thus, the manner in which the shock wave is coupled to the body is critical.

**Water-bath lithotripters**

The “first-generation” lithotripters (e.g. Dornier HM3) were EHLs and used an open water bath in which the patient was immersed. Thus, there is nothing but water between the shock source and the patient. This is ideal except that bubbles that drift up from the spark gap or the cavitation bubbles that form along the path of the shock wave, have the potential to collect against the skin of the patient and interfere with the propagation of subsequent shock waves. To help prevent this, the ellipsoidal reflector of the shock source is fixed off vertical (in the Dornier HM3 the angle is 14°) and the water in the bath is continuously degassed.

**Dry lithotripters**

Most current lithotripters have the shock-wave source mounted in a “therapy head” which is filled with water. The therapy head is capped by a thin rubber membrane, which is pressed against the patient and through which the shock wave passes. The water in the therapy head of most lithotripters is continuously recirculated and degassed to remove any bubbles that might interfere with the shock-wave propagation. Although this design is more convenient in the clinic than a water-bath-type lithotripter, it is inherently less effective at allowing shock waves to pass because the presence of the rubber (although well matched to water and tissue) adds additional reflecting interfaces. A coupling agent such as gel or oil is used to marry the rubber membrane of the treatment head to the skin [66]. Although seemingly simple, this procedure may spell success or failure for treatment. Laboratory studies have shown the coupling interface to be prone to developing voids and such defects covering just 10% of the interface can reduce the breakage efficiency of model stones by around 60% [23]. The typical methods used to apply gel, such as dispensing from a squeeze bottle and rubbing by hand across the skin and treatment head, create defects and are highly variable. An improved method for coupling has been described [22].

**Focal zone of the lithotripter**

In lithotripsy, acoustic energy is focused to a relatively small focal zone surrounding the focal point of the lithotripter. The *focal point* is a geometric point in space, e.g. in an EHL this point is the second focal point (F2) of the ellipsoidal reflector [24], and it is usually the location at which the stone is placed for treatment. All lithotripters have a focal point, but lithotripters differ in the dimensions of the zone of high pressure (*focal zone*) that surrounds this point. The dimensions and the pressure characteristics of the focal zone are the most important features that distinguish one lithotripter from another.

There are many definitions of the focal zone that may be appropriate for lithotripsy. The IEC standard for measuring lithotripter pulses [20] defines it as the volume within which the measured peak acoustic pressure is at least half the maximum peak positive pressure. The peak positive pressure (p+) of a waveform (see Figure 49.1) is the highest positive pressure in that waveform. The maximum peak positive pressure is the highest value of p+ in the field of the lithotripter and the location of the maximum peak positive pressure is defined as the focus [20]. The maximum peak pressure will vary with the power setting of the machine. The resulting focal zone is normally an elongated, elliptical, “cigar-shaped” volume. It is worth noting that maximum peak pressure does not necessarily occur at the same location as where the manufacturer indicates a stone should be placed, and that the location of the focus and the dimensions of the focal zone may change as the power setting is changed.

The Dornier HM3 has arguably been studied and characterized more extensively than any other lithotripter. As such, data for the Dornier HM3 are a useful standard for reference. Because different lithotripters, even the same type of lithotripter, may perform somewhat differently, and because investigators have used different means to map the acoustic field of their lithotripters, published values for peak pressures and dimensions of the focal zone of a given type of lithotripter may not coincide perfectly. Representative focal zones of selected lithotripters are shown in Figure 49.15. Typical published values for the Dornier HM3 EHL report the maximum peak positive pressure to be 40 MPa at 20 kV and the focal zone to be about 60 mm long by 12 mm in diameter. In contrast the Storz Modulith EML has a maximum peak positive pressure around 100 MPa at energy level 8 and a focal zone that is about 35 mm long and only 4 mm in diameter. Reported values for PELs indicate a maximum peak positive pressure of 80 MPa and a focal zone 20 mm long and 3 mm in diameter [67, 68]. Thus, there is a considerable difference in the dimensions of the focal zone between lithotripters and those lithotripters with the narrowest focal zone have the highest peak pressures.

The half-maximum focal zone (also known as $–6\text{dB}$ focal zone as the contour corresponds to the pressure being $6\text{dB}$ less than at the maximum) is recommended...
Figure 49.15 Comparison of the focal zones of selected clinical lithotripters showing their dimension along the axis of the lithotripter (ellipses) and in the focal plane at the focus (circles) (image courtesy of P. Blomgren).

in the IEC standard, but this may not necessarily be the best descriptor of the focal zone of a lithotripter. For example, in a Storz lithotripter, with a peak pressure of 110 MPa at energy level 9, the focal zone will correspond to the region where the peak pressure exceeds 55 MPa. For a Dornier HM3, which only has a peak pressure of 40 MPa, the focal zone will correspond to a region where the pressure exceeds 20 MPa. Therefore, when comparing the focal zones of these machines the absolute pressure levels are very different; indeed, the focal zone of a Dornier HM3 would be zero if the 55 MPa level of the Storz focal zone were used. Other suggestions for the focal zone include: half the peak negative pressure, half the energy density, the surface where the peak pressure is 5 MPa, or even using the energy that passes through a volume with a diameter of 10 mm (about the size of a typical stone). Until there is a better understanding of how shock waves fragment stones, it is unlikely that an alternative metric will be agreed upon within the literature.

The smaller, tighter focal spot of an EML or PEL would at first glance appear to be advantageous as it should allow for more accurate targeting on the stone, and thus, less damage to the surrounding tissue. However, in vitro experiments (where stones are stationary) indicate that the EMLs or PELs, with their very high pressures, are no better at breaking stones that an EHL and often are not as effective [69, 70]. High peak positive pressure does not appear to correlate with enhanced stone fragmentation in the clinic [4, 15, 16].

Further, stone motion due to respiration means that with a tight focal zone fewer shock waves actually hit the stone and more shock-wave energy is deposited directly into tissue [71]. When it is considered that some tight focal zone lithotripters have peak pressures in excess of 100 MPa, this suggests that tissue is being subjected to a very high dose of acoustic energy. This may help explain the increased incidence of adverse effects such as subcapsular hematomas observed with these machines [6, 13].

Device equivalency/equating lithotripter performance

At present there are no agreed metrics by which the acoustic output of different lithotripters can be compared, and there is no straightforward means to operate a given lithotripter so that it is equivalent to another. This is partly due to the fact that, although all lithotripters produce shock waves that have similar waveforms, the amplitude and focal zone of different lithotripters is not the same, and measurements of the properties of the acoustic field can yield very different values. This is illustrated in Table 49.1, where a number of physical measurements made on an EHL and an EML are shown [72]. For the settings chosen, the only parameter that was roughly equivalent was the energy incident on a 6.5-mm diameter stone (0.484 mJ vs 0.528 mJ). However, other physical measurements varied tremendously, e.g. the peak positive pressure in the EML was three times that of the EHL.

Therefore, although it is possible to find settings on two given machines that give equivalency on one physical property, it is unlikely that there will be equivalency on other properties and, indeed, there is likely to be significant differences. For example, if the power setting
Mechanisms of shock-wave action

Acoustic waves in stones

A number of mechanisms have been proposed by which lithotripsy shock waves may destroy kidney stones. The acoustic field in stones is more complex than the acoustic theory described for shock waves in tissue. Kidney stones are elastic solids and support two types of waves: a longitudinal or compression wave (which is akin to an acoustic wave) and transverse or shear waves, where the motion of the vibration is transverse to the direction of propagation. In a shear wave, the transverse vibration does not result in the molecules being compressed and rarefied, but rather they oscillate in a manner analogous to the wave motion of a rope excited by a snap of the wrist. Longitudinal waves and shear waves travel at different speed and the longitudinal wave speed ($c_L$) is always faster than the transverse wave speed ($c_T$).

When a shock wave passes from urine or tissue into a stone, the transmitted energy is divided between the longitudinal and transverse waves in the stone. How the shock-wave energy is divided between the longitudinal and transverse waves depends on the material properties of the stone and the angle of incidence. If the wave is normally incident on the stone surface, then all the energy is converted into a longitudinal wave in the stone and no energy is available for transverse waves. As the angle of incidence increases, less energy is converted into a longitudinal wave and more is converted into transverse waves. The complex shape of many natural stones results in a nontrivial partition of energy between the two types of wave.

The basic features of the interaction of shock waves with a stone can be illustrated by means of a computer simulation. The computer simulation solves the equations of motion for particles in an elastic solid [17]. Figure 49.16 shows a series of snap-shots of the interaction of a lithotripter shock wave with a cylinder-shaped stone. The snap-shots show the distribution of the maximum tensile stress inside the stone at each instant of time. In the first two frames, the shock wave can be seen to enter the stone as compressional waves. The third and fourth frames show that the longitudinal wave inside the stone and the acoustic wave outside the stone result in the generation of shear waves from the lateral walls. Further, between the third and fourth frames, the shock wave reflects from the rear wall. Because the impedance of the surrounding fluid is less than that of the stone, the reflected pressure wave is inverted (because the pressure reflection coefficient $R_p$ given in Eq 49.19 is negative if $Z_2 < Z_1$) and the leading compressive wave is reflected as a tensile wave. The last two frames show constructive interference between the shear and longitudinal waves to produce the high tensile stresses in the stone.

Acoustic properties of stones

From the point of view of determining the stress inside a kidney stone, the important material properties are: density ($\rho_0$), longitudinal sound speed ($c_L$), and shear wave velocity ($c_T$). Figure 49.17 shows reported measurements from human stones [74–77]. There is a large variation in the reported properties for uric acid, calcium oxalate monohydrate, and cystine stones, e.g. the sound speed in calcium oxalate monohydrate varies between 3000 m/s and 4500 m/s. This variation is likely due to

<table>
<thead>
<tr>
<th>Length$_{z2}$</th>
<th>Width$_{z2}$</th>
<th>$p_+$</th>
<th>$p_-$</th>
<th>$E_{z2}$</th>
<th>$E_{STONE}$</th>
<th>$t_c$</th>
</tr>
</thead>
<tbody>
<tr>
<td>EHL</td>
<td>54 mm</td>
<td>9 mm</td>
<td>37.5 MPa</td>
<td>−7.8 MPa</td>
<td>4.25 mJ</td>
<td>0.484 mJ</td>
</tr>
<tr>
<td>EML</td>
<td>32 mm</td>
<td>3.5 mm</td>
<td>115 MPa</td>
<td>−14.6 MPa</td>
<td>3.35 mJ</td>
<td>0.528 mJ</td>
</tr>
<tr>
<td>Ratio</td>
<td>1.69</td>
<td>2.57</td>
<td>0.33</td>
<td>0.53</td>
<td>1.27</td>
<td>0.92</td>
</tr>
</tbody>
</table>
Figure 49.16 Snap-shots of the tensile stress generated by the propagation of a lithotripsy shock wave through a model kidney stone surrounded by fluid. The cylindrical stone (6.5 mm wide × 7.5 mm high) has the following properties $\rho = 1700 \text{ kg/m}^3$, $c_L = 3000 \text{ m/s}$, $c_S = 1500 \text{ m/s}$. The shock wave is incident from below and the color scale depicts the tensile stress (in MPa), where yellow through red indicate regions of tensile stress, and blue regions of compression. (A) (at 3.6 μs) The leading compressional phase of the shock wave in the fluid (L1) is almost incident on the proximal surface of the stone. (B) (4.6 μs) The shock wave has entered the stone as a longitudinal wave (L2). Note because the propagation axis is normal to the surface, no shear waves are generated at this interface. Because the speed in the stone is higher than in the fluid, the wave in the stone advances ahead of the wave in the fluid (L3). The reflection of the shock wave by the proximal surface can be seen leaving the bottom of the image (L4). (C) (6.0 μs) The tensile tail of the incident shock wave can be seen in the stone (L5) following the leading compressive phase. The interaction of the longitudinal wave in the stone with the lateral walls of the stone results in the production of shear waves (S1) that propagate towards the axis of the stone. (D) (6.8 μs) The leading compressive phase has been partially transmitted (L6) and reflected (L7) at the distal surface. The reflection coefficient is approximately $-0.5$ and results in a tensile phase (L7) that generates significant tensile stress (red region) near the distal surface; this is spall. The wave on the outside of the stone (L8) is inducing further shear waves (S2) inside the stone. (E) (7.6 μs) The reflected longitudinal wave and the shear waves interact to produce a large region of tensile stress in the center of the stone (L/S). (F) (9.2 μs) The shear waves interact near the distal surface to generate another region of high tensile stress (S3).
Spall fracture

Spallation occurs after the shock wave enters the stone and subsequently reflects from the rear of the stone (see Figure 49.16). The stone–urine interface inverts the large positive pressure pulse, resulting in a large tensile stress. This stress is added to the tensile stress of the, still incoming, negative pressure tail, resulting in a very large tensile stress near the back wall [78–81]. Most solids are much weaker in tension than in compression, so the large tensile stress near the rear of the stone can be expected to make the material fail. Recent fundamental research suggests that spall plays a small role in the fragmentation of kidney stones [17, 19].

Squeezing

Squeezing occurs because of the difference in sound speed between the stone (>2500 m/s) and the surrounding fluid (~1500 m/s). The shock wave inside the stone “runs away” from the shock wave propagating through the fluid outside of the stone (see Figure 49.16B,C). The shock wave that propagates in the fluid outside the stone results in a circumferential force on the stone (known as a hoop stress). It has been proposed that this will result in a tensile stress at the proximal and distal ends of the stone and lead to an axial “splitting” failure [82]. More recent dynamic studies suggest that squeezing produces shear waves that then produce high tensile stresses internal to the stone towards the distal surface [17]. It is these regions of high tension associated with shear waves that appear to correlate best with in vitro stone fragmentation [19].

Shear stress

The shear waves and compressive waves that propagate inside a kidney stone (see Figure 49.16) will generate regions of high shear stress in the stone. Figure 49.19 shows in vitro experiments in which regions of high...
shear in a stone can be visualized. Many materials are weak in shear, particularly those that have layers. In the case of kidney stones that consist of layers, the bonding strength of the matrix between the layers often has a low ultimate shear stress [78, 79, 83, 84]. Further, the organic binder of kidney stones is much softer than the crystalline phase, and as the shock front passes through the stone, it will induce very large shear stresses at the binder–crystal interfaces which likely contribute to the fracture of the kidney stone [85].

Superfocusing

Superfocusing is the amplification of stresses inside the stone due to the geometry of the stone. The shock wave that is reflected at the distal surface of the stone can be focused either by refraction (associated with the high sound speed and geometry of the stone) or by diffraction from the corners of the stone. It has been shown that these reflected waves can be focused to caustics (regions of high stress) in the interior of the stone and that this can lead to failure [84, 86]. The regions of high stress (both tensile and shear) can be determined from the geometry of the stone and its elastic properties, e.g. density, longitudinal wave speed, and shear wave speed.

Cavitation

Cavitation refers to small bubbles/cavities that grow in the urine surrounding the stone in response to the large negative pressure tail of the acoustic pulse. When a cavitation bubble collapses near a solid surface (e.g. a kidney stone), a microjet of fluid is formed that pierces the bubble and impacts the surface (Figure 49.20) with speeds upwards of 100 m/s [39]. This jet likely plays a role in cavitation-induced damage to kidney stones [39–40]. The collapse of the cavitation bubble also results in the emissions of secondary shock waves [87, 88] that are radiated into the stone. These secondary shock waves have an amplitude comparable to that of the focused shock wave [89].

In vitro experiments where cavitation is suppressed show significant reduction in stone fragmentation [59, 90, 91]. Cavitation is principally a surface-acting mechanism, and experiments indicate that it acts most strongly on the proximal (shock-wave incident) surface of the stone [40, 80, 92]. Studies suggest that cavitation is
Section 4 Shock-Wave Lithotripsy

Section 4 Shock-Wave Lithotripsy

There are two pieces of evidence that strongly support the argument that stone comminution is a fatigue process. First, the internal structure of stones has been shown to affect how they fragment in lithotripsy [100–103]. Second, normally more than 1000 shock waves are required to progressively fragment stones into sufficiently small pieces; the use of multiple stress cycles to fracture a material is a classic hallmark of fatigue [97, 98].

Although, present understanding of SWL indicates that the stones fail through a fatigue process, it is not clear which mechanism drives the fatigue. The two most commonly cited mechanisms are direct stresses (tension and shear) and cavitation, or some combination of them [93]. Part of the problem in determining which mechanism is in action is that only limited data on the material strength of kidney stones has been reported, e.g. ultimate strength in compression, fracture toughness, Knoop hardness, and Vickers micro-hardness [74–76, 104–107]. Of note is the paucity of data for the tensile and shear strength of kidney stones. This is most likely because determining these properties in brittle materials is fraught with technical difficulties. Further, most of the data have been measured in quasi-static tests, with the stress applied over many minutes, and the results may not be representative of the material properties when subject to shock waves, where the stress is applied and removed in microseconds [85]. At present the data on material strength of kidney stones are insufficient for the fracture process to be described.

Mechanisms of tissue damage

It is now well recognized that SWL results in trauma to the kidney, and that in some cases the injury can be severe [9]. The clinical implications of such adverse effects are still under investigation and a thorough discussion of the complications associated with lithotripsy is given in Chapter 53.

The notion that lithotripter shock waves can pass harmlessly through the body is simply not true. It is likely that all patients who receive at least an average dose of shock waves (2000 shock waves at mid-range power or higher) experience some degree of tissue trauma. Lithotripsy has been very beneficial for a large number of patients, but it has also led to severe, even catastrophic adverse effects for others [9, 11, 108]. To better understand how shock waves have the potential to cause tissue trauma, the physics of the problem needs to be considered.

As discussed above, lithotripters produce a focused acoustic pulse. The acoustic field is broad at the source and narrow at the focus. The focal zone, the area of highest acoustic pressure, is elongated and of dimensions that cannot be localized exclusively to a stone.
Figure 49.21 Cavitation bubble cluster collapse on a stone 6.5 mm in diameter × 7.5-mm long. (A) Orientation of shock wave to stone. (B) 100 μs after shock-wave arrival a bubble cluster has formed on the proximal surface and a few bubbles have formed in the surrounding fluid. (C–E) (200, 300 and 400 μs after shock-wave arrival, respectively) the cluster continues to grow. (F) The cluster begins to collapse in a mushroom-like shape. (G) Final collapse of the cluster at the centre of the stone (reproduced from Pishchalnikov et al. [52] with permission).

Although shock waves are targeted onto the stone, the surrounding tissue is also subject to significant mechanical forces. The length of the focal zone of most lithotripters is about 50 mm (see Figure 49.15) and this means that the entire thickness of the kidney is subject to high amplitude shock waves. In addition, patient motion, due to respiration or discomfort, likely results in the stone spending a good portion of the treatment time out of the focal region and, thus, many of the shock waves will interact solely with tissue.

Fortunately, tissue has physical properties that make it far less susceptible to damage by shock waves than kidney stones. For example, the fact that the acoustic impedance of tissue is close to that of water means that shock waves can pass through a tissue–water interface without significant reflection. Thus, tissue is not subjected to the extreme tensile forces that cause stones to fail. Further, the sound speed in tissue is almost constant and so tissue will not be subject to squeezing, which can generate tensile and shear stresses. However, tissue is subject to deformation by the pressure wave and to cavitation induced by the tensile phase of the shock wave. We briefly describe the mechanisms that may contribute to tissue injury.
Mechanical stress

The positive pressure of a lithotripter pulse leads to significant compression of tissue. The fact that the shock rise time in vivo is of the order of 70ns means that the spatial extent of the shock front is about 100μm. Therefore, tissue structures in the range of 10μm to 1mm will experience a significant variation in stress across them as the shock wave passes. The short rise time associated with the shock will lead to nonuniform straining of the tissue, resulting in shear forces. It is generally recognized that tissue structures are sensitive to shear stress and the distortion of the tissue by the shock wave could induce enough shear to cause damage [109, 110].

Cumulative shear

As mentioned above, the positive phase of the shock wave can induce a displacement in tissue of tens of microns. The tissue returns to its original state during the negative phase. However, there can be a small net displacement remaining in the tissue after the passage of the shock wave. Eventually the elasticity of the tissue will relax the tissue back to its state. However, for kidney tissue it has been estimated that this time scale may be of the order of 1 s [111]. This means that for shock waves that are delivered at rates faster than 1 s, i.e. faster than 60 shock waves/min, the tissue will not have time to return to its rest state before the next shock wave is delivered. Over thousands of shock waves the net displacement could build up to sufficient levels to result in tearing of the tissue: a process referred to as cumulative shear [111].

Shear induced by inhomogeneities

Tissue is an inhomogeneous medium at multiple length scales. Spatial variation in the sound speed on the millimeter length scale can have a dramatic effect on the focusing of ultrasonic pulses in tissue [112]. As the shock wave focuses, parts of the wavefront that passed through tissue with high sound speed will be advanced and the parts that passed though low sound speed tissue will fall back. This distortion in the wavefront will lead directly to shear stresses in the tissue. Again, these shear stresses could be strong enough to induce mechanical damage of the tissue [109].

Cavitation

Cavitation is known to occur in tissue during lithotripsy [51, 113–115]. Measurements using passive cavitation detection in both humans and pigs have detected the unique acoustic signature associated with cavitation. Measurements have indicated the presence of cavitation in the perirenal fat, collecting system, parenchyma, and subcapsular hematomas.

Cavitation has been well documented to have a significant biologic effect in many in vitro settings [116–121]. Experiments in lithotripsy indicate that damage to in vitro cells and in vivo tissue is dramatically reduced when cavitation is reduced or eliminated [42, 121, 122]. This makes cavitation the most likely candidate for tissue damage in tissue. The onset of detectable cavitation in the parenchyma requires about 1000 shock waves to be delivered using a Dornier HM3 in the pig model [115]. This is consistent with the approximately 1000 shock-wave dose needed to observe widespread damage in the pig kidney for the same shock-wave settings [9]. These data suggest that in SWL widespread damage to the kidney can be attributed to cavitation.

For cavitation bubbles to undergo a violent growth and collapse cycle, the bubble needs to be in a fluid environment. This implies that cavitation activity, at least initially, is more likely to occur in the fluid environment of blood vessels than within the surrounding tissue. There are at least two mechanisms by which bubbles could produce mechanical damage to vessel walls and once a vessel is ruptured it is likely that cavitation activity progresses quickly.

Bubble expansion Bubbles may rupture vessel walls during the expansion phase of the bubble cycle. That is, as the negative pressure of the shock wave passes through the vessel it causes the bubble to undergo explosive growth (see Figure 49.10) pushing outward on the vessel and rupturing it. Experiments using cellulose tubes as capillary phantoms in an in vitro setting support the explosive bubble hypothesis [123, 124]. However, measurements of bubble activity in isolated mesentery vessels suggest that bubble growth results in little distention of vessel walls and that bubble collapse is more likely to result in injury for thin-walled vessels particularly, such as venules [125].

Bubble collapse When cavitation bubbles collapse near rigid surfaces they can form high-velocity microjets that are forceful enough to etch metal surfaces (Figure 49.20). These jets would also appear to be capable of puncturing the wall of a capillary or other blood vessel; however, bubble dynamics near elastically “soft” boundaries do not result in jetting towards the surface. High-speed movies of bubble activity in mesentery vessels found no evidence of jetting towards the wall [125]. Rather the dominant effect of the collapsing bubbles was to produce a focal invagination of the order of 10μm, with deformation that has the potential to induce vessel rupture.
**Cavitation progression** Once blood vessels have been ruptured, either by mechanical stresses or cavitation bubbles inside intact vessel, blood will collect in pools, e.g. in a hematoma, and there is a greater potential for cavitation to occur. The pooling of blood provides a large fluid-filled space for cavitation bubbles to grow and collapse. Also, existing bubbles, which can act as nuclei for subsequent cavitation events, will not be swept away by blood flow, but will remain in the pooled region. This explains the intense PCD cavitation signals and B-scan ultrasound echogenicity collected from hematomas during SWL [113, 114, 126]. The ensuing violent cavitation could result in liquefaction of the tissue and the spread of a wave of cavitation-induced damage through the kidney [127].

Research continues in this area with the goal of confirming whether the physical processes outlined here, or some other processes, are responsible for tissue damage in SWL. This is partly due to the fact that the mechanical response of the tissue, at least at strain rates relevant in SWL, is not well understood, and so damage criteria are also not well defined. Further, there are few experimental systems that can be used to test and validate different hypotheses. Although the general consensus among researchers is that cavitation is the primary mechanism for tissue injury, this field still requires much study.

**Evolution of the lithotripter**

It is more than 30 years since the introduction of lithotripsy to clinical practice and there have been a number of noteworthy changes in equipment design, but none that has involved a fundamental change in the acoustics of the lithotripter. That is, lithotripters have changed (they are now compact, modular, use dry shock heads, have improved imaging), but the acoustic signature of the lithotripter pressure pulse remains the same. The focal waveform generated by a Dornier HM3 is virtually the same as the waveform produced by any of the numerous lithotripters available on the market today. This is not to imply that the lithotripter industry has been static. Indeed, there has been a very active effort on the part of manufacturers to produce machines that are easier and more practical to use. In this regard lithotripsy has seen numerous refinements. At the same time, however, it is essential to note that success rates in lithotripsy have declined [4, 15–16]. One refinement was to develop lithotripters that produced a tight focal zone of high peak positive pressure. However, as discussed above, data suggest that this led to decreased efficiency of stone breakage and to an increase in collateral damage. More recently, manufacturers have developed lithotripters with broader focal zones [128–130] or in the case of the Storz SLX-F2, a user-selectable focal zone. This is an example of progress in lithotripsy which comes at the expense of poorer outcomes and it appears that a step back is now being taken.

The first lithotripters were electrohydraulic devices in which the shock wave was generated by underwater spark discharge and shock-wave coupling was achieved by immersion of the patient in a water bath. The Dornier HM3 was a very popular lithotripter of this era, and at some centers is still in use today. By today’s standards the Dornier HM3 produces moderate peak positive pressures (∼40 MPa) delivered to a generous focal zone (∼12 × 60 mm). The Dornier HM3 was a very successful machine and it is probably safe to say that the early success and rapid acceptance of lithotripsy was built on the back of this particular lithotripter.

Even though the Dornier HM3 was a very effective lithotripter, it was perceived by some to have several significant drawbacks. It used an open water bath to couple shock waves to the body; treatment was painful, necessitating that the patient be sedated or even anesthetized; the shock wave firing had to be gated to the cardiac cycle (to reduce risk of arrhythmias), which slowed the treatment; and the lithotripter was a large, stationary piece of equipment that required a dedicated water treatment plant. What physicians (and patients) really wanted was lithotripsy that was painless, fast, and convenient with minimal to no anesthesia; a fully ambulatory walk-in-walk-out therapy. Lithotripter manufacturers responded with a number of modifications. Problems related to the overall physical design of the lithotripter were challenging but solvable. For example, the issue of the open water bath was addressed by enclosing the shock head and by using a rubber membrane to couple the shock wave to the body. This was not a perfect solution, as there is no better way to achieve acoustic coupling than through a water–tissue interface [23]. However, elimination of the water bath meant that medical staff had much easier access to the patient, the lithotripter did not necessarily have to be tied down to a dedicated facility, and lithotripters could be designed as modular systems. Many modern lithotripters have been designed to be portable and are used in mobile lithotripsy units.

The cardiac gating was found to be unnecessary for most patients and so most lithotripters run in “ungated” mode in which the urologist can chose the rate at which shock waves are delivered. Commonly, this will be at a rate of 2 Hz or 120 shock waves/min, which results in a much quicker treatment time. However, *in vitro* studies indicated that firing at higher rates resulted in worse stone fragmentation [131]. The likely explanation for this effect is that bubbles from a previous shock wave can last for many seconds [91] and that these bubbles can end up shielding the stone from the next shock wave [132]. The *in vitro* studies were followed up by animal studies which confirmed better stone comminution at
slower rates [73]. A number of clinical trials have reported data on the effect of rate and a meta-analysis of those studies indicates that firing at a lower rate does result in improved stone comminution [133]. Further data in the pig model indicate that delivering shock waves more slowly results in less tissue damage [64]. It may also be that at slower rates the cavitation bubbles from the previous shock have more time to dissipate and in addition cumulative shear should produce less damage [111].

The attempt to design a lithotripter so that it can be operated "anesthesia free," on the other hand, has proven to be a much more difficult problem. Discomfort during shock-wave treatment is due primarily to the sensation of cutaneous pain over the area of shock-wave entry at the surface of the body. One attempted solution was to widen the aperture at the shock source in order to spread the energy over a broader area. A wider aperture broadens the acoustic field along the shock-wave axis, but it narrows the focal zone of the pressure pulse. Many current lithotripters have a very narrow focal zone, of the order of 5 mm or less. Some of these lithotripters also generate huge peak positive pressures (in excess of 100 MPa). Because of respiratory motion, shooting at a stone using a narrow focal zone proves to be harder than when using a broad focal zone. Even if the shock wave could be kept directly on target, use of a narrow focal zone is less effective at delivering energy into the stone. Further, regardless of which lithotripter is used, lithotripsy is uncomfortable for the patient. If the patient is not sedated they will move to try to get more comfortable. Thus, attempts to build a totally anesthesia-free device have not yet been successful.

Another perceived disadvantage of the Dornier HM3 was the limited lifespan of the electrodes; it is necessary to replace the electrode one or more times during a treatment. EMLs and PELs do not use electrodes, which is an advantage in terms of cost, time, and convenience. In addition to the need to periodically change electrodes, electrode wear is an issue with EHLs. As the spark gap widens with use, there is increased variability in the path of the arc discharge. Also, as the spark gap widens, it takes higher a voltage to initiate a spark. Several manufacturers of current EHLs have found various ways to improve electrode life and some use designs such as encapsulation in an electrolyte-filled housing that extend the life of the electrode [17, 134].

**Future directions in lithotripter design**

More recent developments in lithotripsy could herald a positive change for the future of SWL. That is, there has been an effort to introduce novel approaches in lithotripter design that build upon well-tested theory and positive experimental results, targeting ways to improve stone breakage and reduce tissue injury. One approach is a response to the recent trend toward tight focal zone, high acoustic pressure machines. The Xi Xin Eisenmenger lithotripter (Model CS-2012) is a wide-focus and low-pressure lithotripter that generates the largest focal zone (18 × 180 mm) and lowest range of acoustic pressures (10–25 MPa) currently in use in clinical practice [128]. This machine was developed to test the hypothesis that a very broad focal zone could be used to enhance stone breakage by circumferential squeezing [82]. It has been reported in an early trial that this machine delivers a high stone-free rate (86%) and can be used anesthesia free [128].

Cavitation control may be a means to improve lithotripsy. The cavitation bubble cycle, i.e. the time for a bubble to grow and then collapse, lasts of the order of 300 μs in the free field and ~600 μs at the surface of a stone [135]. Studies have shown that cavitation bubbles generated by one lithotripter pulse can be manipulated by a second pulse [136, 137]. If the second pulse arrives while bubbles are in their early growth phase, further expansion is stopped and the bubbles collapse with minimal damage. If, however, the second pulse arrives later in the cycle, bubble collapse is accelerated and damage is enhanced. Thus, the timing of the two pulses is critical. Bailey originated dual-pulse lithotripsy and in his studies used twin shock sources oriented coaxially facing one another [138]. Others have built upon this concept and have developed lithotripters that fire multiple pulses along the same axis [41, 139–140] or machines that use dual treatment heads offset at an angle to accommodate the constraints imposed by the anatomy of a patient [141–143]. At the current time, dual-pulse lithotripsy is under development and testing. The concept holds promise, as this may be a means to tailor acoustic forces within the focal zone for better breakage of stones, hopefully with reduced collateral damage [144, 145].

The safety of lithotripsy is a very important issue. Shock waves cause trauma and any strategy that lowers the dose of shock waves needed to treat a patient should be welcomed. One way to reduce unnecessary shock wave impact on tissue is to track the stone during treatment and to only fire when the shock wave will hit the stone. Devices have been proposed which monitor stone location and only allow shock waves to be fired when the stone is at the focus of the lithotripter [146–150]. A device has also been proposed to exploit acoustic time-reversal to dynamically change the focus of the lithotripter and so hit the stone even as it moves [151, 152]. Such concepts have the potential to dramatically reduce the number of shock waves required to break a stone. However, clinical devices do not currently employ real-time tracking.
Conclusions

SWL is a superb example of the successful transition of engineering technology into the clinical area. We have outlined the underlying acoustic principles that describe: the generation of the shock pulse, focusing, nonlinear distortion, coupling of the shock source to the body, and absorption of sound by the body. The exact mechanisms by which shock waves can damage stones and tissue are still not fully understood, although it is likely that direct stresses and cavitation are dominant in stone fragmentation and that cavitation is dominant in tissue injury. Improvements in lithotripsy, whether through improved use of existing lithotripters or through the development of new technologies, are likely to come only from an improved understanding of the acoustics and the physics of this problem.

In this chapter we have attempted to make the following main points:

• Most lithotripters produce a similar type of shock wave, which consists of a leading positive pressure shock front (compressive wave), lasting about 1 μs, followed by a negative pressure trough (tensile wave), which lasts about 3 μs. There is a large range in the amplitude of the shock waves used with peak positive pressures of 30–110 MPa, depending on the type of shock source and the power setting.

• Various types of shock-wave sources and focusing mechanisms have been exploited in lithotripsy. EMLs and EHLs dominate the lithotripsy market today.

• The size and dimensions of the focal zone are controlled by diffraction. Typically, EMLs have a smaller focal zone than EHLs and generate substantially higher peak positive pressures. A smaller focal zone is not necessarily an advantage as patient motion means that the stone can easily spend a significant amount of time outside the focal region. Currently, there is no good metric to determine equivalent action of different types of machines.

• Shock waves are coupled into the body using a water path that ideally is devoid of bubbles. Most current lithotripters use an enclosed water path in which the shock head is capped by a rubber membrane of low acoustic impedance. Such dry lithotripters tend not to be as efficient as the older water-bath lithotripters in which the patient is immersed in water during treatment. This reduced efficiency could be due at least in part to poorer coupling. Coupling can be improved by minimizing handling of the gel.

• The acoustic properties of tissue are very close to those of water, but tissue absorption and inhomogeneities lead to modest reduction of the peak positive pressure of the shock wave at the kidney. This is a significant finding for it validates in vitro experimentation as being representative of the in vivo condition.

• It appears that multiple mechanisms are responsible for stone fragmentation. Most likely shear waves generated at the outer surface of the stone generate high tensile stresses to induce the first fractures. Then cavitation grinds down the fragments into pieces small enough to pass.

• For tissue injury it appears that once detectable cavitation occurs, the kidney tissue suffers widespread damage. The initiation of detectable cavitation appears to be associated with rupture of smaller order vessels and this may be due to mechanical rupture of the vessel wall by direct stress or by localized invagination induced by a collapsing bubble.

• Firing shock waves more slowly results in better stone comminution and reduced tissue damage.

• Since its inception, lithotripsy has undergone a fascinating evolution. Water-bath-type, EHLs have given way to modular, highly portable lithotripters, many of which employ electromagnetic shock-wave generators. Most lithotripters are now performed using mobile units delivered by truck to subscribing hospitals. This improved convenience has come at a price, as stone retreatment rates have increased and reports of collateral damage are on the rise. One explanation is that the newer lithotripters are not as efficacious and have the potential to cause more collateral damage.

• New technologies of shock-wave delivery are now being applied to patient treatment. Dual-pulse lithotripsy uses two shock heads to fire separate pulses with the potential for control over the properties of the acoustic field leading to improved efficacy and safety. Likewise, initial success with a new lithotripter that produces a very broad focal zone and is operated at low peak positive pressures suggests that a return to some of the features of the original lithotripter could also be a step toward improved lithotripsy.

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