CONTROLLING ACOUSTIC CAVITATION FOR BIOMEDICAL AND MICROBIOLOGICAL APPLICATIONS

A. Loske, F. Fernández and A. Méndez
Centro de Física Aplicada y Tecnología Avanzada, UNAM, Querétaro, Qro., México
loske@fata.unam.mx

Abstract

We describe the conversion of a piezoelectric lithotripter into a device to control acoustic cavitation. In contrast to standard systems, our device produces *two* shock waves with an adjustable time delay (50 – 950 μs). One objective is to enhance cavitation – induced damage to kidney stones during extracorporal shock wave lithotripsy. In standard lithotripters, cavitation is produced near the stone after arrival of each individual wave. Bubbles expand, stabilize and collapse, creating stone-damaging microjets. Hundreds of shock waves, administered at 1 – 2 Hz, are needed to disintegrate the stone. Bubble collapse and microjet emission can be intensified by arrival of a second shock wave, having larger phase duration. Fragmentation efficiency and pressure measurements were compared to that of a standard lithotripter. Results indicate that efficiency was enhanced. Dual-pulse systems, as presented here, could also be used as a method for *in vivo* drug delivery and as a new food preservation method.

Introduction

Extracorporeal shock wave lithotripsy (SWL) is an effective treatment for urinary tract calculi [1,2]. Hundreds of shock waves are generated outside the patient's body and focused on the stone. These shock waves consist of positive pressure pulses of about 30 to 150 MPa, with a 10 ns rise time and 0.5 to 3 μs phase duration, followed immediately by a negative peak of about 3 to 20 MPa. Usually they are administered to the patient at a rate of about 1 - 2 Hz. SWL is reliable, nevertheless, clinical devices (lithotripters) are still evolving and improvements to increase efficiency are constantly sought. So far, three shock wave generation methods have been developed for SWL: electrohydraulic, electromagnetic and piezoelectric. This work was done using a piezoelectric system; however, the physical principles presented here could also be valid for the other two methods.

Piezoelectric lithotripters generate shock waves by a fast electric discharge (5 – 10 kV) applied to a set of up to 3000 piezoelectric crystals laid out on a bowl-shaped aluminum backing (see Figure 1). The crystals expand due to the high voltage peak, producing a pressure wave. A negative pressure peak results when the crystals return to their initial shape. The shock wave arriving at the center of the sphere (focus *F*) is generated by superposition of the pressure wave formed by each crystal. Water is used as a coupling media to transfer the energy into the patient's body through a latex membrane. The standard electric circuit consists of a capacitor charging unit (CCU), a spark-gap trigger (SG), and a spark gap driver (SGD). A pulse generator controls the discharge frequency.

During SWL, kidney stones disintegrate due to cavitation, spalling, layer separation and circumferential squeezing [2,3]. Cavitation bubbles are generated in the fluid (urine) near the stone by the negative phase of each shock wave. These tiny (1 μm – 1 mm) bubbles expand in 50 – 100 μs, stabilize and collapse violently after approximately 250 – 500 μs, producing stone-damaging high-speed microjets and secondary shock waves [4]. Microjet velocity is proportional to the initial bubble radius and to the bubble collapse energy. It has been shown experimentally that sending a second shock wave a few hundred microseconds after the first, may intensify bubble collapse and increase stone comminution efficiency [5-7]. This happens if the positive peak of the second shock wave arrives at the bubbles during, or shortly after, their stable phase [8]. Even if the negative phase of the second shock wave is not strong enough to reverse bubble collapse, it certainly slows it down. To prevent this, in our device it is possible to lengthen the phase duration of the second shock wave so that the negative pulse of the second shock wave arrives after bubble collapse. This innovation has not been tested before. The amplitude of both shock waves and the phase duration of the second wave may also be varied. Our objective is to enhance cavitation near the focus of the shock wave generator.

Methods

A *Piezolith 2300* (Richard Wolf GmbH, Knittlingen, Germany) lithotripter was modified to generate two discharges with an adjustable time delay between 50 and 950 μs, by installing a second capacitor charging unit (CCU 2), spark-gap (SG 2), and spark-gap driver (SG D). As shown in Figure 1, both units are charged via the high voltage supply (HVS). The system may be operated in manual or repetition mode. For experimental purposes, a cylindrical Lucite water tank and a *XYZ* positioner were mounted on top of the shock wave generator. The phase duration of the second shock wave can be modified by adjusting a variable resistor (3 – 85 Ω) connected between the second capacitor and the piezoelectric crystal array. A specially designed pulse generator (PG) triggers both spark-gap drivers.

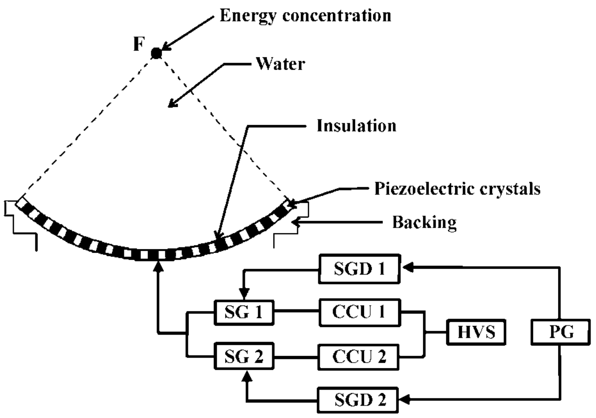

Figure 1: Block diagram of the dual-pulse piezoelectric shock wave generator (SG = spark gap, SGD = spark gap driver, CCU = capacitor charging unit, HVS = high voltage supply, PG = pulse generator).

Pressure measurements were done at *F* with *PVDF* needle hydrophones (Imotec GmbH, Würselen, Germany), having a 20 ns rise time (time for the output to rise from 10 to 90% of its final value). Signals coming from the hydrophones were fed into a 100 MHz digital oscilloscope (Tektronix, Inc., Beaverton, OR, USA, model 2430A). Ten pressure waveforms were registered with the conventional single pulse device and with the *tandem* system at different delays. *AST 110* rectangular kidney stone phantoms (High Medical Technologies, Kreuzlingen, Switzerland) were used to test fragmentation efficiency. Three phantoms were placed horizontally at *F* and exposed, one by one, to 600 shock waves at each delay. The size of the crater formed by the shock waves was compared by subtracting the final weight of the model from its initial weight. Stone comminution efficiency was defined as $E = 100\ (W_i - W_f)/W_i$. (W_i and W_f are the mean initial and the exposed stone phantom weight.)

Results

Mean positive and negative amplitudes recorded with the standard single-shock wave generator were about 38 and 18 MPa. Using the *tandem* system, the amplitudes of the second shock wave were 10 % lower. *E* was found to be about 20% higher for 600 pairs of shock waves than for 1200 single (standard) shock waves at delays between 350 and 450 µs.

Discussion

The emission of two time-delayed shock waves may produce better stone fragmentation than using standard single shock wave lithotripters. At small delays (< 300 µs), the second positive pulse seemed to suppress bubble growth and bubble collapse was less violent (smaller crater). At delays between 350 and 450 µs, the second shock wave intensified bubble collapse, increasing *E*. This happened because the second positive peak arrived during collapse of previously generated bubbles. At larger delays (> 500 µs), this pulse arrived after bubble collapse and no additional damage was produced to the phantoms.

Tissue damage is not expected to increase when using *tandem* shock waves, because bubble expansion *in vivo* is constrained by the tissue.

Since cavitation has been identified as an important mechanism to destroy *E. coli* [9], *tandem* systems as described here, may be useful to increase the bactericidal action of shock waves. They also could be used to improve *in vivo* drug delivery to target cells [10].

References

[1] C. Chaussy, E. Schmiedt, D. Jocham, W. Brendel, B. Forss-mann, V. Walther, “First clinical experience with extracorporeally induced destruction of kidney stones by shock waves,” Journal of Urology, vol. 127, pp. 417-420, 1982.

[2] A.M. Loske, “Shock waves in medicine,” in: “Handbook of Shock Waves”, vol. 2, chapter 12, Academic Press, New York, 2001, ISBN 0-12-086432-2.

[3] M. Lokhandwalla, B. Sturtevant, “Fracture mechanics model of stone comminution in ESWL and implications for tissue damage,” Physics in Medicine and Biology, vol. 45, pp. 1923-1949, 2000.

[4] L.A. Crum, “Cavitation microjets as a contributory mechanism for renal calculi disintegration in ESWL,” Journal of Urology, vol. 140, pp. 1587-1590, 1988.

[5] A.M. Loske, F.E. Prieto, “Improving underwater shock wave focusing efficiency,” Urolithiasis, C.Y.C. Pak, M.I. Resnick, G.M. Preminger (eds.), Millet The Printer, pp. 401-402, Dallas TX, 1996.

[6] P. Zhong, X. Xi, S. Zhu, F. Cocks, G. Preminger, “Recent developments in ESWL physics research,” Journal of Endourology, vol. 13, pp. 611-617, 1999.

[7] A.M. Loske, F.E. Prieto, F. Fernández, J. van Cauwelaert, “Tandem shock wave cavitation enhancement for extracorporeal lithotripsy,” Physics in Medicine and Biology, vol. 47, pp. 3945-3957, 2002.

[8] M.R. Bailey, "Control of acoustic cavitation with application to lithotripsy", Ph.D. dissertation, University of Texas, Austin TX, 1997.

[9] A.M. Loske, U.M. Álvarez, C. Hernández-Galicia, E. Castaño-Tostado, F.E. Prieto, "Bactericidal effect of underwater shock waves on *Escherichia coli* ATCC 10536 suspensions," Innovative Food Science and Emerging Technologies, vol. 3, pp. 321-327, 2002.

[10] S. Gambihler, M. Delius, J.W. Ellwart, "Permeabilization of the plasma membrane of L1210 mouse leukemia cells using lithotripter shock waves," Journal of Membr. Biology, vol. 141, pp. 267-275, 1994.

Acknowledgements

The authors acknowledge important technical assistance by Ulises M. Álvarez, Edgar Méndez, René Preza, and Helmuth Busch. The Richard Wolf *Piezolith 2300* lithotripter was donated to our university by *Dewimed SA de CV*, Mexico City. This project was sponsored by DGAPA IN101202 UNAM.