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Abstract: A transient pressure wave created in photoablation-based laser corneal refractive surgery propagates along the optical axis of the eye from the cornea towards the fundus. Along its path, it is attenuated, reflected several times at tissue interfaces, focused due to the spherical shape of the cornea and partially phase-converted to a tensile wave after passing the acoustic focus. Employing the numerical acoustic eye model, the dynamics of the acoustic focusing is described in detail. The simulated pressure waveforms are first validated against the existing measurements. Then, the calculated full pressure fields are used to obtain the extreme values of the pressure amplitude at every point within the eye over the whole timespan of interest. Over these pressure maps, various threshold contours obtained from the existing literature are superimposed, to help ophthalmologists investigate the consequences of acoustic focusing within the human eye during laser keratorefractive surgery. The comparison indicates that the simulated negative peak pressure values used in the contemporary broad beam and flying spot LASIK/PRK treatments exceed some of the reported threshold values. The critical volume where the thresholds are surpassed is located near the lens-vitreous boundary and extends to the retina for wide beams. The photoablation-induced pressure wave focusing is therefore a mechanism that needs to be researched carefully and included in the list of potential risks accompanying refractive laser surgery.

Key Words: laser ultrasonics, negative pressure, mechanical index, laser refractive surgery

INTRODUCTION

The cornea is the prevalent part of the eye that focuses both optical and acoustic waves. Due to its spherical geometry, the air-cornea interface accounts for approximately 2/3 of the eye's total optical power and focuses visible light on the retina, located at a depth of about 23 mm from the entrance. Human eye as an optical apparatus for adaptive focusing of light is an evolutionary feature, which has evolved by natural selection to endow our species with vision. As such, side effects of optical focusing have naturally been minimized. On the other hand, when man-made transient acoustic waves (medical ultrasound¹ or laser-induced shock waves²), to which human eye has been exposed only recently, impinge on the cornea³ or are created on the cornea^{4,5}, its spherical shape focuses them to a volume located somewhere between the ocular lens and the anterior vitreous, which coincides with the $R = 7.8$ -mm radius of curvature of the cornea⁶. Within the focal volume, the pressure fields may surpass the thresholds for the inception of various physical effects that may potentially damage ocular tissues. This paper describes the dynamics of acoustic focusing in detail and gives numerical estimates of pressure fields alongside with the threshold contours that can help ophthalmologists investigate the consequences of acoustic focusing within the human eye during laser keratorefractive surgery.

The prevalent approach in laser corrective surgery involves modification of the refractive characteristics of the eye by cornea photoablation with excimer laser pulses in the UV wavelength region⁷⁻⁹. The photoablation mechanism, also known as photochemical ablation or ablative photodecomposition, includes material ejection, resulting in reaction force and creation of a transient compression within the treated tissue, propagating into the interior of the eye. The mechanical wave undergoes a series of changes during propagation, such as focusing, reflection at boundaries, phase shift and attenuation. Some aspects of this complex evolution have already been studied in detail using the acoustic eye model (AEM) simulation to investigate the photoionization inside the eye as the source of mechanical waves³. In that study, an important phenomenon was revealed, namely how the geometry of the eye evokes the phase conversion of the mechanical disturbance, resulting in the creation of large negative pressures. High negative pressure is important in this context, because tissue is much more susceptible to damage by tension than by compression¹⁰, the effects ranging from delaminations and lacerations to tearing, cavitation-induced lesion formation and cutting of tissue¹¹, caused by acoustic (inertial) cavitation.

From the physical point of view, focusing of photoablation-induced mechanical transients is similar to the therapeutic technique employing low- (LIFU) or high-intensity focused ultrasound (HIFU), where tissue in the focal volume is either stimulated (LIFU)¹² or destroyed by heating (HIFU)¹³, or to the destructive ultrasonic histotripsy, where the targeted tissue experiences mechanical breakdown by intense sonification¹⁴. Here, bursts of multicycle ultrasonic pulses with frequencies in the MHz range or continuous ultrasound are generated by bowl-shaped phased array transducers, resembling the spherical shape of the cornea. In contrast, during transient absorption of laser light in ocular tissues¹⁵, causing photoionization¹⁶ and photoablation¹⁷, short acoustic pulses with a typical duration of 100 ns and spatial dimension around 0.2 mm are emitted. The latter laser pulse-tissue interaction mechanism is employed in photoablation-based laser refractive surgery to reshape the cornea and thus permanently correct vision problems such as nearsightedness, farsightedness and astigmatism, most commonly used procedures being PRK (photorefractive keratectomy) and LASIK (laser-assisted in situ keratomileusis). Even though laser keratorefractive surgery is promoted as minimally invasive and safe, there is only limited information on the accompanying unwanted effects, such as retinal tears¹⁸ and the US Food and Drug Administration (FDA) advises potential users that long-term data of the risks is still not available¹⁹. As pointed out in the review by Mirshahi and Baatz²⁰, this can partly be attributed to the absence of comprehensive methodic study which would include long-term follow-up reports. There are several reports on the irreversible changes observed in human eye tissue following laser surgical procedures and the mechanical effects following the laser pulse impact were often considered as a possible mechanism that causes these changes^{17,21,22}. Lately, in the literature, some more instances of possible complications following LASIK/PRK procedures have been mentioned that are described as possibly related to the excimer laser-induced shock wave. These include: corneal damage, endothelial cell loss, retinal detachment, posterior vitreous detachment, alteration in fragile subretinal vessels, Bruch's membrane

and retinal epithelium, macular hemorrhage and choroidal neovascularization⁹, opening of the gap in Bruch's membrane and generation of lacquer cracks²³ and serous macular detachment, increase of the gap in Bruch's membrane, posterior vitreous detachment, structural damage and intraocular inflammation of the retina²⁴. The above mentioned complications are the motivation for our work which examines the frequently overlooked mechanical effect that accompanies laser corneal refractive surgery, namely the evolution of the compression wave, formed during the photoablation of the cornea.

The threshold values of negative pressures having effect on tissue are reported from -40 bar causing cell lysis²⁵, while -200 bar already produces free-field cavitation in water²⁶, followed by the cavitation bubbles collapses that can cut the eye tissues^{11,25}. Free field cavitation in tissues commonly occurs at a slightly lower threshold than in water²⁷. Positive peak pressures that affect cell viability are reported to be much higher ($+400$ bar²⁸ or $+800$ bar¹⁰). These processes are of probabilistic nature and the threshold data usually refer to 50% probability. For that reason, the mentioned physical processes may occur also at pressure values lower than the reported threshold values, although with a smaller probability. The minimum pressure level capable of inducing a process can be twice as small as the 50% probability threshold²⁹.

As for the cornea photoablation, relevant measurements were performed on artificial samples and ex vivo on porcine^{4,17} and human eyes⁵. UV laser pulses with flat spatial profile were used, with variable apertures up to 6.5 mm in diameter⁴. Siano et al.⁴ used a needle hydrophone with a 0.5-mm diameter to record pressure waveforms at different positions on the symmetry axis inside the eye, following corneal tissue photoablation. As they stated, the main result of their time-resolved shadowgraphy in artificial samples was the "evidence of a significant focusing of the acoustic wave front for the larger irradiation spot diameters". Esposito et al.³⁰ report on similar effects of the constructive superposition of acoustic waves during primary blast injuries of human eyes due to explosion events in the proximity of the eye. Siano et al.⁴ analyzed the evolution of the peak positive and the peak negative pressures along the eye symmetry axis for different photoablation spot diameters. They measured that, in the case of the 6.5-mm-diameter irradiation spot, values up to $+250$ bar (of compressive stress) and -90 bar (of tensile stress) were reached in the focal volume. As they also pointed out, the large value of peak negative pressure is more critical from the point of view of unwanted structural and cellular damage.

TABLE I. Properties of the eye tissues used in the model

Tissue	Density [kg/m ³]	Bulk modulus [GPa]	Wave velocity [m/s]	Attenuation coefficient @ 10 MHz [cm ⁻¹]
Cornea	1087	2.6420	1559	0.42
Sclera	1050	2.7150	1608	2.6
Lens	1040	2.7667	1631	1.2
Fat	950	1.9974	1450	0.69
Ciliary body Ciliary zonule	1065	2.6587	1580	0.80
Vitreous humor Aqueous humor	994	2.2455	1503	0.12

ACOUSTIC EYE MODEL (AEM)

The acoustic eye model (AEM)³, based on finite element method, is capable of modeling wave propagation starting with virtually arbitrary input loading. The AEM was built in the ABAQUS 6.16 finite element software (Simulia, Providence, 2018, RI). In this work, the model proposed previously³ was upgraded to recreate the above described measurements, taking into account as accurately as possible the geometric (same as in Ref. 3) and material characteristics (listed in Table I), and the initial

input loading. In the axisymmetric model, the eye was considered as a lossy acoustic medium, excluding nonlinear effects, but including the absorption coefficients of soft tissues, taken from the work of de Korte et al.³¹. To increase the accuracy of the simulations, the finite element mesh was refined to consist of more than 15 finite elements over the acoustic wave front. Additionally, eight node quadrilateral finite elements with quadratic formulation were used with the finest element mesh size of 7.5 μm .

RESULTS AND DISCUSSION

The dynamics of the photoablation-induced pressure waves within the eye is first validated against the existing, but scarce measurements, followed by an elaborated description of intricate physical effects accompanying the temporal evolution of the pressure field. Finally, we discuss the implications of the acoustic focusing related to keratorefractive surgery by comparing the simulated pressure contour maps with the recommended values and threshold values for the inception of secondary physical effects such as the potentially harmful inertial cavitation.

Comparison of AEM results to measurement

The temporal pressure profile of the input mechanical loading in the AEM was taken from Siano et al.⁴ who measured laser photoablation-induced pressure waveforms immediately below the inner boundary of the cornea, on the axis 1 mm inside the eye. For photoablation, they used the commercial surgical ArF excimer laser (Apogee Summit Technology, Waltham, MA, USA) at 193-nm wavelength and 15-ns (FWHM) pulse duration, with a fluence of 175 mJ/cm^2 , repetition rate of 1–10 Hz and different apertures, from $2r = 6.5$ mm in diameter to 1 mm, keeping the light fluence of the flat top profile constant. In order to match our simulation results to their measurements, we found that an initial monopolar compression pulse with a temporal FWHM of 51 ns and with a constant transient loading of $p_0 = +86$ bar throughout the circular area with a radius r , and an additional, 0.75-mm-wide, smooth cosine taper to zero at the edge, gives the best fit. To validate the AEM, three selected broad beam profiles were modelled. The physical parameters of the corresponding light pulses and the generated pressure distributions are listed in the first three columns in Table II.

TABLE II. Fluence and pressure profiles used in the simulations

Type	Fluence				
	Flat top				Gaussian
Radius [mm]	3.25	1.5	0.75	0.5	0.5
Max. fluence [mJ/cm^2]	175	175	175	200	400
Avg. fluence [mJ/cm^2]	175	175	175	200	200
Energy [mJ]	58	12	3.1	1.6	1.6
Type	Pressure				
	Flat top + taper				Gaussian
Max. pressure [bar]	86	86	86	99	197
Total radius [mm]	4	2.25	1.5	0.555	0.5
Flat part [mm]	3.25	1.5	0.75	0.444	/
Cosine taper [mm]	0.75	0.75	0.75	0.111	/
Force [N]	360	96	35	7.7	7.7

Temporal evolution of the pressure field

A single frame of the animation showing the calculated pressure field evolution for the largest spot diameter (6.5 mm) is presented in Fig. 1 (Multimedia view). A few snapshots from the animation are presented in Fig. 2, at 1 μs , 3 μs , 5 μs and 8 μs . The largest aperture size is chosen for the general

description and discussion of the pressure field evolution because of the convenience of visualization. The same features can be observed also with smaller spot sizes.

Fig. 1. Animation at 4.85 μs shows the calculated pressure field for the largest spot diameter (6.5 mm) at the focus. (Multimedia view).

Fig. 2. Cross section of the eye with calculated pressure field at 1 μs , 3 μs , 5 μs and 8 μs for the 6.5-mm initial spot diameter.

After the laser-induced material ejection, a compression pulse is created at the outside boundary of the cornea. Initially, the temporal evolution of the pressure field follows the theoretical description of the complex acoustic field of a pulsed circular piston^{32,33}, where the radiated field can be viewed as having two contributions: (i) a transient, monopolar (compression only) plane wave travelling normally to the radiating surface and confined to a region of interior points inside the cylindrical volume representing the geometrical shadow of the radiating surface and (ii) a hemi-toroidal transient wave emanating from the edge of the radiating surface. The two contributions can be clearly resolved in the beginning of wave propagation (Fig. 2, at 1 μs). The thin grey arcs help the observer to distinguish the propagation of the edge wave and the central wavefront. The theory also accounts for the π phase jump of the edge wave³², namely the outer part of the edge wavefront has the same phase as the central wave, seen in Fig. 2 as a compression (red), while the inner part of the edge wave has the opposite phase, a rarefaction (blue).

In the cornea-photoablation case, there is a distinction from the flat piston source, because the cornea is curved, resulting in the same curvature of the initial central field wavefront. This discrepancy from the 'flat' case is later on responsible for the difference in the far field propagation. The two contributions, i.e. the central part and the edge wave, are superimposed, resulting in a complex field. The pressure field evolution is better followed if their propagation is considered separately: (i) Because of the curvature of the source, the central part of the wavefront is focused to a small volume just beyond the eye lens (Fig. 2, at 5 μs). The propagation of this part of the wave is confined approximately to the two cones having a common apex. The approximate boundary of this volume is depicted in Fig. 2 with two thin grey, crossing lines. Upon focusing, the compression wave undergoes a Gouy phase shift, thus transforming to a rarefaction, described in detail elsewhere³. In the animation (Fig. 1, Multimedia view), the positive peak pressure rises from the initial 86 bar to over 1200 bar when it reaches the focal volume, yielding a pressure gain of $G = 14$. The transformation from the compression to the rarefaction after passing the focal point cannot be clearly observed at this time, because of the superposition with the simultaneous arrival of the edge wave. The Gouy phase shift, manifested in the described transformation of the central part of the wave from a compression to a rarefaction, becomes obvious after a few μs , e.g. at 8 μs , after the contributions have already spatially decoupled. As can best be seen in the animation (Fig. 1, Multimedia view), after passing the focus, the leading compression can be identified as originating exclusively from the edge wave. (ii) The edge wave expands evenly in a hemi-toroidal fashion. Note also, that in Fig. 2, at 5 μs , multiple reflections can be observed on the axis of symmetry behind the leading wave. A reflection is generated each time the wave encounters a jump in acoustic impedance at each interface of different tissues within the eye. Such reflections were experimentally observed by Pini et al.³⁴

Examining the components contributing to the total field at 8 μs (Fig. 2) and tracing them back, it can be determined that after passing the focal volume, the leading positive pressure (red) originates solely from what was initially the outer portion of the edge wave. Regarding the negative pressures (blue), the examination shows that the central part of the rarefaction (on the axis) originates from the initial monopolar compression wave which had experienced a Gouy π phase shift after passing the focus. The already mentioned leading compression wave on the axis originates from the edge wave and it precedes the π -shifted rarefaction because of the slightly shorter path it had to travel. The outer portion of the rarefaction is the originally inner part of the edge wave. After the passing of the focal volume,

from approximately 6 μs on, the compression and the rarefaction wave on the symmetry axis are thus coming from different parts of the bowl-shaped source and are even separated spatially and temporally, as can best be visualized in the animation video (Fig. 1, Multimedia view).

The evolution of the acoustic field for the 3.0-mm and 1.5-mm irradiation spot diameters follows along the same lines and can be described in the same manner. Since the central portion of the field in those cases is narrower, the focusing is less apparent and the overlapping of the two contributions even more pronounced. The overall effect is that during the propagation, the wave does not spread out laterally as much as when the largest spot size (6.5 mm) is used. This also means that the relative rise in peak pressure during propagation, both positive and negative, is smaller. For the same reason, the focal volume is more elongated along the axis for smaller photoablation areas.

In a few selected points along the axis, we compared the simulated pressure waveforms with the measurements of Siano et al.⁴ for two spot diameters (Figs. 3(a)–3(b)). To make direct comparison, we had to take into account the dimension of the pressure sensor (0.5 mm in diameter), which was comparable to the typical spatial dimension of the wave. For this reason, in the simulation, the pressure pulse arriving at the sensor was averaged over the aperture of the sensor to yield comparable results. Given the fact that other influencing parameters of their experiment are known only to a limited degree, the matching of the waveforms is relatively good. It is not unexpected that overall larger absolute pressures (positive and negative) are obtained with the simulation, compared to the measured ones. The eye model in the simulation has perfect rotational symmetry, unlike the real samples, used in the experiments. This means that the constructive interference of the contributing waves, especially on the symmetry axis, is more pronounced in the simulation, leading to higher pressure amplitudes. We believe that the reason for the relatively large mismatch between the measured and simulated negative pressure values lies in the contact nature of the measurement. Namely, in a related research by the same group^{34–36} they stated that they used the PVDF needle-probe hydrophone pressure sensor (Imotec GmbH, Germany). This sensor is described in detail in the book by Loske² (p. 35–36), mentioning that “A disadvantage is that water–metal shielding adhesion at the tip of the gauge does not withstand the tensile phase of a shock wave. As a consequence, the duration and amplitude of the negative pulse is underestimated. This has to be considered when calculating energy values from pressure records obtained with needle hydrophones.” This explains the relatively larger discrepancy between the simulated and the measured waveforms (Fig. 3) for the negative pressure part of the wave, compared to the compression part. A similar effect, although not explicitly stated, can be noticed in the comparison of fiber tip hydrophone measurement of the laser generated focused ultrasound to the simulated waveforms³⁷ (Figs. 6(c) and 6(d) in Ref. ³⁷). This suggests that because of the nature of the measurement, all contact sensors are bound to underestimate the actual negative pressure in liquids, due to the detachment at the boundary between the propagating medium and the sensor at high tensile stresses.

The same considerations regarding the limitations of the pressure sensor have to be taken into account in the analysis of the peak pressure on the symmetry axis. In Figs. 3(d)–3(e), the comparison between the measured⁴ (white circles) and calculated (black circles) peak pressure is presented for two different irradiated spot diameters. The calculated data again includes the integration over the sensor area. The underestimation of the measured negative pressures, explained above, is observed again. The simulated point pressure peaks on the symmetry axis are shown as blue solid lines. From the comparison of the point peak pressure to the measured pressure which represents spatial average of the pressure over the sensor area, it can be deduced that the local point-pressure peaks are in fact considerably larger than the measurements suggest, in the particular case up to 3 times larger for the compression and, because of the sensor limitations, possibly even 7 times larger for the rarefaction.

Fig. 3. Left column: Comparison of simulated pressure waveforms [(a)–(c)] to measurements by Siano et al.⁴ [(a)–(b)], for different spot diameters: (a) 6.5 mm, (b) 3.0 mm and (c) 1.5 mm at three selected depths (P_1 , P_2 and P_3) along the axis. Right column: Calculated (black circles) and measured⁴ (white circles) peak pressure (positive and negative) for three different irradiation spot diameters: (d) 6.5 mm, (e) 3.0 mm and (f)

1.5 mm. The solid blue line is the calculated local peak pressure at a given point while the black circles represent the same pressure, but averaged over the sensor area.

Figs. 3(c) and 3(f) show the results for the 1.5-mm spot diameter. Acoustic focusing is less pronounced for 1.5 mm laser spot diameter compared to the 6.5-mm and 3.0-mm diameters. The point-pressure peak is similar to the aperture-averaged pressure peak, since the wave impinges on the sensor perpendicularly with relatively uniform pressure over the whole sensor tip (see Fig. 3(f)).

Based on Figs. 3(d)–3(f), the numerical gain (maximum positive peak pressure $p_{\max}$ divided by the surface loading pressure $p_0 = 86$ bar) is estimated to be 14 for the 6.5-mm spot, 3.5 for the 3.0-mm spot and 1.7 for the 1.5-mm spot. The gain can also be calculated analytically using a simplified reasoning. Assuming a spherical cap array of point radiators with the radius of the base of the cap r and the radius of the sphere $R = 7.8$ mm, the focal positive pressure $p_{\max}$ is enhanced compared to the loading surface pressure p_0 , by

$$G = \frac{p_{\max}}{p_0} = \frac{r^2}{2R^2} \sqrt{1 + \left(\frac{2\pi f R}{c} \right)^2} e^{-\alpha R}, \quad (1)$$

where $f = 10$ MHz is the dominant frequency of the acoustic excitation, $c = 1500$ m/s is the approximate speed of sound in the eye tissues, and $\alpha = 0.8 \text{ cm}^{-1}$ is the average pressure attenuation coefficient of the anterior part of the eye⁴. Using this approximate equation¹² and taking into account the acoustic attenuation, the gain is calculated to be 15 for the $2r = 6.5$ -mm spot, 3.2 for the $2r = 3.0$ -mm spot and 0.81 for the $2r = 1.5$ -mm spot. The match between the numerical gain and the approximate, analytically obtained gain is very good, except for the smallest spot, where the additional, 0.75-mm-wide taper should have been effectively added to the spherical cap size r .

Implications for medical procedures inside the eye

Due to the expanding use of ultrasound for diagnostic as well as therapeutic purposes, its biological effects are studied extensively^{38,39}. Two non-dimensional parameters: mechanical index (MI) and thermal index (TI) were introduced as an assessment tool in the safety recommendations and regulations by ultrasound professional community of World Federation for Ultrasound in Medicine and Biology (WFUMB) regarding the use of ultrasound in diagnostic applications⁴⁰. The mechanical index is defined as being proportional to the negative peak pressure of the ultrasonic wave, reflecting the fact that from the standpoint of tissue damage, the negative pressure represents greater concern than the positive pressure of the same absolute value, although the amplitude of the positive pressure also plays an important role²⁹. There are several mechanisms by which the ultrasonic wave can harm tissue mechanically. As pointed out by ter Haar⁴⁰, most of the mechanically induced effects are due to acoustic (inertial) cavitation, which is a threshold, statistical phenomenon, so higher MI represents higher probability for unwanted tissue damage. The recommendations of various professional associations for the safe use of diagnostic ultrasound have set the limits for MI and TI. For most applications, the recommended highest value for MI is 1.9, with a marked exception of ophthalmological applications where the recommended limit for MI is set to 0.23, an eight times lower value⁴⁰. From the work of Apfel et al.⁴¹ who studied effects of short pulse, low duty cycle diagnostic ultrasound and who proposed a similar index regarding acoustic cavitation, it can be assumed that the general results regarding MI concept in connection with unwanted tissue damage can also be used for single cycle pulses, like the ones that are created during laser eye surgery.

Here, the ultrasonic waves are an unwanted phenomenon. It is thus important to identify the volume inside which the pressure is larger than the value at which substantial probability for damage is expected. Fig. 4 depicts the cross section of the volume inside which the chosen limits, shown as the contour lines, were exceeded at any time during the wave propagation in our model. The limits used in Fig. 4 were determined according to the values reported elsewhere: (i) -7 bar, calculated for the mechanical index recommended for diagnostic ultrasound in ophthalmologic applications (value $MI = 0.23$, at the central frequency of 10 MHz), (ii) -40 bar for cellular damage²⁵, (iii) -200 bar, reported as the pressure value

for the free field cavitation in water at 10 MHz²⁶ and (iv) +800 bar, reported as the value for structural damage in tissue¹⁰.

Fig. 4. Cross section of the eye showing the critical volume inside which the thresholds (see the legend) for negative and positive pressure peaks are exceeded for a variable broad beam with (a) 6.5-mm, (b) 3.0-mm and (c) 1.5-mm irradiation spot diameter (all 175-mJ/cm² fluence, corresponding to initial 86-bar peak pressure) and for a small-size flying spot beam with 1.0-mm irradiation spot diameter at 200 mJ/cm² average fluence for (d) flat top profile and (e) Gaussian profile, corresponding to the typically used parameters of the contemporary laser systems. For the purpose of comparison, the corresponding energy per pulse is also indicated.

The pressure contours in Figs. 4(a)–4(c) were calculated with the input broad beam parameters for various spot sizes used in the experiments by Siano et al.⁴, used also in this work, while the contours in Fig. 4(e) were calculated with the typical parameters of 1-mm-spot Gaussian excimer laser pulse that are commonly used today in LASIK/PRK surgery with small size flying spot laser systems. Figure 4(d) presents the pressure extremes for the flat top pulse that has the same energy and width as the Gaussian beam (compare the fourth and the fifth column in Table II). Namely, the overall trend in LASIK/PRK procedures is to use smaller irradiation spots (up to 1 mm diameter) to achieve better spatial resolution of photoablation^{42,43}, keeping the average fluence near the optimal value of 200 mJ/cm². The average fluence of a flat top beam equals its maximum fluence, while the maximum fluence of a Gaussian beam is 2 times larger than its average fluence. Except for the taper, added to the flat top pressure profile, the fluence profile of the beam for the fluences between 50 mJ/cm² and 600 mJ/cm² maps to the pressure profile linearly with the coefficient of about 0.5 bar/(mJ/cm²). Here, we assumed linear dependence of the initial loading peak pressure in the tissue to the laser fluence, experimentally determined by Krueger et al.⁵ and Spörl et al.¹⁷.

TABLE III. Characteristics of laser pulses of various contemporary excimer lasers

Company	Bausch & Lomb	Carl Zeiss Meditec AG	Alcon Surgical	Schwind eye-tech-solutions	Kera Harvest	Nidek	Johnson & Johnson Vision
Product	TECHNOLAS TENE0 317 Model 2 (M ²)	MEL 90	Allegretto Wavelight EX500	Amaris 1050RS	IsoBeam D200	EC-5000 CX-III (Navex Quest)	STAR S4 IR
Type	Flying spot					Broad beam/slits scanning	
Beam size/diameter [mm]	1	1.2 (0.7 FWHM)	1	0.9 (0.54 FWHM)	two spots 0.6 × 0.8	1.0 spot & slit 2 × 10	0.65–6.5
Shape	Soft spot (truncated Gaussian)	Gaussian	Gaussian	Super-Gaussian	NA	Flat top	Flat top, slits
Maximum repetition rate [Hz]	500	500	500	1050	450 (2 × 225)	50	1.5–20
Average fluence [mJ/cm ²]	200	150	200	200–500	100–180	300	160
Pulse duration [ns]	5–11	5	7	10	21–26 μs	25	20

Both, the broad beam delivery and the flying spot approach are still in use in the contemporary ArF (193 nm) excimer laser systems for LASIK/PRK. Table III presents a list of these lasers with an emphasis on the properties of the delivered laser pulses. Five of the contemporary excimer lasers are of the flying spot type (columns 1–5) while two (columns 6 and 7) use broader beams in combination with slits or variable iris diaphragms.

To interpret Figs. 4(a)–4(c) properly, it has to be noted that in the simulations and experiments, the input fluence was taken to be constant. This means that the critical volume is smaller for smaller irradiation spots primarily as a consequence of the total input energy being smaller (see Table II). Due to the general linearity of the AEM, the results presented in Fig. 4 are scalable.

Generally, regarding the shape and the absolute pressure values inside the critical volumes depicted in Fig. 4, the negative pressure in the broad beam cases (Figs. 4(a)–4(c)) is concentrated near the lens-vitreous boundary and a wave with a non-negligible negative amplitude eventually even reaches the retina, relatively far from the targeted area. On the other hand, the shape of the critical volume in the case of smaller spots, flat top or Gaussian [Figs. 4(d)–4(e)], is more elongated. Using the flying spot approach, the negative pressure reaches its highest values closer to its acoustic origin, extending from within the cornea through the aqueous humor up to the anterior part of the lens but does not extend as far as the retina.

The repetition rate of the contemporary flying spot lasers for corneal refractive surgery (up to 1 kHz) is too small for the two consecutive pulses to yield any superposition-based enhancement of the two pressure fields, since it takes about 20 μ s for the single-pulse pressure wave to exit the eyeball into the neighboring tissues, including any significant reflections, while the time interval between the pulses is 1 ms or more. Thus, repetition rate has no additional effects on the pressure wave focusing. In this respect, each laser pulse photoablation can be thought of as a separate event with no correlation between the consecutive pulses. However, if one of the laser pulses gives rise to an inertial cavitation bubble, the remnant gas bubble has not got enough time to escape the focusing volume of the subsequent pressure pulse. This remnant gas pocket effectively lowers the inception threshold for the subsequent cavitation structures that form in front of the bubble, because the reflection of the initial positive phase of the pressure wave interferes with the incident negative phase (see Figs. 3(a)–3(c) for the shape of the pressure wave), thus enhancing the peak negative pressure⁴⁴. The time interval between laser pulses is long enough for the oscillatory dynamics of the inertial cavitation bubble to die away, thus we believe that the inertial bubble dynamics is independent of the repetition rate. Moreover, it is known that interaction of the pressure pulse with the already existing gas bubble located in the vicinity of (or in contact with) delicate tissues may result in cellular damage⁴⁵.

While it is important to know the risks that arise from the mechanical effects and the conditions that bring about such risks, the interpretation and assessment of the implications of the presented analysis must be left to professionals in the ophthalmologic community.

CONCLUSION

The propagation of the photoablation-induced mechanical wave was described in detail with the help of a numerical acoustic eye model (AEM). The AEM has been validated by comparing the simulated pressure fields to the existing, but scarce experimental data. The propagation of the acoustic wave, generated by a broad beam pulse, can be best followed if the field is decomposed into the central monopolar compression and the hemi-toroidal edge wave. The effects of acoustic attenuation, focusing, phase-conversion, interference and reflections at tissue interfaces add to the complexity of the temporal evolution of the initial ultrasonic transient.

The comparative analysis of the existing experimental data on the evolution of the pressure fields soon after the laser pulse application in the photoablation of cornea reveals the existence of a focal volume where the peak negative pressure exceeds various published threshold values, potentially leading to tissue damage. This volume depends on the size of the irradiation spot and the input laser fluence. It is elongated along the optical axis of the eye and extends from the interior of the ocular lens to the vitreous humor. Until the risks due to this effect are objectively determined by ophthalmologists,

a combination of large spot diameters with large fluences, such as in the wide area (broad beam) approach to photoablation, should advisably be avoided.

A future step towards reducing the possible collateral effects due to the focusing of acoustic shock waves might also be a reduction of pulse duration into the picosecond or femtosecond range. This can be achieved with frequency-quintupled solid-state lasers. The photoablation threshold for sub-nanosecond pulses is lower than for the classical, few 10-ns-long ArF excimer pulses which in turn yields shock waves of up to 10-times-smaller amplitudes if femtosecond pulses are used instead^{9,46}. Lowering the fluence from the efficiency sweet spot near 200 mJ/cm² closer to the 50 mJ/cm² photoablation threshold would also beneficially reduce the amplitude of the shock waves. Alternatively, a novel laser-based refraction corrective mechanism was recently introduced⁴⁷, reported to be non-invasive even in the sense that it produces no damaging shock waves. Similarly, the existing small incision lenticule extraction (SMILE) technique⁴⁸ is also free from generating high-amplitude pressure waves.

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6.5 mm spot diameter

p [bar]

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$t = 4.85 \mu\text{s}$

6.5-mm spot diameter

3.0-mm spot diameter

1.5-mm spot diameter

Broad beam (flat top)

(a) 6.5-mm spot

(b) 3.0-mm spot

(c) 1.5-mm spot

Flying spot (flat top vs. Gaussian)

(d) 1.0-mm spot - flat top (e) 1.0-mm spot - Gaussian

$p < -7$ bar

$p < -40$ bar

$p < -200$ bar

$p > 800$ bar

z

25 mm