Shock Wave Pulse Pressure After Penetration of Kidney Tissue

L. Filipczynski, Jerzy Etienne, and T. Kujawska

***Abstract*—Lithotripter generated compressional peak pressure (p^+) was measured after penetrating tissue samples 0.5, 1, 2.5, and 4 cm thick. The samples were taken from fresh pig kidneys and insonated by shock wave pulses with positive peak pressures varying from 14 to 72 MPa. The measurement results allow one to draw some qualitative conclusions concerning the attenuation process in the lithotripter beam and show that in practice the compressional peak pressure values generated in the lithotripter may decrease almost by a factor of two before reaching the kidney stone. On the contrary, theoretical estimates indicate that negative pressures are practically not influenced by attenuation due to their different spectral composition.**

I. Introduction

MEASUREMENTS *in situ* of shock wave pulses administered to a patient during the lithotripsy treatment are nowadays difficult to carry out. In animals, however, such measurements were performed invasively by Vergunst *et al.* [9] in gallbladders of pigs. To find the amplitude and the shape of the shock wave pulse in the kidney we have carried out [5] simulation computations based on the shock wave pulse measured in water. We have introduced into the calculations the tissue attenuation coefficient A which—due to nonlinear effects—was assumed to increase m times with respect to the small amplitude value, according to the relation

$$A = mA_1 f^n \qquad (\mathrm{N_p/cm}) \qquad (1)$$

where A_1 denotes the small amplitude attenuation coefficient at the frequency of 1 MHz, f—frequency (in MHz). In our simulations it was assumed $A_1 = 0.1$ Np/(cm · MHz) and $n = 1$. According to Goss *et al.* [6] for small amplitudes $1 \leq n < 1.2$.

One can show that higher values of n introduce only small changes to the pulse shape and to its front steepness. This is shown in simulation results presented in Fig. 1(a) and (b) where the expected shock wave pulse was numerically reconstructed. The reconstruction was performed on the basis of the shock wave pulse $p_p\phi_p(t)$ measured in water by means of a PVDF membrane hydrophone with the sensitive electrode 0.5 mm in diameter (see curve W in Fig. 5). p_p represents the value of the positive pressure peak and $\phi_p(t)$ is the dimensionless time function describing the pulse shape.

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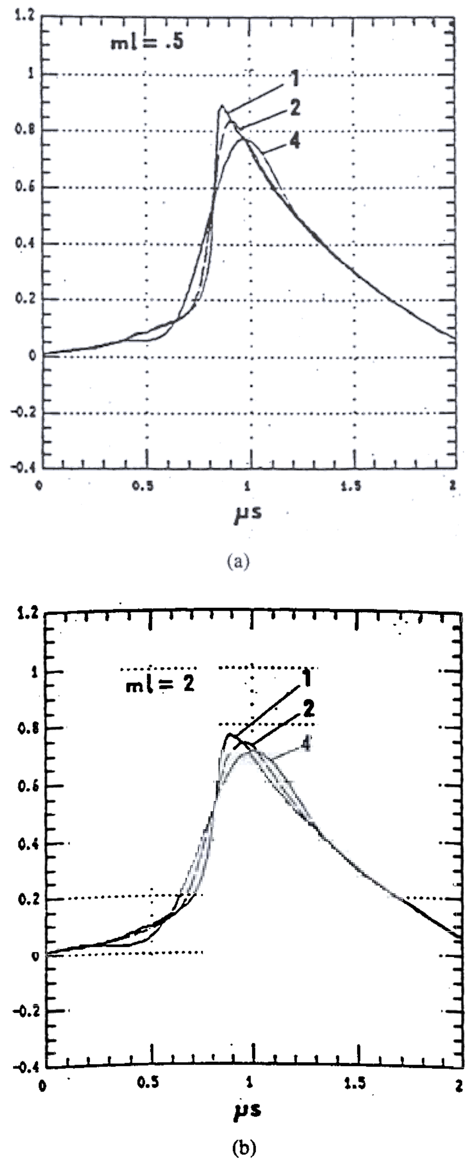

Fig. 1. The reconstructed shock wave pulse for $A_1 = 0.1$ Np/(cm · MHz), $n = 1, 2$, and 4 [see (1)]. (a) For $ml = 0.5$ cm (l is tissue thickness). (b) For $ml = 2$ cm. The vertical axis denotes relative pressure, the horizontal-time.

The reconstructed shock wave pulse can be expressed by the relation [5]

$$p_r\phi_r(t) = p_p F^{-1}\{F[\phi_p(t)]\exp(-A_1 f^n ml)\}$$

where l denotes thickness of the tissue sample (in cm).

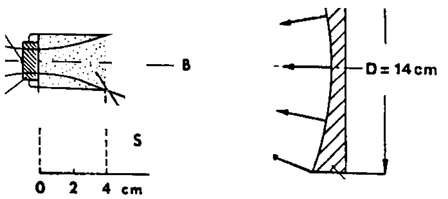

Fig. 2. Simplified geometrical relations in our experimental measurement system. B—shock wave beam, F—focal length, W—water, S—tissue sample, P—plastic bottomless container, HE—hydrophone's electrode (2 mm in diameter), HP—hydrophone's front metal plate (10 mm thick, 25 mm in diameter), L—plastic lens, D—lens diameter, l—thickness of the tissue sample.

First, the original time wave form $\phi_p(t)$ of the measured pressure pulse was transformed into the frequency domain by means of the Fourier transform F [see (2)]. Next, every frequency component was multiplied by the corresponding attenuation factor expressed by the exponential term. Finally, the inverse Fourier transform F^{-1} was computed in order to synthesize the attenuated shock wave pulses presented in Fig. 1(a) and (b). One should notice that in the case of small attenuation, i.e., when in the exponent of (2) $ml = 0.5$ cm, the pulse amplitude decreases by about 6% when n changes from 1 to 2. For higher attenuation values (e.g., $ml = 2$ cm) the amplitude decrease is even smaller [Fig. 1(a), (b)].

These results seem to be in qualitative agreement with the paper of Watson *et al.* [10] where the finite-difference solution of the KZK parabolic wave equation was used to investigate focused ultrasonic beams in tissue-like media in the steady state. Fig. 1(a) and (b) also demonstrate the small change of the pulse steepness when changing the value of n from 1 to 2.

One should notice that absorption of finite amplitude waves cannot in principle be described by the exponential dependence on distance as was done in (2). However, it is permissible as an approximation for the short wave path in tissue when compared to the long focal distance. This can be shown by means of exact formulas published by Naugolnych [8].

In our calculations it was assumed that the attenuation coefficient due to nonlinear effects increases m times with respect to the small amplitude attenuation coefficient and that the shock wave is approximately plane. In reality one should expect that attenuation depends on pressure as a continuous although generally unknown function.

The purpose of this paper is to demonstrate measurements of shock wave pulses penetrating kidney tissue samples of various thicknesses and at various pressures which are or were used in lithotripsy, as well as to explain theoretically questions connected with the influence of attenuation on the shape of the shock wave pulse.

II. Experiments

Measurements were carried out in our experimental lithotripter [2] in which an electromagnetic membrane generator was used. The radiated beam was focused by means of a plane-concave plastic lens 14 cm in diameter and 20 cm focal length (Fig. 2). The focus itself was 8 cm long and 0.5 cm in diameter when measuring the −6 dB positive pressure amplitude of the generated shock wave pulse [3].

For pressure measurements our capacitance hydrophone with a 2 mm diameter electrode was used [3]. It is similar to a microphone, however, instead of a thin membrane, employs a 10 mm thick metal plate, 25 mm in diameter. The polarized electrode is situated near to the back surface of the plate in an air cavity. When placed in the focal plane of the lithotripter narrow beam, the hydrophone's metal plate can be considered as a large reflecting surface which does not influence the shape of the incident wave. The hydrophone's relatively large electrode causes a certain effect of surface averaging. For the focal pressure of 40 MPa, the −6 dB focal beam diameter in our lithotripter equals 5 mm [3]. In such a case the surface averaging decreases the maximum pressure value by about 2 dB.

The capacitance hydrophone does not reproduce exactly the descending slope of the shock wave pulse and its further portion which forms an overshoot. This is caused by the fact that the incident pulse pressure generates at the front surface of the metal plate simultaneously a high longitudinal wave pulse and a different elastic wave type component which propagates with a much lower speed. After a short time of about 1.5 μs this wave component reaches the back surface of the plate, making the exact measurement impossible [3].

In spite of this fundamental drawback the capacitance hydrophone was especially useful in our measurements because of its very high durability even at highest pressures. More than one thousand shock wave pulses were necessary to observe very small changes on the hydrophone's metal front surface.

The requirements for shock wave hydrophones were discussed in detail in [7].

In every one of 45 measurements three pressure curves were recorded on one plotter chart, thus verifying the repeatability of measurements. Fig. 3(a) and (b) show as an example positive peak pressures measured in water and after penetrating a kidney sample 1 cm thick. The hydrophone was always situated in the beam focus while tissue samples were placed on the hydrophone's front plane (Fig. 2).

The samples were taken from fresh pig kidneys. They were stored only few hours in a temperature of about 5°C and measured at the temperature of 24°C. To obtain thicknesses equal to 0.5, 1, 2.5, and 4 cm plane sections of the cortex and medulla region were sliced and grouped together to form the desired thickness. Transverse dimensions of the samples were greater than the diameter of the hydrophone's plate, i.e., 25 mm (see Fig. 2).

Fig. 4 presents results of our measurements. The horizontal axis denotes thickness of the tissue and the vertical denotes peak positive pressure. Vertical bars represent measured positive peak pressures. The pressures measured in water was varied from 14 to 72 MPa in 9 steps. Results for both water value ($l = 0$ cm) and after penetration of kidney samples of the thickness $l = 0.5$, 1, 2.5, and 4 cm, respectively, are shown on each horizontal line.

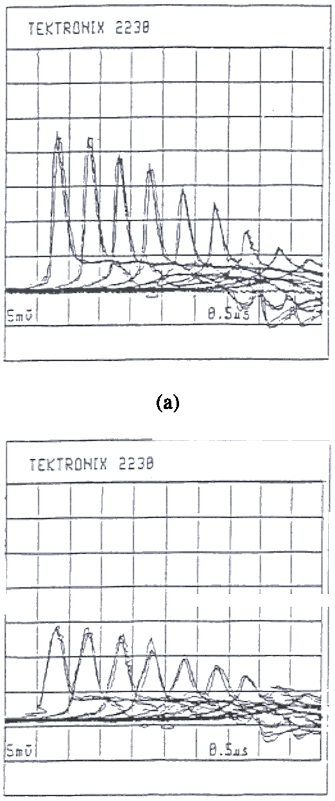

Fig. 3. Positive peak pressure pulses obtained by means of the capacitance hydrophone. Vertical axis division = 5 mV, horizontal = 0.5 μs. (a) Measured in water. The generator voltage was equal to 25, 24, 23, 22, 21, 20, 19, 18, and 17 kV, producing the corresponding pressure values of 72, 69, 60, 59, 46, 40, 26, 19, and 14 MPa, respectively. (b) Measured (in the same sequence) after penetration of the 1 cm thick kidney sample.

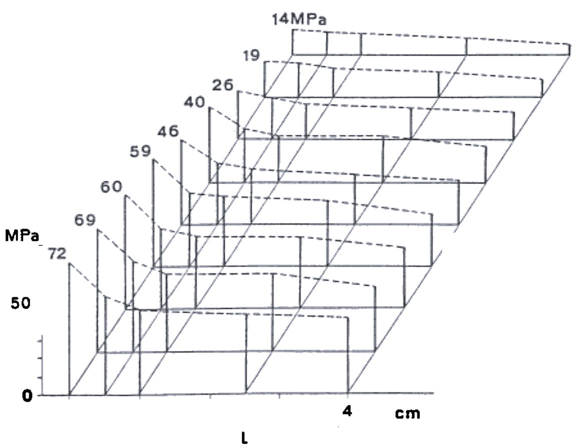

Fig. 4. Positive peak pressure values measured at the focus for various peak pressures generated in water ($l = 0$), at intervals equal from 72 to 14 MPa, and measured after penetration of kidney samples of the thickness $l = 0.5$, 1, 2.5, and 4 cm.

III. Discussion and Conclusions

At present there is no analytical solution of the general problem of nonlinear focused sound fields [1]. The attenuation problem in a nonlinear focused pulse system is very complex since with the pressure variation one obtains complicated variations of the pulse shape and its spectrum. Higher harmonic content causes higher attenuation due to its frequency dependence as shown by (1). Nevertheless one can draw interesting conclusions from our results shown in Fig. 4.

The slope of the dashed lines can be used for a qualitative estimation of attenuation for the small value of the sample thickness $l = 0.5$ cm which is situated in the focal region (Fig. 2). The penetrating wave is there approximately plane (see also [5, Fig. 1]). In this case one can assume that the changes of the pulse spectral composition along the short path of l are small. For higher values of l, outside of the focal region, where the beam is converging all these assumptions are no more valid. In these cases the slope of the dashed lines shows the rate of the peak positive pressure loss which in a complex way depends on many factors including attenuation.

For the lowest pressure in water (14 MPa), when considering $l = 0.5$ cm, attenuation is several times smaller than at higher pressure. One observes then the saturation of the shock wave pulse attenuation; the increase of the pressure pulse in water from 59 to 72 MPa has practically no effect on the attenuation value. This high and almost constant attenuation may be caused by the well-developed, steep shock wave pulse and the accompanying higher harmonics.

A certain scatter of experimental results and possible defocusing effects should be taken into account when interpreting these measurements. However, we do not expect defocusing could be increased by tissue when compared with water since the wave speed changes in tissue and water are almost the same as a function of pressure [5]. Also dispersion in tissue can be practically neglected—as was shown in [5].

In renal lithotripsy, where the distance between the stone and the patient's body surface may be equal to 4 cm and where the peak pressures are about 20–40 MPa, the pulse pressure at the stone surface may be about half the peak value measured in water as can be seen from Fig. 4. In fact, the shock wave pulse is attenuated by the outer layer of skin, muscle, and fat tissues surrounding the kidney and by the kidney itself before reaching the stone. The attenuation in these layers can be approximately estimated to be the same as in kidney samples used in our measurements.

The above conclusion is valid for positive peak pulse pressures. Negative pressures could not be measured by the authors since the capacitance hydrophone reproduces exactly only the pulse front and the positive pulse amplitude while deforming the negative part of the shock wave pulse. On the other hand our conventional PVDF membrane hydrophone, which reproduces exactly positive and negative parts of the shock wave pulse could not be applied in such a great number of measurements due to its low durability, especially at the highest pressure used in our measurements.

However, the negative part of the pulse is most likely influenced only minimally by attenuation. This conclusion can be drawn from Fig. 5, which presents the shock wave pulse measured in water by means of the PVDF membrane hydrophone which follows exactly the positive and negative portions of the shock wave pulse. It shows also the numerically reconstructed pulses for high attenuation values ($ml = 5$ cm and 15 cm). Even if the positive peak pressure value decreases by a factor of two, the negative peak value decrease is about

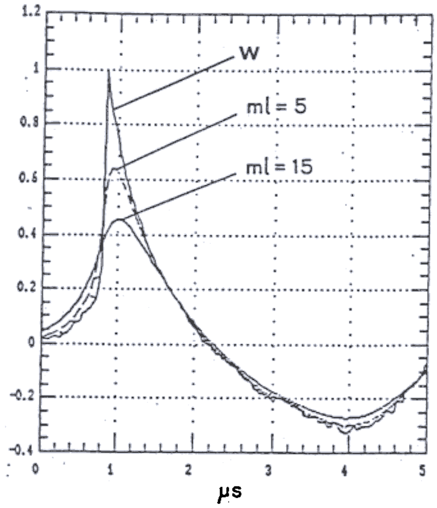

Fig. 5. The shock wave pulse pressure (relative) measured in water (W) by means of a PVDF hydrophone [3] and reconstructed numerically for two high attenuation values (ml = 5 cm and 15 cm).

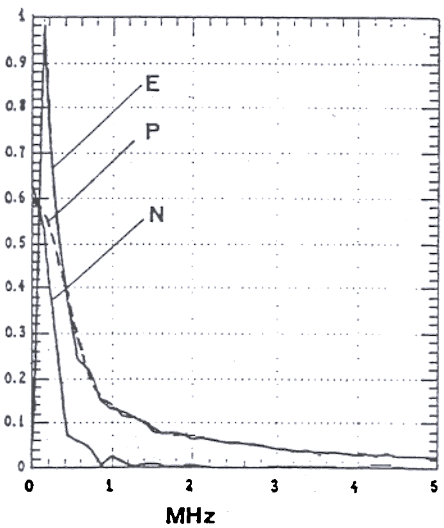

Fig. 6. Computed amplitude density spectra of the entire shock wave pulse (E) measured in water (see Fig. 5) and of its positive (P) and negative (N) parts.

ten times lower. This effect can be explained by the fact, that the negative part of the pulse contains mainly low frequency harmonics as it is presented in Fig. 6 (see also [4]). Therefore, their attenuation, according to (1), is lower than that of higher frequency harmonics which are contained in the positive part of the shock wave pulse spectrum.

Our measurement results obtained *in vitro* may differ from those which would be obtained *in vivo*. For example, cavitation within tissues might be different. It seems that this question is at the moment open although Vergunst *et al*. [9] have not found differences when comparing lithotripsy measurements in animals *in vivo* with those *in vitro*. If this observation is correct, then our results show realistic values of pressure peaks in kidney which can be expected during the lithotripsy treatment.

The results obtained may be useful in the study of mechanisms of kidney stone disintegration as well as in the study of harmful effects of shock wave pulses in living tissues.

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