Shock Wave Physics
for Urologists
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To my parents, whose example, love, and teachings are with me every day.

To my brother Mario, for his unconditional support for many years.

To Cecy for giving me peace and so much love.

To my daughter Melanie, for being the greatest gift of my life.
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Extracorporeal shock wave lithotripsy (ESWL) was introduced clinically in February of 1980 in Munich, Germany. Since that time ESWL has become the primary, non-invasive treatment modality for patients with stones in the kidney or ureter. Seventy-five per cent of non-selected patients can receive ESWL as monotherapy with stone clearance rates of approximately 80%, while 25% of patients require endoscopic surgery either as monotherapy or in conjunction with ESWL. Open surgery is basically extinct for the management of urological stone disease. Although ESWL is the mainstay of urological stone treatment, proper patient selection and proper execution of the ESWL procedure, which govern its successful outcome, are still complex even for the experienced endourologist.

In this book the author, a renowned physicist who has spent more than 20 years conducting basic and applied research in the field of acoustic shock wave propagation, provides the urologist with an excellent, comprehensive, and easy to understand explanation of the physical properties of the shock wave phenomenon. Urologists can learn how they can improve treatment outcomes by appreciating the physical properties of the shock wave propagation and the components of a lithotripter. By following the author’s easy-to-read explanations and practical instructions every urologist can indeed improve his/her performance with this technology.

In addition, the chapter on the different ESWL machines gives valuable insight in selecting a device best suited for one’s individual practice. The research chapter is essential reading for any urologist with a research interest in shock wave treatment and as such provides an up to date overview of current research endeavors and future directions.

The author needs to be applauded for having taken on single-handedly this enormous task of presenting the practicing urologist for the first time with a comprehensive manual of the basics and inner workings of this complex and highly successful technology.

Gerhard J. Fuchs, MD., FACS
Los Angeles, CA, U.S.A.
Misconceptions regarding shock wave physics are still common among primary care physicians, patients, lithotripter operators, and sometimes urologists already in practice. This book, written by a physicist for urologists, represents a starting point to understand the basics of shock wave physics in order to perform safer and more efficient ESWL treatments. It may also serve as a tool for teaching graduate students and young urologists and as a guide for technicians at the beginning of their ESWL training. Easy to understand and useful advice from the standpoint of physics can increase ESWL success rates. Hospital managers and urologists who have the responsibility to choose a convenient lithotripter for their hospital may also find part of this book helpful. Well-designed treatment protocols are needed in many hospitals to improve results and to prevent over-treatment and re-treatment of the patient. Intensive and certified training should be required by hospital authorities and offered by lithotripsy societies worldwide to increase patient safety. The present book is intended to be a step in this direction. Some concepts treated in this book could also be useful to physicians using shock waves for medical applications different from ESWL or to direct manufacturers towards the design of more efficient and safer equipment.

The physics behind shock wave lithotripsy is a large subject. Because of this, the range of phenomena discussed is extensive; however, biomedical engineering and research progresses so rapidly that a book on applications of shock waves to medicine can never be up-to-date. The idea of writing this book evolved from courses on principles and medical applications of shock waves for graduate students, urologists, and scientists at hospitals and meetings in different countries.

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Introduction

Urolithiasis is a significant health problem that has been increasing worldwide over the last decades (Anderson 1966). Between 5 and 15% of all people will develop a urinary calculus during their lifetime (Soucie et al. 1994; Kerbl et al. 2002). In the United States, the annual costs for treatment of urolithiasis are about $1.8 billion USD (Clark, Thompson & Optenburg 1995).

Before 1980, the majority of patients with proximal urolithiasis and symptomatic nephrolithiasis needed open surgery. Fortunately, percutaneous nephrolithotomy, retrograde ureteroscopic intrarenal surgery, laparoscopic surgery, and extracorporeal shock wave lithotripsy now allow almost any urinary stone to be removed from the upper urinary tract without open surgery. *Litho* is the word used for stone in Latin. *Tripsy* means to crush. Extracorporeal shock wave lithotripsy, generally abbreviated ESWL\(^1\) or SWL, refers to the use of shock waves, generated outside the body, to break up kidney stones, obviating the need for invasive techniques. It is one of the great medical advances of all time (Chaussy et al. 1979; Chaussy, Brendel & Schmied 1980; Chaussy et al. 1982a; Chaussy et al. 1982b; Chaussy & Schmiedt 1984; Brendel 1986; Chaussy & Fuchs 1987; Newman et al. 1987; Chaussy & Fuchs 1989; Delius & Brendel 1990; Rassweiler, Henkel & Köhrman 1992; Chaussy et al. 1995; Loske 2001). The apparatus to perform ESWL treatments, called *lithotripters*, are composed of an energy source or shock wave generator, a focusing unit, an imaging system, a coupling device, and a patient treatment table (Lingeman 2007). If the patient is properly positioned, shock waves enter the body with some attenuation, become focused on the calculus, and fracture it. Shock wave targeting is performed by fluoroscopy or ultrasound imaging systems. Several thousand shock waves may be needed to comminute the stone completely. Calculi fragments pass spontaneously through the urinary tract, and patients are free of stones a few days after treatment. The goal of ESWL is to provide efficient fragmentation of calculi with the least amount of tissue damage. ESWL requires the smallest amount of anesthesia of all methods used to treat renal calculi. However, many patients have hematuria after treatment.

\(^1\) © Dornier Medizintechnik GmbH
A physical exam, x-ray, blood and urine tests, an intravenous pyelogram, and in some cases a spiral computerized tomography (CT) scan are done before ESWL. Most cases can be treated with local anesthesia and intravenous sedation. Epidural or general anesthesia may also be used. Protective headphones may be worn, because shock wave generation can be loud.

After its introduction, ESWL changed the management of renal and ureteral calculus disease. Endoscopic intracorporeal lithotripsy is reserved as the treatment for stones that are more complex or when initial ESWL treatment fails (Skenazy et al. 2005). In some cases, ESWL has been used to treat gallbladder stones, pancreatic concrements, and stones of the salivary glands.

Considering that an ESWL patient can return to work in about 48 hours, and that, for instance, in the USA about 700,000 patients per year are ESWL candidates, the benefit to the health care services is not in doubt (Clark, Thompson & Optenburg 1995; Levy, Adams-Huet & Pak 1995; Pearle, Calhoun & Curhan 2005). Today, lithotripters are in use in about 35 countries, and millions of ESWL treatments have been performed successfully. Nevertheless, we have not reached the end-point of technical developments in ESWL. Almost three decades after the first ESWL, the techniques and clinical devices are still evolving, and improvements to increase stone fragmentation efficiency and reduce pain and tissue trauma are constantly being sought. Research is still necessary to improve understanding of calculi fragmentation and tissue-damaging mechanisms. Therefore, a chapter on ESWL research was included.

Currently shock waves are also used in Orthopedics and Traumatology (Haupt et al. 1992; Haupt 1997; Loske 2001). Shock wave therapy as the first-line treatment for Peyronie’s disease has been reported (Skolarikos et al. 2005). A device for extracorporeal shock wave therapy in cardiology, called Modulith SLC (Storz Medical AG, Kreuzlingen, Switzerland) is already on the market. The uses of shock waves in oncology and gene therapy as well as the transfection of cells are promising approaches for treatment and research (Russo et al. 1986; Russo et al. 1987; Hoshi et al. 1991; Brümmer, Suhr & Hülser 1992; Steinbach et al. 1992; Oosterhof et al. 1996; Lauer et al. 1997; Bao et al. 1998; Armenta 2005; Armenta et al. 2006). The bactericidal effect of shock waves is another interesting research topic, with possible applications in medicine and chemistry (Loske et al. 1999; Loske et al. 2002a; Álvarez et al. 2004).
I
Brief history of extracorporeal shock wave lithotripsy
To trace the history of extracorporeal shock wave lithotripsy to its origin is not easy. As in many research fields, innovative ideas seem to have appeared in different institutions and countries. A spark gap with an ellipsoidal reflector for medical uses of shock waves was described in the late forties (Rieber 1947). The invention refers to a device for generating shock waves in oil using the so-called spark gap method in order to destroy brain tumors. This shock wave generator (see Figure 1) is surprisingly similar to modern electrohydraulic shock wave generators.

In 1950, Yutkin proposed the idea of disintegration of urinary tract stones by using an endoscopic electrohydraulic shock wave generator (YPAT-1 Medexport, USSR, Moscow). This apparatus was composed of a condenser rack to store electrical energy and a flexible tube with coaxial bipolar electrodes at the tip. The device was also used successfully for bladder stones in the early 1970s. Another successful electrohydraulic endoscopic lithotripter was the German Riwolith. Since 1951, experiments for the contact-free fragmentation of urinary calculi and gallstones have been conducted using continuous wave ultrasound (Mulvaney 1953; Häusler & Kiefer 1971; Häusler 1985). Stone fragmentation was achieved, but it was accompanied by severe damage when applied to living tissue. Therefore, extracorporeal ultrasound lithotripsy was not pursued.

In 1966, the German aircraft manufacturer Dornier performed experimental work on shock waves associated with high-speed phenomena in Friedrichshafen, Germany. Pitting seen on supersonic aircraft by shock waves generated after collision with micrometeorites was a main concern. Erosion damage to aircraft caused by rain was another stimulus to research in this field. During experiments, a Dornier employee touched a plate at the precise moment when a high-speed projectile hit it. He felt an “electric shock”, but measurements revealed that his sensation was not due to electricity. We now know that shock waves passing through the body may cause the same sensation as electric shocks.

In 1969, the physicist Eberhard Häusler from the Technical University of Saarbrücken, Germany and Armin Behrendt and Günther Hoff of Dornier discussed the possibility of using extracorporeal shock waves...
for medical purposes. Studies on the degradation of animal tissues resulting from shock wave application were done at Dornier from 1969 to 1971. Underwater in vitro experiments did prove the feasibility of kidney stone destruction by shock waves in 1972 (Hepp 1972). Häusler reported initial in vitro studies during a conference of the German Physical Society in 1971, leading to studies with Manfred Ziegler, a German urological surgeon who was teaching at the University of Saarbrücken.

In 1974 the German government supported a huge research program in order to develop a clinical device. Egbert Schmiedt, director of the Department of Urological Surgery, and Walter Brendel, director of the Institute of Surgical Research at the Ludwig-Maximilians University in Munich, led important animal studies (Eisenberger, Chaussy & Wanner 1977). Ferdinand Eisenberger and Christian Chaussy were named representatives for these two departments. Thanks to their effort, ESWL developed at astonishing speed. Chaussy created a model to implant human kidney stones into the renal pelvis of healthy dogs in order to test ESWL (Chaussy & Staehler 1980). Bernd Forssmann from Dornier was significantly involved in the technical development.

It is noteworthy that the first experimental lithotripter had a rubber membrane and no water tub to couple shock waves into the body of the animal. However, since the membrane did not permit effective transmission of the shock wave, the second experimental device had a water bath. Shock waves were generated using the electrohydraulic principle described in Section IV.1. An ultrasonic localization system was developed between 1976 and 1977, but the device was not good enough for localizing stones. An x-ray system was installed in 1978, and extensive animal experiments
were performed during 1978 and 1979. Chaussy and coworkers published results describing successful ESWL (Chaussy et al. 1978; Chaussy et al. 1979). The first Human Model Lithotripter (HM1) was finished by Dornier in 1979 and installed in the Institute of Surgical Research of the Ludwig-Maximilians University, Munich, Klinikum Grosshadern (see Figure 2). The first clinical application of shock waves was performed in February 1980 by Christian Chaussy and Dieter Jocham (Chaussy, Brendel & Schmiedt 1980; Brendel 1981). ESWL gained rapid acceptance and revolutionized the treatment of urinary stone disease. Two hundred and twenty patients were treated on the HM1 between 1980 and 1982. A second Human Model (HM2) was tested in 1982. The famous HM3 was developed in 1983 (see Figure 3). It was the first commercial extracorporeal lithotripter in clinical service worldwide and was introduced in the United States in 1984 at the Methodist Hospital, in Indianapolis, Indiana (Evan & McAteer 1996). FDA approval for the HM3 was obtained in 1984, and Japanese approval followed one year later. By 1986, more than two hundred Dornier lithotripters were installed worldwide and about 250,000 successful treatments had been performed. In 1986, Drach et al. published the first report of results from six sites in the United States that had an HM3.

According to the original operating manual, the Dornier Kidney Lithotripter HM3 is indicated for use in the disintegration of upper urinary system stones and contraindicated for treatment of gallstones, lower ureteral stones, and bladder stones. At present, this machine is still used at some hospitals and is considered “The Gold Standard” of ESWL (Preminger 1995). The majority of animal and human studies reported so far have utilized the HM3 lithotripter. The term unmodified HM3 stands for the original machine.
having an 80 nF capacitance. The modified HM3 (and HM4) shock wave generator has a 40 nF capacitance and slightly larger reflector aperture to produce a tighter focal zone. In 1998 the Dornier HM3 was still one of the most widely used lithotripters in the United States (Evan et al. 1998b).

After the first ESWL systems, several companies developed so-called second- and third-generation lithotripters, offering improvements in patient positioning and decreased anesthesia. However, only a few of these machines were comparable in treatment efficacy to the HM3. The HM4, a “dry” lithotripter with a water cushion shown in Figure 4, was the last device of the Human Model series manufactured by Dornier.

A different principle to generate shock waves for ESWL, using piezoelectric crystals, was developed by Kurtze and Riedlinger at the University of Karlsruhe, Germany in 1980 (Coptcoat, Miller & Wickham 1987). The first ESWL using a piezoelectric lithotripter was performed in 1985 on a machine manufactured by Richard Wolf Endoscope GmbH (Figure 5). Electromagnetic lithotripters were also developed at the beginning of the 1980s (Figure 6). Some of them were configured with bilateral shock wave generators (Wilbert et al. 1987; Loske & Prieto 1999). The first treatment with this system was in 1986 (Coptcoat, Miller & Wickham 1987).

ESWL in the gallbladder was first reported in the late eighties (Delius, Heine & Brendel 1988; Sackmann et al. 1988; Sauerbruch & Stern 1989; Vergunst et al. 1989;...

In 1988, the first shock wave treatment of a non-union fracture was successfully performed on a human patient in Bochum, Germany. A special orthopedic shock wave device, called OssaTron (High Medical Technologies, AG) became available in 1993 (Valchanou & Michailov 1991).

The use of C-arms for ESWL systems was introduced by Direx Medical Systems Ltd., Petah-Tikva, Israel, in the late 1980s on a compact lithotripter and is now very common. Compact lithotripters (Figure 7), manufactured by several companies, followed.


**Figure 7** Photograph of the Tripter X-1 electrohydraulic extracorporeal lithotripter, manufactured by Direx Medical Systems in Petah-Tikva, Israel. This was the first compact lithotripter manufactured by Direx (Bierkens 1992).
II Basic physics
This chapter will present a few principles of physics in a clear, logical manner in order to strengthen the understanding of concepts often found in apparatus descriptions, user manuals, and research articles related to ESWL. It is meant to be a brief reminder, not a complete basic physics course. Mathematical equations are kept to a minimum. Each section deals with a topic that would need, for a detailed description, a whole book by itself. More information on these topics may be found in the literature cited at the end of the book (Tippens 2001; Serway & Jewett 2004; Halliday, Resnick & Walker 2005). Readers with some background in physics may skip this chapter and move forward.

II.1 Work and energy

II.1.1 The physical concept of work
The physical concept of work is fundamental in many sciences. It is defined as the external force acting upon an object to cause a displacement, or as the energy transfer that occurs when an object is moved over a distance by a force. In order for a force to do work on an object, it must cause a displacement. If the force is constant, work equals the distance multiplied by the component of the force acting along the direction of motion. Work can also be done by compressing a gas, by moving an object with a magnetic force, or by crushing a kidney stone with a shock wave. In physics, it is understood that no work is done unless the object is displaced in some way. For instance, holding a book without moving it for a certain time can cause fatigue; however, no work is done, because there is no displacement. Work is considered to be negative if the force is opposite to the motion. In this case, physicists say that energy is taken from the system.

Work should be defined only relative to a frame, i.e., an initial reference state. The selection of this reference is arbitrary. In most cases it is selected in order to simplify the problem to be solved. There are also types of work that are not evidently mechanical. An example is electrical work. In this case an electric force acts on a charged particle. The transfer of energy in the form of heat can also be associated with work. This was established in the 19th century, when the principle of the conservation of energy (first law of thermodynamics) was formulated.

Any unit of force, for instance newton (N), multiplied by any unit of distance, like meter (m), equals units of work. Work is expressed in the same units as those used for energy, like the joule (J), which results by multiplying one newton by one meter. The erg is another unit of work, used when the displacement is measured in centimeters and the force in dynes. In the English system, the foot-pound is used as unit of work.

II.1.2 Energy
In physics, energy is defined as the capacity for doing work. There are several types of energy. Energy can be potential, kinetic, electromagnetic, etcetera, and it can be converted from one form to another. Stored energy is named potential energy, because it has the ability to do work. One possibility is that a system stores energy as
the result of the position of its parts. An example is the elastic potential energy in a spring-mass system. There are several other types of potential energy, like nuclear or chemical energy. Kinetic energy is always related to motion. Any object at motion has kinetic energy and could do work. The energy due to vibrational motion, the energy due to rotational motion, and the energy due to translational motion are all examples of kinetic energy. An increase in speed produces an increase in kinetic energy. Heat is another form of energy. It is transferred from hot regions to cool regions because of a temperature difference.

In principle, energy can be reduced to either kinetic or potential energy. Energy cannot be destroyed or created; it can only be transformed from one type to another. Potential energy is converted into kinetic energy as an object falls to the ground. Another example is the conversion of chemical energy into electrical energy inside a battery. As already mentioned, units of work and energy are the same.

II.1.3 Power
A given amount of work can be done rapidly or slowly. Power gives information on the rate at which work is done. Power can also be described as energy flow, i.e., the rate of change of the energy, and it can be calculated by dividing work by the time taken to do it. A low-powered engine would do certain work in a longer time than a motor having more power. The standard metric unit of power is the watt (W), named after the British inventor James Watt (1736-1819). A unit of power is equivalent to a unit of work divided by a unit of time, that is, one watt is equivalent to one joule per second. Another famous unit of power is the horsepower.

In electricity, power can be calculated by multiplying the voltage by the electric current (see Section II.4). The resulting unit is also the watt, i.e., one volt multiplied by one ampere equals to one watt (1V × 1 A = 1 W). Electric power can be varied using different combinations of current and voltage. Passage of electric current through a wire produces heat. The amount of heat per second is equal to the product of the electrical resistance of the wire and the square of the current. The heat produced per second equals the power. This relationship is called Joule’s law.

Extracorporeal lithotripters deliver large amounts of electric energy, previously stored in a capacitor bank, in very short periods of time. As an example, the electric power of an electrohydraulic shock wave generator (see Section IV.1) is the product of the voltage on the electrodes and the current flowing between them. Nevertheless, electric power of a lithotripter does not mean “power to disintegrate a stone” or “time to do the work of breaking a stone.” Lithotripters operating at high electric power are not necessarily better than lithotripters using lower electric power.

In some areas of physics, like optics, the word power is also used to define the ability of a lens to focus light, which is a completely different concept. In this case, power is equal to one divided by the focal length of a lens.

II.2 Elastic properties of solids

Solid, liquid, and gas are the states of matter normally mentioned in textbooks. Sometimes, plasma is referred to as the fourth state of matter. Solids have a definite volume and shape. Liquids have a definite volume, but their shape depends on the shape of the container. Nevertheless, there are some solid-like materials that tend to flow like liquids. Since they flow so slowly it is not easy to realize that they are liquids. Plastics are an example of this category. Liquids and gases are also named fluids. Unconfined gases have neither definite volume nor definite shape. The elastic properties of solids are important for the study of shock wave propagation through human tissue and for the design
of physical models to describe the interaction of shock waves with kidney stones.

The difference between rigid and flexible solid materials is related to the interatomic forces inside them. The atoms of some materials form very rigid lattices, i.e., repetitive arrangements in which all atoms have well-defined distances between each other. Flexible solids are made of atoms that are aligned in long flexible chains. The forces between chains are relatively weak. All materials are elastic to some extent. Their shape can be changed by compressing, pulling, or twisting them. In order to change the shape or destroy a solid object, it is necessary to apply external forces on it; however, internal forces will oppose the deformation.

Two physical concepts are generally used to study the deformation of solids: stress and strain. Stress is defined as the external force applied to an object per unit cross-sectional area. The consequence of applying stress to an object is strain. Strain is a quantity that gives information on the deformation achieved after applying stress. Strain is proportional to stress, as long as the applied stress is not too large. The ratio of stress to strain is called the elastic modulus of a material. The value of the elastic modulus depends on the mechanical properties of the material, but also on the way stress was applied to it. For instance, deformation of a rod of nylon will be different if we try to compress it longitudinally than if we try to bend it. Thus, three different types of deformation are normally considered for solid objects. Each type has its own elastic modulus. The mechanical resistance of a solid to variations in its length is measured by the Young's modulus; the resistance to shifting of planes inside a solid is determined by the shear modulus; the resistance to a change in volume is measured by the bulk modulus. If the stress on an object exceeds a certain value, the material will fracture.

Considering the arrangement shown in Figure 8, if the external force \( F \) is applied perpendicular to the cross section \( A \), tensile stress is defined as the ratio \( F/A \). The resulting tensile strain is defined as the ratio of change in length to initial length, i.e., \( \Delta l/l \). The Young modulus \( Y \) is the ratio of tensile stress to tensile strain:

\[
Y = \frac{F/A}{\Delta l/l}.
\]

The rod shown in Figure 8 will return to its original length after removing the force only if its elastic limit was not exceeded. It seems obvious that such an experiment could not be done with a fluid.

When a force acts on an object parallel to one of its faces, as shown in Figure 9, a different type of stress, called shear stress, appears. Shear stress is defined as the tangential force \( F \), divided by the area \( A \) of the face that is being sheared. In this case, \( \Delta d \) is the distance that the sheared surface has moved and \( h \) is the height.
of the sheared object. Shear strain is calculated as the ratio $\Delta d/h$. Shear modulus is defined as the ratio of shear stress to shear strain:

$$S = \frac{(F/A)}{(\Delta d/h)}.$$  

It plays an important role in buckling of shafts that rotate. Shear stress is also involved in bone fractures caused by bending. Once again, we realize that no shear modulus could be measured for a liquid. During ESWL, considerable shear stress may appear inside a kidney stone. It is generated by compressive waves and shear waves that develop as shock waves pass through the stone.

The bulk modulus gives information on the properties of a solid when it is under uniform pressure. In this case (see Figure 10) the object undergoes a change in volume without changing its shape. A uniform distribution of pressure on the whole object can be achieved when the object is immersed in a fluid. **Volume stress** is defined as the pressure on a surface, i.e., the ratio of the force exerted on a surface to the area of the surface. **Volume strain** is defined as the change in volume $\Delta V$, divided by the initial volume $V$ of the object. Analogous to Young modulus and shear modulus, bulk modulus is defined as volume stress divided by volume strain:

$$B = \frac{- (F/A)}{\Delta V/V}.$$  

The negative sign that appears is inserted so that the bulk modulus is a positive number. Several handbooks list the **compressibility** of solids and liquids, instead of the bulk modulus. The compressibility of a material is the reciprocal of the bulk modulus.

Solids have Young modulus, shear modulus, and bulk modulus; however, liquids have only bulk modulus. This is because liquids do not sustain tensile and shear stress, i.e., the liquid simply flows when we try to apply a tensile or a shear stress. Some materials, like concrete and most renal calculi, can strongly resist compression but can only weakly withstand tension.

### II.3 Density and pressure

To study the motion of a fluid and the propagation of mechanical waves through it is complicated and mathematically not easy to handle. In general it is important to have information on the density and the pressure as a function of time and space.

The density of a substance is defined as the ratio of mass to volume. It can be expressed in kilograms per cubic meters, and it depends on the pressure and the temperature of the substance. As an example, water has a density of about 998 kg/m$^3$ at 20° C and one atmosphere (atm). At 50 atm its density increases to 1000 kg/m$^3$. Whole blood has a density close to that of water (1060 kg/m$^3$).

The density of materials plays a significant role in propagation of mechanical waves. Strong reflections occur at boundaries where the density varies significantly. The interface between soft tissue and a kidney stone, or between soft tissue and lung cavities are two examples. Huge differences in density between air (about 1.2 kg/m$^3$) and living tissue are one reason why, for clinical applications, shock waves are generated in water and not in air. Since the density...
Shock waves may result from a sudden release of mechanical, electrical, chemical, or nuclear energy in a limited space and are defined as a sharp discontinuity through which there exists a sudden change in pressure, density, temperature, entropy, and particle velocity. Shock waves may propagate in a manner different from that of ordinary acoustic waves. They are mechanical waves that were known long before the discovery of explosive materials. Examples are the acoustic emission of some volcanic eruptions and thunderstorms, where an acoustic boom follows the electric discharge seen as lightning. Extremely fast heating of the huge spark channel displaces the surrounding air, analogous to an explosion.

Detonation of an explosive can generate a spherical shock wave. In this case, the pressure decays as it expands until it reduces to a sound wave. This happens because a given amount of energy is spread out over an increasing area (see Figure 14). Shock waves have been observed in all states of matter (Ben-Dor et al. 2001). Shock waves change the physical properties of solids and, thus, they can be used to study the equation of state (Rice, McQueen & Walsh 1958; Duvall 1976; Prieto & Renero 1982; Icaza, Renero & Prieto 1989).

As already explained, sonic booms are also associated with aircraft shock waves. Airplanes breaking the sound barrier generate very intense sounds. Our understanding of this phenomenon can be traced to the shock wave technology of bullets. Ernst Mach first showed that a projectile flying at supersonic speed produces a hyperbolic shock wave (see Figure 26) moving with the projectile. This bow wave has a higher-than-atmospheric pressure and propagates slightly faster than the ambient sound speed. Another wave is created at the tail. It propagates at less than ambient sound speed and has “negative” pressure.

Shock waves share properties with conventional ultrasound; however, some basic principles of shock waves should not be confused with those of ultrasound (Hamilton & Blackstock 1997; Cleveland & McAteer 2007). The main difference between ultrasound and shock waves is shown in Figures 11 and 12, respectively. Ultrasound is characterized by alternate compressions and rarefactions. It consists of a sinusoidal wave or modulated pulse train, having a defined frequency. Shock waves are high-energy waves, consisting of a single high-pressure peak with a steep onset and a gradual decline into a pressure trough. When transforming the pressure-time plots shown in Figures 11 and 12 into the frequency domain (Figures 28 and 29), it can be seen that the frequency domain of shock waves is
much wider than that of ultrasound. Ultrasound has a characteristic frequency; in contrast, shock waves are composed of many frequencies. This is referred to as the frequency spectrum of a shock wave.

An advantage of shock waves for some medical applications is that they undergo less attenuation than ultrasound when propagating through the human body. This is because high frequencies are damped more than low frequencies. As already mentioned, many years ago, experiments for the extracorporeal fragmentation of urinary calculi were conducted using ultrasound (Mulvaney 1953; Häusler & Kiefer 1971); however, clinical application was not feasible, because of its high attenuation in tissue (low depth of penetration) and the risk of thermal injury to tissue. Even if the pressure generated during ESWL is about one order of magnitude higher than that generated by ultrasound imaging devices, no thermal injury has been reported.

Whether they are technically shock waves or not, in general, ESWL pulses are called shock waves to denote the high-amplitude pulse that is generated.

Formally it is only the sharp positive pressure jump that should be called “shock”. Shock waves are sometimes referred to as lithotripsy pulses, which is a term used to denote nonlinear high-pressure impulses with a very short rise time and a wide spectrum of frequencies. As mentioned in Section II.5.5, the velocity of any mechanical wave, like sound or ultrasound, increases as the compressibility of the medium decreases, i.e., velocity increases as density increases. In water, for instance, sound travels faster than in air. It is also possible that the density of the medium changes as the wave propagates through the medium. As shock waves propagate, they cause the medium to become denser.
Initially, a large positive pulse is formed due to a sudden compression of the media (see Figure 30(a)). The compression pulse will be distorted (Figure 30(b)) as it travels through the medium (water). This happens because high-pressure parts travel faster ($v_1$) than low-pressure parts ($v_2$) of the wave. As the high-pressure parts travel faster, they push the pressure profile forward increasing the pressure at the wave front, until a steep (shock) front is formed (Figure 30(c)). A compression pulse transforms into a shock wave at the instant when the pressure profile does not change (pile up) any more. Shock fronts can have different widths and amplitudes. The shock wave remains in its state until no more energy is delivered from the wave crest. The aging shock wave has a smooth pressure profile. During propagation, the shock front will loose energy and transform into a sound wave. To prevent shock waves from loosing energy, focusing devices like lenses or reflectors can be used (see Chapter IV). Shock wave focusing is the cornerstone of ESWL. After passing through the focus of the lithotripter, shock waves diverge, gradually reducing their amplitude.

Shock waves are not visible, because they are very rapid pressure fluctuations; however, they can be made visible by using suitable optical instrumentation (Schlieren photography). The wavefront of an ESWL shock wave has a thickness less then one millimeter (Eisemenger 1964). Like other compressive waves, shock waves undergo refraction and reflection when passing from one medium to another. Shock waves will refract and reflect more if acoustic impedances at the interface differ more (see Section II.5.9). Whitham studied the relationship between the shape of a shock front and its propagation velocity. His theory is known as the ray shock theory or geometrical shock dynamics theory (Chester 1953; Chisnell 1957; Whitham 1959). Under certain circumstances, this theory may be used as an approximation to predict the propagation of lithotripter shock waves.

The steepness of a shock wave is defined by its rise time (see Figure 12). It depends on the shock wave generation mechanism. Physically it is not possible to generate a completely unipolar positive or negative pressure peak. Time duration, i.e. the time between the positive peak and the negative peak of a lithotripter shock wave, is between about two and four microseconds. Time durations are similar in all lithotripters.

When the wave passes through medium, there is energy loss due to friction. This absorption of energy causes reduction in pressure amplitude. The pressure amplitude decreases exponentially as the travel distance through tissue increases. Tissue penetration depends on the frequency of the wave.

Spark gap generated shock waves (Section IV.1) are formed almost immediately after the electric energy is released. In piezoelectric systems (Section IV.2), the shock wave develops while the pressure pulse, generated by the crystals, is traveling through the medium towards the focal zone. In electromagnetic ESWL (Section IV.3), the shock front develops after passing through the lens. Due to the finite aperture size of the shock wave sources, diffraction takes place and has to be considered in addition to non-linear propagation.

The pressure pulse in the focal region of a lithotripter consists of a short compression pulse with a peak pressure between 30 and 150 MPa (approximately 10 ns rise time) and a phase duration of $0.5 - 3 \mu s$, and a subsequent decompression pulse, referred to as the “negative” pressure peak, with a tensile peak of up to - 30 MPa, and a phase duration of 2 to 20 $\mu s$ (see Figure 12).

Lithotripter shock waves have broad frequency spectra in the 20 kHz to a few MHz range. Most energy is between about 100 kHz and 1 MHz, with a peak at about 300 kHz. Total pulse energies are in the range of 10 to...
100 mJ (Folberth et al. 1992) and energy densities are between about 0.2 and 2.0 mJ/mm². Energy density is defined as the amount of acoustical energy transmitted through an area of 1 mm² per pulse. This should not be confused with the total acoustical energy per released shock wave, defined as the sum of all energy densities across the beam profile multiplied by the area of the beam profile. Most articles dealing with the suppression of tumor growth by pressure pulses refer to these shock waves as “high-energy shock waves”, even if lithotripter shock waves are used (Russo et al. 1986; Oosterhof et al. 1990; Oosterhof et al. 1996).

In general, the temperature behind a shock front is higher than that in front of it. The ratio of the two temperatures depends on the shock wave energy and is proportional to the square of the Mach number (ratio of shock wave velocity and speed of sound). For distances \( R \) (several inches) and energies \( E \) (several hundred joules) characteristic of ESWL, the pressure \( p \) can be approximated by:

\[
p = 72 \left( \frac{E^{1/3}}{R} \right)^{1.18} e^{\theta/\Theta},
\]

where \( p \) is given in pounds per square inch, and \( \Theta = 1.09 \times 10^{-3} \left( \frac{E^{1/3}}{R} \right)^{0.185} \) ms (Marshall et al. 1988). By substituting energy values in this equation, it is possible to obtain a pressure profile (variation of pressure), similar to the one shown in Figure 12, for a given distance \( R \) from the energy source.

Fortunately, nonlinear effects are only prominent within a short distance of the focal area. This simplifies the equations that predict shock wave propagation. The correspondence between pressure and density is linear for low-pressure acoustic waves. In this case, the laws of linear acoustics are valid. As the pressure becomes higher, the medium is more difficult to compress, and the relationship between pressure and speed of sound becomes nonlinear and more complicated to predict. For nonlinear waves, different parts of the wave travel at different speeds.

Peak-pressure attenuation of shock waves in water is about 10 to 20% for a distance of 100 mm. As explained before, shock wave velocity in water depends on the pressure: the greater the pressure, the higher the velocity. Energy attenuation due to passage of a shock wave through the membrane of a lithotripter water cushion is about 20%; however, increasing the voltage on the shock wave generator can compensate for this loss.

The dynamic focus, focal region, or 6 dB zone of a lithotripter is defined as the volume in which, at any point, the positive pressure peak amplitude is equal to or higher than 50% of the maximum amplitude. The size of this volume varies depending on the shock wave generation mechanism, on the design of the shock wave source and on the energy (voltage setting). This should not be confused with the treatment zone or 5 MPa focus of a lithotripter, which is defined as the volume inside which any point has a pressure equal to or larger than 5 MPa. Further information on the physics of shock wave lithotripsy can be found in a chapter written by Cleveland and McAteer (2007).
Shock wave generation and focusing

Since the introduction of ESWL, more than 60 different varieties of lithotripters have become available. Even though continuous modifications are being made to extracorporeal shock wave lithotripters in order to reduce tissue damage and to improve disintegration of calculi, the basic working principle of the systems has not changed. Most extracorporeal lithotripters use one of three different shock wave generation techniques: electrohydraulic, piezoelectric, or electromagnetic (Brümmer et al. 1992; Loske 2001). They generate shock waves outside the patient’s body and concentrate them on the urinary stone. In general, several hundred shock waves are needed to comminute the stone completely. All lithotripters require a focusing system to concentrate the shock wave energy onto the stone. The electric circuit of all shock wave generators is similar: a high-voltage capacitor is charged and rapidly discharged, providing energy to the electro-acoustic transducer. From the beginning of ESWL, the tendency has been to obtain a smaller focal region, a larger aperture, and consequently reduced tissue exposure to the shock wave. In recent years, however, some authors suggest going back towards lithotripters with larger focal regions and lower pressure, and they are reporting good clinical results (Eisenmenger et al. 2002).

IV.1 Electrohydraulic shock wave generation

Electrohydraulic shock wave lithotripters induce shock waves by electrical breakdown (15 - 30 kV) of water between two electrodes located at the focus (F1) closest to a paraellipsoidal reflector (see Figure 22). Peak electric current at the underwater discharge is in the range 10 - 20 kA, depending on the inductance of the circuit. This discharge produces an electromagnetic field that could cause arrhythmia in the patient. Because of this, spark gap lithotripters are sometimes gated to the cardiac cycle of the patient.

Electric energy is stored in a set of capacitors by a power supply. The electrodes are connected to the capacitor terminals and submerged under water. The capacitors are charged and triggered to discharge across the underwater spark gap by means of a secondary spark gap. This secondary spark gap acts as a rapid switch. It consists of a pair of primary electrodes and a third trigger electrode sealed within a housing. A pressurized nitrogen atmosphere inside it prevents electrical breakdown. When a high-voltage pulse is delivered to the trigger electrode, the gas inside the switch is ionized allowing conduction to take place from the capacitor, across the primary electrodes, to the underwater spark gap. This trigger pulse is sent by the electronic circuits of the generator (trigger module). As a safety measure, a high-voltage relay closes when the power supply is switched off so that the capacitors discharge through a set of high-voltage resistors. A simplified diagram of the discharge circuit is shown in Figure 31. When the spark plug fires, dielectric breakdown occurs and a plasma bubble is generated at temperatures of about 20,000 degrees Kelvin (see Figure 32). A sinusoidal half-wave is created by the plasma expansion and isotropically radiated from F1. The positive pressure peak increases in speed during propagation until it overtakes the front end of the wave, forming a shock front, as explained in Chapter III. Shock waves are created at F1, partially reflected,
and concentrated at the second focus F2. Treatment efficiency and patient pain and trauma depend, to a certain extent, on the design of the reflector. It is remarkable that this shock wave generation method was described as early as 1947 by Frank Rieber (see Figure 1).

If the patient is properly positioned with his calculus located at F2, shock waves enter the body with some attenuation, are focused on the calculus, and fracture it (see Figure 33). A water bath or water-filled cushion couples the energy generated at F1 with the patient’s body. Most electrohydraulic lithotripters are loud. Protective headphones should be worn, particularly by personnel working many hours a day next to the lithotripter.
Spark gaps in water generate broadband pressure pulses with very short durations. Their pressures depend on several parameters, some of which can be controlled (discharge voltage, capacitance, water conductivity, temperature) and some which cannot. As explained before (Chapter III), an essential characteristic of spark gap lithotripters is the generation of a shock wave almost from the onset and not after focusing. Spark gap generated shock waves have shorter rise times than shock waves generated with piezoelectric or electromagnetic systems. The rise time of electrohydraulic generators was measured to be about 30 ns, although theoretically, it should be less than one nanosecond (Chitnis 2002). Only about 5% of the total energy of the discharge reaches the kidney stone (Coleman & Saunders 1989), indicating that, from the energy standpoint, this system is inefficient.

Following a single discharge, more than one shock wave is generated. The first (direct shock wave) is due to the water breakdown at F1 (see Figure 34). Its contribution to calculi fragmentation is very low. The reflected pulse reaches F2 tenths of microseconds after the direct pulse. Both waves are followed by another reflected pulse, generated when the plasma bubble at F1 collapses (not shown in Figure 34). There is evidence that the initial focused shock wave is responsible for most of the calculi disintegration. Collapse of cavitation bubbles (see Section V.2) at F2 can also be detected (Pishchalnikov et al. 2005; Chitnis & Cleveland 2006). Part of the initial pressure wave will not be reflected, forming a diverging cone (Figure 35).

Another aspect of shock wave focusing, which has clinical implications but is seldom mentioned, is the fact that focusing of positive and negative pulses is not equal. This is a consequence of what physicists call the nonlinear nature of the shock wave field and is expected from diffraction considerations. In general, the negative tail of the shock wave has its
maximum (negative) amplitude 10 to 30 mm before reaching F2. Since the negative phase of the shock wave generates acoustic cavitation, which is one of the main kidney stone fragmentation mechanisms, some authors (Sokolov et al. 2002; Sokolov et al. 2003) have suggested placing the stone not at F2, but two centimeters closer to F1 (see Section VIII.4).

Disintegration efficiency depends, to a certain extent, on the design of the reflector and the shape of the electrodes (Loske & Prieto 1993; Loske & Prieto 1996). Since the acoustic impedance of the reflector is related to the reflected energy, reflectors are generally made out of brass or stainless steel.

In water, lithotripter shock waves behave similarly to acoustic waves and may have high pressure values for relatively small generating energies. This is why a weak shock wave, generated at F1, is reflected and produces, at F2, an energy density similar to the one in F1. The law of geometrical acoustics described in Section II.5.9, which predicts that the angle of incidence $\alpha_1$ of an incoming spherical wave equals its angle of reflection $\alpha_2$, is valid only for low pressures. The design of a reflector for a lithotripter must take into account, however, that the angle of reflection $\alpha_2$ is larger than the angle of incidence $\alpha_1$ (Müller 1987a and 1987b). This difference increases as $\alpha_1$ becomes larger and also as the pressure rises (Figure 36). The phenomenon
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Shock Wave Physics for Urologists

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V. Shock wave comminution mechanisms

The pressure profile of a shock wave might be altered in order to maximize fragmentation efficiency while minimizing tissue injury; however, the physical mechanisms responsible for tissue trauma and stone comminution have yet to be determined in detail. This research field is still relatively new, and the problem is complex.

It is surprising how little information on shock wave interaction with renal calculi and renal tissue was available during the first ten years of ESWL. Even today, with a concentration of more than one lithotripter per million inhabitants in Europe, US, Canada, and Mexico, and although considerable progress has been made in understanding shock wave interaction with matter (Sass et al. 1991; Vakil 1991; Zhong & Preminger 1994; Lokhandwalla & Sturtevant 2000; Cleveland, McAteer & Müller 2001; Cleveland, McAteer & Williams 2002; Eisenmenger 2001), there is still not complete agreement as to how shock waves destroy kidney stones. Stones are known to fracture due to spalling, cavitation, squeezing, superfocusing, fatigue, and other mechanisms (Figure 56). It is also believed that stress waves and cavitation act synergistically, rather than independently (Zhou et al. 2004a; Zhou et al. 2004b). Nearly all kidney stones can be fragmented into a few pieces by shock waves. However, the challenge is that after ESWL, fragments need to be smaller than 2 mm to pass spontaneously in the days following treatment.

To date the only method to achieve sufficient pressure to fragment a stone is using focused shock waves. From the physical standpoint, material properties determine the response to ESWL. As illustrated in Figure 57, renal calculi have varying chemical composition, shape, and mechanical properties (Singh & Agarwal 1990). The most common type of stone contains calcium combined with either phosphate or oxalate. So-called struvite or infection stone are associated with an infection in the urinary tract. Uric acid and cystine stones are less common. Cystine and calcium hydrogen phosphate dihydrate or brushite calculi are very resistant to ESWL (Wang et al. 1993). Kidney stones, which develop from crystals that separate from urine and build up on the inner surfaces of the kidney, are brittle composite materials having weak spots and flaws, and are generally composed of crystals joined together by organic deposits. These organic deposits are weak. Compression and tension of the stone result in loss of cohesiveness, due to growth of microscopic flaws. Stones with a heterogeneous and laminated structure are more fragile than homogeneous calculi. In general, ductile materials are more resistant to shock wave action than brittle materials, because brittle materials have less deformation capability. More ductile stones,
like gallstones, absorb shock wave energy through plastic deformation. For this reason, only about 20% of all gallstone patients are suitable for ESWL, and ESWL in the gallbladder is performed only in a few centers (Maglinte et al. 1991).

Initially it was thought that shock wave peak pressure is the most important parameter for stone disintegration. The amplitude of the pressure pulse transmitted into the stone is an important factor in the fragmentation; however, it is not easy to precisely correlate shock wave amplitude and duration with stone disintegration. It is known that higher amplitudes generate stronger spalling effects and that the rise time and duration of the shock wave are related to shear forces produced inside the stone; however, as measuring techniques improved, acoustic energy was found to correlate with fragmentation much better than positive pressure amplitude, or rise time of the shock wave. In order to reduce pain, the tendency has been to reduce the maximum pressure of the shock waves used in ESWL.

V.1 Spalling

Because of the inability of tissue, blood or urine to transmit shear waves (see Chapter II), only longitudinal waves can propagate through these media; however, both longitude and transverse wave propagation have been identified inside renal calculi. Urinary stones are brittle materials, i.e., the tensile strength of most urinary stones is about one order of magnitude lower than their compressive strength. This is attributed to the existence of small cracks and cavities that weaken them in tension but not in compression. As already mentioned in Section II.2, an analogous situation is seen in concrete, which withstands large compressions, but only relatively little tension.

Stone damage may occur by conversion of positive pressure pulses into tensile stress inside the concre-
ments. As the shock wave enters the stone, the sign of the wave is not changed. However, on the back surface, where the shock wave exits the stone, a portion of the wave is reflected and becomes tensile. Incident and reflected waves add, and a fracture plane may develop at the point where the net effect of the two waves is sufficiently tensile to induce a cleavage plane (see Figure 58). This phenomenon is very sensitive to the orientation of the stone, i.e., the angle of incidence of the shock wave on its distal surface. The mechanism is called spallation or the Hopkinson effect. Figure 59 shows a photograph of a brittle artificial kidney stone, exposed to a few shock waves in vitro. In this case, damage was only observed at the distal (shock wave-exit) side of the stone.

Fig. 58 Spallation of renal calculi may occur because the compression pulse is reflected as a tensile wave at the distal surface of the stone. Since the tensile strength of most stones is much less than their compression strength, fracture is observed at the shock wave exit surface of the stone. The position of the spall line is determined by constructive interference of the inverted reflected wave and the trailing negative wave.

Theoretically, the fracture depth due to spalling is comparable to half the wavelength of the stress. The distance between the distal surface of the stone and the spall depends on the shock wave velocity inside the stone and on the duration of the shock. Shock waves with a short duration produce smaller spalled fragments. This is illustrated in Figure 60(a). As the duration of the pulse increases, fragments get larger (Figure 60(b)). Furthermore, materials having slower sound speeds generate smaller fragments than stones where shock waves travel at higher velocities. Since shock waves travel faster through materials having high acoustic impedances, hard stones will result in larger fragments. If the stone or fragment is too small, spalling cannot occur. Cleveland, McAteer, and Williams (2002) showed that square stone models, with a flat distal surface, produced a larger region of
high tensile stress than spherical stones. For circular stones the amplitude of peak tensile stress reduced as their size reduced. They concluded that during shock wave application, when fragments became 3 to 4 mm in size, mechanisms like spalling and superfocusing are no longer effective, and cavitation and other mechanisms are necessary to fully comminute the stone. Inner reflections of the two shock fronts create alternative compressions and rarefactions, producing small fissures. As these fissures become filled with liquid (urine), cavitation (see Section V.2) occurs with subsequent shock waves, contributing to a fast disintegration of the stone. The shock wave pressure profile at the focal region and the geometric and mechanical properties of the stone will determine whether fragmentation will occur more due to spalling or due to cavitation.

V.2 Cavitation

Cavitation, i.e., the growth and collapse of vapor bubbles, is a problem occurring in ship propellers, turbine blades, valves, and spillways. The phenomenon was discovered by Leonhard Euler in 1754. Cavitation is generated whenever there is a fast change of positive pressure into tensile stress, and it is known to contribute to erosion in hydraulic machinery and high-speed aircraft. Since the popular work of Rayleigh (1945), the collapse of cavitation bubbles has been treated by many authors (Crum & Fowlkers 1986; Crum 1988; Brennen 1995; Leighton 1997; Young 1999).

There is evidence that shock wave-induced cavitation, sometimes also referred to as acoustic cavitation, is one of the most important mechanisms for stone fragmentation during ESWL; however, it may cause side effects such as vascular damage, and perirenal and intrarenal hematomas. The control of cavitation seems to be a key factor to improve ESWL. Cavitation has been reported to contribute to stone fragmentation both in vitro and in vivo (Field 1991; Wiksell & Kinn 1995; Auge et al. 2003; Pishchalnikov et al. 2005; Chitnis & Cleveland 2006). Focused passive receivers (see Section VIII.3.4) and B-mode ultrasound have been used in attempts to find direct evidence of cavitation within the kidney and tissue (Coleman, Choi & Saunders 1996; Zhong et al. 1997a; Zhong et al. 1997b; Bailey et al. 2005).

When the trailing “negative” pulse of the shock wave is strong enough to make the fluid (water or urine) rip apart, cavitation occurs. This depends on the existence of micro-inhomogeneities in the liquid, the content of dissolved gases, viscosity, surface tension, temperature of the liquid, and applied pressure. Microbubbles of free gas or microscopically small solid kernels can be cavitation nuclei. In urine, nuclei of about 1 \( \mu \text{m} \) to 1 mm may be present. In untreated tap water the typical radius of an air bubble is approximately 3 \( \mu \text{m} \). The threshold for cavitation in water is about 0.5 MPa. This means that a tensile pulse of more than 0.5 MPa can generate cavitation; however, this depends also on the duration of the tensile pulse.

Bubbles may form along the shock wave path; however, a concentrated cloud of bubbles forms at the focus of the lithotripter after passage of each shock wave. The dynamics and form of a cavitation cluster depend on the initial sizes of the cavitation nuclei, the properties of the applied tensile stress, and the character of the surrounding materials. The sudden growth and violent collapse of a cavitation bubble is referred to as transient cavitation. Stable cavitation does not occur in ESWL, and is associated with the oscillation of a gas bubble in an acoustic field (Leighton 1997). Lithotripter-generated bubbles generally expand in about 50 to 100 \( \mu \text{s} \) after shock wave passage, stabilize, and collapse violently after approximately 250 to 500 \( \mu \text{s} \) (Kodama & Tomita 2000; Evan et al. 2002; Bailey et al. 2005). Cavitation bubbles created inside ESWL devices have also been observed to collapse as soon as 100 \( \mu \text{s} \) after
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Shock Wave Physics for Urologists

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VI Shock wave interaction with living tissue and cells

The influence of shock waves on living tissue was first recognized during World War I, when the lungs of castaways were damaged because of underwater explosions. Surprisingly, this could happen without producing damage to other parts of the body. Both World Wars stimulated the study of shock wave interaction with the human body.

In general, acoustic waves can cause various effects as they propagate through a medium. Common examples are heating, structural changes, cavitation, compression, rarefaction, reflection, and turbulence. Heating is the result of acoustic energy absorption, i.e., acoustic energy is transformed into heat within the material (tissue). It depends on the physical properties of the material, the intensity of the acoustic radiation, its duration, and its frequency. Dynamic agitation and shear stress produced by the waves can affect structural properties. Acoustic cavitation is also often observed. Cavitation may produce free radicals. In addition, sound waves cause rapid compressions and expansions of the tissue. Reflection occurs as the wave hits an interface (change in acoustic impedance). If the difference in acoustic impedance between two media at an interface is large, the amount of energy reflected will also be large. Finally, at liquid/solid or at gas/solid interfaces, turbulence or “acoustic streaming” may occur.

During the 1980s, the popularity of ESWL increased enormously because of the premature conclusion that no tissue damage or adverse long-term effects were produced after shock wave application. We now know that unfortunately, this is not always true. Lithotripter shock waves may damage tissue and produce long-term effects. They should be administered with care by a well-trained and experienced physician.

Initial experiments by Chaussy and Eisenberger (Webb, Payne & Wickham 1986; Denstedt, Clayman & Picus 1989) revealed petechial bleeding and damage to the intestine of rats after shock wave application; however, no damage to stone-containing canine kidneys was seen after ESWL using the Dornier HM1 at 12 kV (500 shock waves). Although these initial studies concluded that ESWL had no damaging effect on renal tissue, later it became clear that shock waves may cause tissue trauma, consisting primarily of localized destruction of vascular walls, formation of venous thrombi, and generation of diffuse hemorrhages and hematomas (Mayer et al. 1990; Evan et al. 1991; Evan et al. 1998a; Evan et al. 1998b; Willis et al. 1999; Evan, Willis & Lingeman 2003).

Several authors have described shock wave induced tissue trauma in different animal models (Thibault et al. 1986; Delius, Enders & Heine 1987; Delius, Enders & Heine 1988; Abrahams, Lipson & Ross 1988; Jaeger et al. 1988; Neisius et al. 1989; Recker, Ruebben & Bex 1989; El-Damanhoury et al. 1991; Ryan et al. 1991; Kaji et al. 1991; Raeman et al. 1994; Anderson et al. 1995); however, the determining factors of tissue lesions have not been completely clarified.

Shock waves may lacerate, tear, compress, and rip through the walls of veins located near the focus of a lithotripter. Shear, due to the negative peak of the shock wave, tears the tissue. Cavitation can also lead
to disruption. Furthermore, when shock waves pass through soft human tissue, material flows are induced behind the shock front (Kodama & Tomita 2000). Injury to the skin may also occur at the point where shock waves enter the body, particularly if coupling of shock waves into the patient was not carefully verified. Additionally, interstitial edema, transient macrohematuria, and temporary loss of kidney function have been observed after ESWL (Woodruff & Kandel 1987; Fischer et al. 1988; Evan & Willis 2007). Hematuria is believed to be a result of cortical and medullary hemorrhage, tubular dilation, and glomerular bleeding, and it typically disappears within one day. The significance of some of these lesions is still debated. Organs adjacent to the kidney are usually unaffected (Abrahams, Lipson & Ross 1988; Hill, McDougal & Stephens 1990). Magnetic resonance imaging and computerized tomography have shown intrarenal and subcapsular hematoma after ESWL in 24 to 85% of all patients (Kaude et al. 1985; Baumgartner et al. 1987; Rubin et al. 1987; Littleton, Melser & Kupin 1989; Evan et al. 1991).

It has been reported that renal trauma due to lithotripter shock waves could be more dangerous for elderly patients, patients with hypertension, and children (Janetschek et al. 1997; Lifshitz et al. 1998; Willis et al. 1999). Nevertheless, long-term effects on kidney function and blood pressure have still to be elucidated in detail. An increased risk to develop diabetes mellitus has also been reported. In a case-control study, treating renal stones with a Dornier HM3 lithotripter was associated with diabetes mellitus and hypertension at 19 years of followup. Hypertension correlated with bilateral ESWL treatment. Diabetes mellitus was related to the number of shock waves and the treatment intensity (Krambeck et al. 2006). A relationship between the onset of brushite stones and multiple ESWL sessions was reported by Parks et al. (2004).

During ESWL it is important to keep in mind that both in front of and behind the stone, a large quantity of unneeded shock wave energy is always deposited into the tissue (Figure 66); however, shock waves can hit tissue over a rather broad area, not just in front and behind. The energy deposited outside the stone (or stone fragments) may cause tissue damage, which makes it important to stop ESWL on time. This situation may vary from one lithotripter to another, but it cannot be eliminated. All lithotripters produce cigar-shaped focal zones. A spherical focal zone would require shock wave access from all sides outside the patient’s body, which obviously is impossible. ESWL tissue trauma can be minimized if the focal zone of the lithotripter is targeted on the stone, so that the energy applied reaches therapeutic values only in the vicinity of the stone. Passage of shock waves through the lungs would lead to serious damage, because shock waves would be almost entirely reflected and develop destructive forces at tissue-air boundaries.

![Fig. 66](https://example.com/fig66.png) Schematic of shock wave energy concentration showing that in front of and behind the kidney stone, unneeded shock wave energy is deposited into renal tissue.
Experimental results suggest that shock wave number is a risk factor that limits the extent to which ESWL may be used (Janetsheck et al. 1997; Willis et al. 2005). Unfortunately, patients having fragile calculi are often over treated because of the general practice of applying the standard protocol, that is, using the high voltage settings and the full number of shock waves that guarantee fragmentation. Fortunately, advances in computerized tomography will make it possible to identify fragile stones and avoid over treatment in the near future (see Section VIII.6). According to Rassweiler et al. (1993), lesions to the renal vasculature depend mainly on the voltage applied. High voltage caused rupture of interlobular arteries even after only 25 shock waves. This was not seen at low energy settings after up to 2500 shock waves. A correlation between peak pressure and renal trauma in the pig has been reported by different authors (El-Damanhoury et al. 1991; Rassweiler et al. 1993). Increasing peak pressure from about 40 to 60 MPa resulted in exacerbated kidney hemorrhage.

Bones cannot be considered brittle structures, because their tensile and compression strength are similar. Nevertheless, since the acoustic impedance of bone is about four times higher than that of soft tissue, shock wave reflection at the bone/tissue interface may generate tensile waves within the bone, which can exceed its strength. Because of the high percentage of collagen matrix and crystalline composition, bone is not affected too much by shock waves. Karpman and coworkers (1987) demonstrated that lithotripter shock waves have minimal effect on bone strength. No fractures of bones have been reported during ESWL (Weinstein, Wroble & Loening 1986).

As explained in Chapter III, the pressure pulse in the focal region of a lithotripter consists of a compression pulse with a pressure between 30 and 150 MPa, and a subsequent decompression pulse with a negative peak of up to -20 MPa. Since tensile strength for human renal parenchyma is only about -0.57 MPa, theoretically, tissue rupture could occur near the stone during ESWL. New generation lithotripters with small focal zones and high treatment pressures resulted in higher hematoma rates than first generation devices with larger focal volumes and lower peak pressure (Köhrmann et al. 1995; Piper et al. 2001).

As the wave propagates through tissue, energy is lost due to friction, causing energy absorption, and pressure and intensity reduction. The pressure reduction can be estimated by

\[ p = p_0 e^{-\alpha d}, \]

where \(d\) stands for depth in centimeters, \(p_0\) is the initial pressure, and \(\alpha\) is the absorption coefficient. Since in a biological medium the absorption coefficient equals \(\alpha = bf^m\) (\(b\) and \(m\) are constants, \(f\) is the frequency), high-frequency waves will be absorbed more than low-frequency waves. Sonic absorption in biological tissue is relatively low. Most biological tissues have a value of \(m\) between 1 and 2. An advantage of shock waves is that they have a higher penetration power, because their frequency spectrum includes lower frequencies than ultrasound. Nevertheless, high frequencies composing the sharp portion of the wave front are attenuated more than the low-frequency components associated with the tensile phase, leading to a reduction in the positive peak.

Traumatic effects also depend on the frequency. Low-frequency components produce larger displacement amplitudes compared to the high frequency domain. Therapeutically effective components for ESWL are considered to be above 200 kHz. Components with no effect are between 20 and 200 kHz.

High-intensity ultrasound can create localized heating in tissue, and it is used to cauterize vessels or to necrose tumors. Fortunately, contrary to ultrasound, shock waves transfer virtually no heat, i.e., thermal injury to kidney tissue can be ignored at ESWL exposure levels (Filipczynsky & Wojcik 1991).
Cavitation is known to be a main tissue-damage mechanism during ESWL (Coleman & Saunders 1993; Coleman, Choi & Saunders 1996; Delius, Ueberle & Eisenmenger 1998; Carstensen, Gracewski & Dalecki 2000; Sapozhnikov et al. 2001; Zhong, Zhou & Zhu 2001; Evan et al. 2002). A proof of this is that the lysis of red blood cells in suspension has been reduced by application of excess hydrostatic pressure (Williams et al. 2002). At high hydrostatic pressure, bubbles cannot expand freely, reducing cavitation. To minimize damage due to cavitation, it is desirable that the negative component of the pressure profile along the shock wave path is small. In vitro, cavitation bubbles can expand up to about 3 mm in diameter. This is larger than most of the fluid filled spaces in human tissue, suggesting that bubble expansion in vivo is constrained by the surrounding tissue. Ultrasound B-scan imaging shows the formation of cavitation bubbles during ESWL produced by the negative pressure of the shock waves. Bubble formation has been detected in the renal parenchyma during ESWL. Bailey and coworkers (2005) reported that in urine, cavitation occurred readily, but bubble activity in tissue was detected only after continuous treatment with hundreds of shock waves. As explained in Section V.2, tensile components of the shock waves converging towards the focal region may exceed the threshold for cavitation in urine, blood, or tissue, thereby inducing bubbles. When these bubbles collapse, they can produce secondary shock waves and high-speed liquid microjets. Nevertheless, it is important to mention that cavitation is different in tissue than in water or urine. Fortunately, most mammalian tissue is normally remarkably free of cavitation nuclei. When the size of cavities in tissue is small, either no liquid microjets are formed, or they do not become a dominant factor for causing injury. Furthermore, the rise time of a shock wave is significantly increased as it passes through tissue. This affects growth and collapse of cavitation bubbles. Williams and coworkers (1999) studied the effect in vitro of macroscopic air bubbles on cell lysis by shock wave lithotripsy. The possibility of controlling cavitation activity to limit bubble formation near the kidney stone is a main research topic (Arora, Junge & Ohl 2005).

Ultrasound contrast agents are cavitation nuclei. Injection of ultrasound contrast agents like Albunex® (Molecular Biosystems Inc., San Diego CA) or Levovist® (Schering AG, Berlin, Germany) is not recommended before or during ESWL, because vascular injury could be increased (Dalecki et al. 1997; Matlaga et al. 2006). It seems logical that small blood vessels are at higher risk for mechanical rupture than large blood vessels. This has been confirmed experimentally (Zhong, Zhou & Zhu 2001). The size of microbubbles (shells of denatured albumen) in commercial ultrasound contrast agents is about 1 - 10 µm. At lithotripter pressures, cavitation can be induced in blood vessels having nuclei as small as 20 nm in diameter (Zhong et al. 1998).

Contraction of muscles seen during ex vivo shock wave treatment showed that shock waves excite nerves (Schelling et al. 1994). For this to happen, it was required that small gas bubbles were present in the organ bath where the nerve was immersed, demonstrating that the effects were due to cavitation. Under certain circumstances, shock waves may increase the permeability of cell membranes, allowing large molecules to become trapped inside them without causing cell death. This type of interaction of shock waves with cells is discussed in Section VIII.8.

Results on artificial stone fragmentation cannot be directly correlated to trauma observed in animal models. Extensive experimental research is needed to determine the therapeutic range of any new lithotripter, and basic research is necessary to better understand shock wave interaction with biological tissue and to define techniques that improve treatment efficiency while minimizing tissue trauma. Changes in shock wave delivery rate, power settings, and shock wave
Unlike research lithotripters (Coleman, Saunders & Choi 1989; Prieto, Loske & Yarger 1991; Cleveland et al. 2000), most lithotripters for clinical practice only allow control over few parameters, like the number of shock waves, the voltage, and the rate of shock wave administration. Nevertheless, well-defined protocols for ESWL should be developed in any lithotripsy center to guarantee good results and prevent errors and over-treating. Extensive training by certified technicians and urologists should be mandatory before using any extracorporeal lithotripter. Unfortunately in many hospitals the importance of having a team specialized on ESWL is underestimated. This chapter focuses on a few recommendations given from the standpoint of physics; however, it is the responsibility of the urologist to evaluate the whole clinical scene and decide when to take advantage of them.

VII.1 Choosing a lithotripter

All lithotripters differ in design, performance, price, reliability, maintenance costs, fragmentation efficiency, and potential to cause tissue damage. Improvements have been made with regard to functionality, size, radiation exposure, running costs, anesthesia requirements, and ergonomic aspects; however, clinical results did not improve as expected (Parr, Pye & Ritchie 1992; Lingeman 2007). So far, few lithotripters have achieved the reported stone-free rate and minimal re-treatment rate of the Dornier HM3 lithotripter. The HM3 established the criteria for newer devices. Advantages of most new lithotripters are elegant energy sources, and improved imaging systems. Furthermore, patients can be treated in a variety of positions. Nevertheless, reports of high re-treatment rates are common (Wilson & Preminger 1990; Rassweiler & Alken 1990). This may be due to a poor understanding of the mechanics of stone-fragmentation and tissue-injuring processes. Fortunately, research groups worldwide have recently proposed possible improvements, and new shock wave technologies will be developed to obtain the highest efficacy and patient comfort with the lowest re-treatment rates.

VII.1.1 The “best” lithotripter

The ideal lithotripter should be cheap, easy to use, have good quality, low maintenance costs, and provide maximal fragmentation efficiency with minimal associated trauma. However, as in most clinical procedures, good results will only be achieved with a well-trained team. This requires time and investment. In ESWL the skill and experience of the lithotripter operator is much more important than normally believed. Unfortunately, in many hospitals, operators are switched too frequently and do not always have sufficient training and experience. First class modern lithotripters may result in inefficient and even dangerous devices in hands of inexperienced urologists or technicians.

When choosing a lithotripter, important considerations are the budget, the available facilities, and the patient population. It is not the purpose of this book to compare commercial lithotripters in order to judge which lithotripter is “the best”. Some interesting characteristics of a few commercial lithotripters are
described; however, there are excellent lithotripters on the market that are not mentioned in this book.

During the last ten years, international meetings have been an important “field of battle” between lithotripter manufacturers. New shock wave generators are presented and in vitro results are published before the systems are on the market. Brief reports on initial ESWL treatments using new lithotripters appear to be promising in almost all cases (Zaman et al. 2005; Bergsdorf, Thueroff & Chaussy 2005; Rajan, Pemberton & Tolley 2005; Goren et al. 2005; Tailly 2005; De Marco et al. 2005); however, direct comparison between lithotripter performance reports is generally senseless, because different methods and definitions for “success,” “efficiency”, or “efficacy” are used. Some companies define efficiency as “less expense per patient achieving medical success with minor side effects.” Other manufacturers and research institutions define efficiency as “stone fragmentation efficiency.” Technical characteristics and clinical results can be presented in such a way that almost any lithotripter could appear to be better than another one.

There are several articles reporting the efficacy of lithotripters; however, most are longitudinal trials making it meaningless to compare the instruments directly. Clinical studies have variable practice patterns (Rassweiler et al. 1987; Taillery 1990; Tan, Tung & Foo 1991; Sofras et al. 1991; Bierkens et al. 1992; Cass 1996). Nevertheless, some comparative results have been reported (Schultz-Lampel & Lampel 2001; Netto et al. 2002). Teichman and coworkers (2000) performed a study to compare overall success in reducing calcium hydrogen phosphate dihydrate, calcium oxalate monohydrate, cystine, and magnesium ammonium phosphate hexahydrate stones to fragments smaller than 2 mm, using seven different lithotripters. In this study the Dornier HM3, the Storz Modulith SLX, and the Siemens Lithostar 2 gave excellent results.

Comparing lithotripters requires not only one, but several parameters. Some manufacturers only report the “most convenient” parameter for their machine; however, one single parameter means little. Results depend on the type, number, and location of stones, as well as on the number of applied shock waves, the pressure profile, the discharge frequency, the type of anesthesia, the auxiliary measures, etc. From the physical standpoint, there are important recommendations for selecting a suitable lithotripter. A few of them will be discussed in the following sections.

VII.1.2 Pressure field and electric energy
As already mentioned in this book, high peak positive pressure does not necessarily correlate with enhanced fragmentation (Chuong, Zhong & Preminger 1992; Teichman et al. 2000). It is common that lithotripter companies publish peak positive pressure values measured by in-house testing without mentioning the type of transducer and methods used. This data gives no useful information on the potential of the lithotripter and should not be used for comparison unless these measurements have been performed under the same experimental conditions, using the same transducers and techniques (see Section VIII.3). Furthermore, to compare electrical energy only in terms of voltage is not relevant if the capacitance of the shock wave generator is not known. Moreover, high electrical energy does not necessarily achieve high fragmentation power. Parameters like rise time of the shock wave, peak positive and negative amplitude, and shape of the focal volume may be more important than maximum generator voltage and capacitance. Furthermore, electric energy should not be compared between lithotripters using different shock wave generating principles. Smaller, easier to use, and cheaper lithotripters, capable of producing extremely high peak pressure have been relatively easy to sell; however, as already mentioned, the clinical results are not encouraging.
It is important to realize that the focal-zone definition commonly used in ESWL depends on the maximum peak positive pressure ($p^+$) and does not necessarily represent the volume of “clinical efficiency”. This definition is useful to a certain extent; however, a focal zone based on the maximum negative peak pressure ($p^-$) could also have been defined. A $p^-$-based focal zone does not have the same size and position as the “positive” focal zone. Since negative pressure is responsible for cavitation-induced damage to the stone (see Section V.2), the “negative” focal zone could be a better indicator of lithotripter performance. In vitro and in vivo studies with electrohydraulic lithotripters have shown that fragmentation can be enhanced if the stone is positioned several millimeters in front of the geometrical focus of the lithotripter (see Section VIII.4). This is believed to happen because the maximum peak negative pressure is not located at F2, but closer to F1.

Some authors prefer to use the 5 MPa focus or treatment zone definition. Any point inside this volume has a pressure equal to or larger than 5 MPa. The treatment zone is defined as the volume where we can expect stone fragmentation. This volume may be larger or smaller than the focal zone, depending on the energy level selected. The focal zone does not necessarily change its size as the energy varies; however, the size of the treatment zone always depends on the energy of the shock wave. The treatment zone of a lithotripter operated at 18 kV is different in size and shape from the treatment zone of the same lithotripter at 21 kV. The advantage of this definition is that the treatment zone is related to the absolute amount of energy deposited into the tissue (and stone).

VII.1.3 Fragmentation efficiency
An in vitro comparison of ESWL machines (Teichman et al. 2000) revealed that extracorporeal lithotripters vary in the ability to reduce stones to small fragments. Chitnis (2002) reported that an electromagnetic lithotripter was more efficient in breaking artificial stones than an electrohydraulic device; however, some clinical literature indicates the opposite. Comparing the efficacy of spark gap versus electromagnetic shock wave lithotripters, Sebesta and Bishoff (2003) found that an electrohydraulic system was more effective in treating urolithiasis than an electromagnetic machine. In a single center, retrospective study comparing ESWL outcomes using a piezoelectric, an electrohydraulic, and an electromagnetic lithotripter, Ng and coworkers (2003) found that the electrohydraulic lithotripter had the best outcomes for stones of the lower calyx.

Because of their varying chemical composition, fragmentation mechanisms, and mechanical properties, in most cases human calculi are not suitable for in vitro fragmentation tests. Well-standardized kidney stone models, like HMT (High Medical Technologies, Kreuzlingen, Switzerland) or U-30 (van Cauwelaert 2004; McAteer et al. 2005b) stone phantoms, placed in a water tank at the focus of the lithotripter are a good way to compare fragmentation efficiency (see Figure 67). Artificial stones provide reproducible results and can be used to compare the effectiveness of various lithotripters. A stone model using natural stone materials has also been developed for this purpose (Heimbach et al. 2000). In vitro fragmentation produces stone breakage equivalent to that seen in vivo (Paterson et al. 2002); however, to evaluate a...
lithotripter, other aspects like tissue injury should be considered (Evan & Willis 2007). For in vitro experiments to test fragmentation efficiency of lithotripters, stone comminution generally is determined by the percentage of fragments smaller than 2 mm. The fragmentation coefficient FC of a lithotripter is defined as:

$$FC = \frac{(W_i - W_f)}{W_i} \times 100 / W_i.$$ 

In this equation, $W_i$ and $W_f$ stand for initial (intact) stone and final stone (fragment) weight.

**VII.1.4 The shock wave generator**

The shock wave generator is the most important part of a lithotripter. It determines running costs, fragmentation efficiency, potential tissue damage, and anesthesia needs. The first challenge before buying a new lithotripter is to choose between three shock wave generation principles: electrohydraulic, piezoelectric, and electromagnetic. Some clinical studies have shown significant differences between electrohydraulic, piezoelectric, and electromagnetic lithotripters. Size of the focal zone and type of lithotripter may influence induced renal damage (Neisius et al. 1989; Morris et al. 1989; El-Damanhoury et al. 1991; Morris et al. 1991).

For many years, there has been a discussion amongst manufacturers and scientists, whether a larger or a smaller lithotripter focal zone is better in terms of fragmentation and side effects. Electrohydraulic lithotripters have large focal zones. Piezoelectric systems have small focal zones. Most piezoelectric machines are intermediate between electrohydraulic and electromagnetic lithotripters in peak pressure (see Figure 68). Some companies have increased the pressure amplitude and reduced the size of the focal zone in attempts to reduce the number of shock waves and tissue damage (Moody, Evan & Lingeman 2001). On the one hand, selecting a shock wave source with large aperture (small focal zone) is desirable because cavitation threshold values may only be exceeded close to the focal zone (see Chapter VI). However, with these lithotripters, good imaging systems and precise patient positioning, as well as an experienced staff are essential to achieve good results, because small focal zones leave little margin of error for targeting the stone. If the focal spot is small, the stone will spend much less time within this volume. In this case, an automatic stone-tracking system would be useful. Patient movement and deep breathing may reduce the success rate significantly (see Figure 69). During ESWL, respiratory motion causes stone displacement of up to 50 mm. Piezoelectric lithotripters have a greater probability of missing the stone during treatment.

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**Fig. 68** The three types of shock wave sources produce different focal zones and different peak pressure. Pressure values and focal-zone dimensions are only examples for each category. Not all lithotripters of the same type have focal zones with the same dimensions.

**Fig. 69** Respiratory motion causes stone displacement. Large apertures (small focal size) reduce the time the stone will spend within the treatment zone.
VIII Research

Since the development of ESWL, countless patients have been successfully treated worldwide, and the number and models of commercial lithotripters have increased significantly. In the early days of ESWL, progress was achieved more by obtaining practical experience rather than by understanding the detailed physical mechanisms causing stone comminution and tissue injury. Research has been done in many countries in order to find improvements in technology as well as modifications in clinical practice that might increase fragmentation efficiency while reducing renal injury. Nevertheless, 27 years after the first ESWL, several publications demonstrate that lithotripsy is not getting better. Complications like long-term hypertension (Knapp et al. 1995; Knapp et al. 1996) and renal trauma (see Chapter VI) indicate that basic and clinical research is necessary. In about 85% of all ESWL treatments, morphologic changes are observed in the kidney (Köhramm et al. 1994). For this reason, today there are more research groups working on the application of shock waves to medicine than during the development of ESWL, indicating that this is still a promising research field.

A chapter on research on such a vast area as ESWL, which includes so many different areas of science, will never be exhaustive. Furthermore, advances in technology are so fast that no manuscript dealing with research on medical applications can be up to date. This part of the book is only a small selection of topics currently being studied in some universities.

VIII.1 Research extracorporeal lithotripters

Even though lithotripters developed for clinical applications can be used for research purposes, it is not easy to find spare time in hospital schedules for non-clinical research and, on financial grounds, sometimes it is difficult to justify the use of expensive ESWL devices. Furthermore, commercial lithotripters have few adjustable parameters and are not always suitable for experimental studies. The design of experiments can be improved by the use of specially designed research devices. Effects like those related to the propagation of the shock wave beam in water and tissue can be more easily studied under well-controlled laboratory conditions. For these reasons some universities have decided to build their own research lithotripters (Coleman, Saunders & Choi 1989; Prieto, Loske & Yarger 1991; Cleveland et al. 2000; Loske et al. 2003).

Fig. 84 Photograph of a small test tank, fastened to the shock wave generator of a Dornier Compact Sigma lithotripter. This system gives a functional assessment of alignment and operation of the lithotripter. Artificial kidney stones, placed on the mesh inside the test tank may be used to demonstrate fragmentation. (Photograph courtesy of Elí Arellano, Centro Médico Puerta de Hierro, Guadalajara, Jalisco, México).
Most of these devices were designed to reproduce the pressure waveform generated by the Dornier HM3. Some companies also donate shock wave generators to research laboratories; however, this is generally done in order to test their systems. Small water tanks with flexible rubber membranes are sometimes fastened to shock wave heads of lithotripters in order to perform pressure measurements, stone fragmentation tests, and other analyses (see Figure 84).

VIII.1.1 The UK research lithotripter
One of the first experimental shock wave generators for lithotripsy studies was developed at the Medical Physics Department of the St Thomas’ Hospital in London. The device was designed to simulate the acoustic field generated by a Dornier HM3 lithotripter so that experimental and clinical results could be compared. The shock wave was focused by a solid brass ellipsoidal reflector with the major and minor axes of the Dornier HM3 and a focal length of 115 mm. The focal length was slightly shorter than that used on the HM3, in which the reflector is truncated to allow space for the x-ray beams of the fluoroscopic imaging system. The spark plug and its orientation inside the reflector were the same as those used on the HM3. A photograph of the device is shown in Figure 85. The capacitor and the discharge circuit were contained in a perspex box positioned above the reflector. The reflector was mounted at one end of a water tank. Pressure transducers could be suspended from a mounting on an optical bench. A low-power laser, also mounted on the bench, indicated the axis of the reflector. Details on the design and evaluation of this small, research lithotripter have been published (Coleman, Saunders & Choi 1989).

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**Fig. 85** Photograph of the experimental shock wave generator for lithotripsy studies developed at the Medical Physics Department of the St Thomas’ Hospital, London UK, in 1989 (Coleman, Saunders & Choi 1989).
VIII.1.2 MEXILIT research lithotripter

The MEXILIT (see Figure 86) designed at the Shock Wave Laboratory of the Center of Applied Physics and Advanced Technology at the National Autonomous University of Mexico (UNAM) consists of a pulsed power circuit, providing multiple pulses to a spark gap immersed in water (Prieto, Loske & Yarger 1991). The total capacitance of the circuit can be varied by installing between one and ten cylindrical 10-nF capacitors between two circular brass discs. A trigger switch is mounted above the set of capacitors (Figure 87). The spark gap electrode assembly is at the focal point of a stainless steel reflector, mounted on the bottom of a 1200 × 800 × 600 mm fiberglass water tank. Application of high voltage (up to 30 kV) to the electrodes generates a shock wave that propagates into the surrounding water and reflects off the reflector. A manual and a computer-controlled positioning system are placed on top of the device (Figure 86) in order to fasten and move any probe, pressure transducer, test tube, or sample to any desired position within the tank. Basically, the electric circuit consists of a computer-controlled capacitor charging system and a discharge device. Electrodes, with the shape of a truncated cone (Loske & Prieto 1993), are burned in for about 200 discharges at 18 kV before starting any experiment (Figure 88). Either tap or degassed water is used. The

Fig. 87 Trigger switch and set of capacitors to store electric energy in a research electrohydraulic lithotripter. (Photograph: G. Trucco & F. Fernández).

Fig. 88 Spark plug used in a research lithotripter (Loske & Prieto 1993). (Photograph: F. Fernández).

Fig. 89 Set of stainless steel reflectors with different geometries used to concentrate shock waves generated in a research lithotripter. (Photograph: A. Sánchez).

Fig. 90 Ellipsoidal stainless steel reflector with lasers to show the location of F2 in a research lithotripter. (Photograph: A. Sánchez).
MEXILIT is similar to electrohydraulic shock wave generators used in ESWL. An advantage for research purposes is that there is a set of different reflectors (see Figure 89), including a parabolic reflector that can be used to apply shock waves to vials containing cell suspensions (see Section VIII.7). Three diode lasers fastened to adjustable mountings at the periphery of the reflectors are used to locate the focus of the ellipsoidal reflectors (Figure 90).

VIII.1.3 The US research lithotripters
A group of specialists from Boston University, University of Washington, Indiana University School of Medicine, California Institute of Technology, and H-Tech Laboratories designed an electrohydraulic research lithotripter as part of a multi-institutional United States research program to investigate mechanisms of stone comminution and tissue damage during ESWL (Cleveland et al. 2000). Three research lithotripters were constructed, individually tailored to the needs of each group. All lithotripters provide performance equivalent to the Dornier HM3. Two capacitors (40 nF each) and a spark gap trigger are enclosed in an acrylic housing. The reflector, bolted to the bottom of a water tank, has the same dimensions as that of the unmodified HM3 lithotripter. Three different rectangular test tanks were made to the specifications of each research group. A water degassing system was designed for each tank. One of these research lithotripters is shown in Figure 91.

VIII.1.4 HM3 research lithotripter
Hospitals around the world have been replacing their first generation machines with new lithotripters. Therefore, many HM3 lithotripters are now in warehouses and basements of hospitals; however, their shock wave generator may still be useful for research. This generator is especially valuable since the majority of animal and human studies reported so far have utilized the HM3, known as ‘The Gold Standard’. A few representative examples are cited at the end of this book (Graff, Schmidt & Pastor 1987; Whelan & Finlayson 1988; Jaeger et al. 1988; van Arsdalen et al. 1991; Miller, Thomas & Thrall 1996; Willis et al. 1996; Cleveland et al. 1997; Lauer et al. 1997).

The sketch of a modified HM3 lithotripter is shown in Figure 92 (Loske et al. 2003). Due to its large dimensions, the original stainless steel tub was replaced by a small rectangular Lucite water tank fastened to the frame of the original lithotripter. The patient gantry and the fluoroscopy system were not installed (see Figure 93). Pressure measurements performed near the focus of the research lithotripter were compared with data.
Shock Wave Physics for Urologists

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While ESWL techniques are still evolving and improvements that increase its efficiency to disintegrate urinary stones and reduce pain are constantly sought, different applications of shock waves lie in an increasing variety of medical indications, especially in orthopedics. ESWL changed the management of urinary calculi. This inspired the use of shock waves as an alternative to treat stones in the gallbladder, common bile duct, pancreatic duct, and salivary gland ducts. It is a matter of general education for the urologist to know about other uses of shock waves in medicine.

IX.1 Non-urological ESWL

Shock waves were first used in a patient with gallbladder stones in 1985 and became an alternative in the treatment of gallbladder stones, pancreatic concrements, and stones of the salivary gland. ESWL can be performed safely on an outpatient basis. Tissue damage appears to be temporary, and side effects, such as biliary pancreatitis and liver haematoma, were found to be rare. Unfortunately gallbladder stone recurrence is high. Because of this, laparoscopic cholecystectomy has become the standard treatment of gallbladder stones today, and ESWL is used to treat bile duct stones resistant to endoscopic extraction. Furthermore, as mentioned in earlier sections of the book, ESWL is not as efficient for treating gallstones as in urology. This is because of the physical properties of gallbladder stones. In general gallstone fragments are difficult to eliminate. They either drain into the intestine on their own or have to be extracted endoscopically (Bland et al. 1989; Wenzel et al. 1989; Chapman et al. 1989; Sackmann 1992; Nahrwold 1993; Chang & Pamies 1994).

ESWL is useful in patients where surgery is contraindicated, and it has shown good results for the treatment of small solitary gallbladder stones, but a low success rate in the case of multiple stones (Rabenstein et al. 2005). Sackmann and colleagues (2001) reported that in bile duct calculi that are difficult to extract endoscopically, ESWL is effective regardless of stone size. It should be considered as a treatment, especially in high-risk patients.

ESWL should also be considered as a useful tool to treat pancreatic duct calculi, since it allows fragmentation of stones and facilitates spontaneous stone passage or endoscopic removal. It has been reported to achieve complete duct clearance in up to 50%, and improvement in duct decompression and symptoms in up to 70% of patients (van der Hul et al. 1993; van der Hul et al. 1994; Kim 2005; Choi et al. 2005; Tadenuma et al. 2005; Conigliaro et al. 2006).

A common pathology of the salivary gland is sialolithiasis. Most frequently localized in the submandibular gland, stones also occur in the parotid gland. Before ESWL, the gland had to be removed to get rid of the stones, in spite of the associated risks to adjacent structures, especially the facial nerve. ESWL for salivary gland stones was introduced in 1989. It can be performed on an outpatient basis. The relatively painless treatment and the elimination of the need for an operation with its surgical risks are advantages of ESWL, so it provides a useful option, particularly for stones less
than 7 mm in diameter (Iro et al. 1991; Hessling et al. 1993; Fokas et al. 2002; Escudier et al. 2003; Zenk et al. 2004; McGurk, Escudier & Brown 2005). Due to the large size of the focal zone, electrohydraulic lithotripters are not convenient for sialolithiasis (Bayar et al. 2002); however, electromagnetic and piezoelectric lithotripters are suitable for this type of treatment (Capaccio et al. 2004). The procedure usually consists of hundreds of shock waves. Earplugs are applied to patients treated for parotid gland stones. Anesthesia is normally not required. About 60% of the patients with parotid gland stones or submandibular gland stones obtain either total stone clearance or sufficient fragmentation to permit spontaneous passage (Kater et al. 1994).

**IX.2 Orthopedic application of shock waves**

As already mentioned in Chapter VI, bones are fairly resistant to shock waves because of their crystalline composition and high percentage of the collagen matrix. Since their tensile and compression strength do not differ much from each other, they cannot be considered brittle. Nevertheless, *acoustic impedance* of bone is higher than that of soft tissue. As a consequence of this, shock wave reflection generates tensile forces that might induce trabecular fractures.

The use of shock waves in orthopedics, named extracorporeal shock wave therapy (ESWT) or extracorporeal shock wave application (ESWA) was introduced in Germany in 1989. Modified lithotripters and specially designed shock wave generators for orthopedics have been developed by several companies (Loske & Prieto 1999). Most of these machines have therapy heads suspended on articulated arms with unrestricted three-dimensional manual movement and in-line ultrasound. These devices can also be complemented with a mobile C-arm to perform x-ray localization. Shock waves are sent directly to the area of pain. Treatments generally take about 30 to 45 minutes and are performed under local anesthesia as out-patient procedures. Energy densities range between about 0.004 and 0.6 mJ/mm².

The mechanics of ESWT are not yet fully understood. Shock waves are used mainly: a) to perform closed osteotomies or to stimulate local hematoma formation and bone healing at the site of non-union (Valchanou & Michailow 1991; Haupt et al. 1992; Schleberger & Senge 1992), b) to fragment polymethylmethacrylate (PMMA) cement and separate it from bone (Braun et al. 1992), and c) to bring relief to patients suffering from chronic pain syndrome (Rompe et al. 1995; Loew, Jurgowski & Thomsen 1995; Haupt 1997; Zimmermann et al. 2005). For the first two applications, energies as in ESWL are used. The treatment of pain requires less energy.

Treatments of patients with delayed union or non-union of fractures were first reported by Valchanou and Michailow (1991) and Schleberger and Senge (1992). Application of several hundred shock waves induces bony union in about 85% of the patients. Microtrauma is produced by the shock waves, leading to revascularization, which triggers the healing process. The main lesion from shock waves focused to the bone is hemorrhage at the periosteum and in the bone marrow. After shock wave treatment the bone reacts by an apposition of new bone at the site of the lesion.

Shock waves have been used to loosen bone cement during revision arthroplasty (Braun et al. 1992). This is possible because there is no chemical bonding between bone and cement and since the *acoustic impedance* of bone is about twice as large as the impedance of bone cement. At the bone/cement interface, conversion of positive pressure into tensile pulses leads to disintegration of the cement. Weinstein and coworkers proposed a preoperative shock wave application to loosened cemented arthroplasties (Weinstein, Wroble & Loening 1986); however, shock waves may liberate bone marrow particles, which could cause fat embolism (Braun et al. 1992).
ESWT to treat some chronic musculoskeletal diseases received the approval of the Food and Drug Administration of the USA (FDA) in 2001. Candidates for shock wave treatment are patients in whom the pain has lasted for several months without responding to conservative therapies such as massage, exercises, cortisone injections, and anti-inflammatory medications. About two million cases of chronic heel pain syndrome (plantar fasciitis) are reported in the United States each year. Fortunately, ESWT has proven to be very effective not only in treating plantar fasciitis, but also shoulder calcifications (calcific tendonitis), and Achilles tendonitis. ESWT therapy has also become important in sports medicine, including tennis elbow (epicondylitis humeri radialis) and golfers’ elbow (medial epicondylitis).

To treat the elbow, three treatments, each consisting of approximately one thousand shock waves having an energy density of about 0.06 mJ/mm² are successful in about 90% of patients (Rompe et al. 1995). This is a very low energy flux density compared to ESWL, where densities in the 0.2 to 2 mJ/mm² range are common. For more than ten years the tendons in the shoulder have been freed from calcareous inclusions using shock wave therapy (Loew, Jurgowski & Thomsen 1995). In the case of plantar fasciitis, shock waves are directed at the plantar fascia, reducing inflammation and pain from the affected ligament; however, the benefits may take up to three months to be fully effective.

Shock waves seem to trigger the patient’s natural healing ability, stimulating metabolic response and revascularization of tissue, changing the membrane permeability and causing the development of stress fibers. They also create cavitation bubbles, which break calcific deposits and induce an analgesic effect by stimulating the axons. As a result, the patient’s pain threshold is increased. Discomfort during or up to 24 hours after treatment may be expected; however, there are no known adverse side effects with the use of ESWT.

Pneumatically generated shock waves are also used to reduce pain. In the Masterplus MP 100 (Storz Medical, Kreuzlingen, Switzerland), a pneumatic pulse accelerates a projectile located inside a small applicator. A shock wave is produced every time the projectile strikes the shock wave transmitter at the end of the applicator. The transmitter surface is placed on the skin of the patient at the pain region. This shock wave generation method is not used for ESWL.

Shock wave acupuncture, sometimes referred to as AkuST is also used to reduce pain. Shock waves are coupled into acupuncture points. Osteoarthritis and back pain are considered to be indications for AkuST. Treatment sessions are short, and no anesthetics are required. Shock wave action improves blood circulation and relieves pain. Energy flux densities, provided by so-called ballistic shock wave generators, vary between 0.01 and 0.23 mJ/mm². Shock wave rates go up to 10 pulses per second.

In equine medicine ESWT is popular to treat racehorses. Shock wave generators specially designed for veterinary applications have been on the market for many years.

**IX.3 Treatment of Peyronie’s disease**

Shock waves have been used for patients suffering from Peyronie’s disease, reducing pain on erection and measurably decreasing fibrotic plaque and penile curvature (Hauck et al. 2000; Mirone et al. 2000; Lebret et al. 2002). From 2000 to 4000 shock waves are applied to the flaccid penis and focused on the plaque, previously located by ultrasound. The procedure has a success rate of about 80% and does not require anesthesia, as the associated pain is reported as tolerable. More studies are still needed to understand the exact mechanism of action of the shock waves on the plaque. A theory is that shock waves increase vascularity, leading...
to the induction of an inflammatory reaction, which enhances the breakdown of the plaque. Even if initial results seemed to be promising, some authors are not so optimistic (Hauck et al. 2004), and long-term results have to be evaluated (Manikandan et al. 2002).

IX.4 Treatment of refractory angina pectoris

Shock waves are also used to treat refractory angina pectoris. The Modulith SLC (Storz Medical, Kreuzlingen, Switzerland) is the first clinical equipment for controlled application of several hundred shock waves on ischemic zones of the heart muscle, generating neo-angiogenesis and increasing blood circulation. The treatment does not require anesthesia and seems to be free of side effects. ECG triggering ensures shock wave release during the refractory phase of the cardiac cycle. The benefit is reported to be significant and lasting. Due to the proximity of the lungs and the vulnerability of the cardiac system, strict control of the shock wave field and dose is obligatory. Focal distance and generator aperture are specially designed for cardiac use. Shock waves are concentrated only on predetermined areas of the heart muscle. Targeting is controlled manually within millimeter precision using in-line ultrasound localization.

IX.5 Shock waves in ophthalmology

Phaco-emulsification and aspiration for cataract removal was introduced by Kelman (1967). During ultrasound, phaco-emulsification, energy is released at a probe tip to the cataract; however, non-target tissues such as the corneal endothelium, iris, and posterior capsule may be affected. Shock waves were used in ophthalmology for the first time by Dodick (1991). A pulsed Nd:YAG laser transmits light through a quartz fiber optic that enters into an irrigating probe. Less than 2 mm in front of the fiber end, a titanium target acts as a transducer to convert laser energy into shock waves. When the laser strikes the target, plasma formation occurs. The shock waves spread out to the mouth of the aspirating port. As the probe is held close to the cataractous tissue, shock waves cause the cataractous material to break off and be aspirated. The main advantages of this technique over ultrasound are that the laser system transmits considerably less heat to the tissue and that the laser probe does not need a motor and is much smaller and easier to handle than ultrasound probes.
Muestra del libro

Shock Wave Physics for Urologists

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