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MINIMIZING THERMAL DAMAGE IN CORNEAL ABLATION WITH SHORT PULSE MID-INFRARED LASERS

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ABSTRACT

Photospallation is proposed as the primary mechanism behind our recent animal studies involving corneal ablation by nanosecond-pulse mid-IR laser beams. Following a brief summary of earlier work directed to refractive procedures in the mid-IR, a preliminary analysis is presented, based on simple one-dimensional models of thermoelastic expansion developed previously. The results of the analysis indicate that front surface spallation is consistent with the striking tissue ablation characteristics observed in our recent *in vivo* work with short pulse beams, including very small ablation rates and submicron thermal damage zones. This is attributed to the fact that spallation is a mechanical—rather than a thermal—mechanism, which allows tissue to be removed in small layers at fluences far lower than those used in the earlier corneal studies with mid-IR beams, typically under 200 mJ/cm^2, resulting in minimal heating of tissue. Unlike prior work in the area of photospallation, we also suggest that the existing theoretical basis supports the use of nanosecond pulses as an effective approach to achieving controlled ablation in the presence of very high absorption. We further suggest that such domain of operation may be preferred over shorter pulses, both from a practical standpoint and to mitigate against potential damage from shock waves. [S1083-3668(99)00204-X]

Keywords mid-infrared; corneal ablation; photorefractive surgery; photospallation; thermoelastic; stress; photothermal mechanism.

1 INTRODUCTION

Ablation at mid-infrared wavelengths has often been suggested as a potential alternative to UV excimer lasers currently used for performing corneal refractive surgery.[1] Radiation near 3.0 μm and particularly at 2.94 μm was considered to be especially promising because of a fortuitous correspondence with the absorption peak of water, which constitutes about 80% of the cornea.[1–3] This results in very small absorption depths in corneal tissue, raising the prospects of achieving clinical effects similar to those observed with the excimer, but using a potentially lower cost, more compact solid state laser system to perform the surgery. The underlying premise is that shallow penetration depths would directly correlate with precise tissue removal and minimal collateral thermal damage, potentially less than 1 μm wide. This is important because submicron ablation rates and correspondingly small damage layers are believed to be key attributes behind the clinical success of current excimer laser based corneal sculpting systems.

To date, however, the quest for such precise laser-tissue interaction characteristics in the mid-IR has proven somewhat elusive. In particular, limiting thermal tissue effects to the desired small levels turned out to be an especially difficult task for mid-IR lasers.[3–5] It appears, in fact, that relatively extensive damage zones may be inherent to the photothermal nature of the ablation mechanism involved in laser-tissue interactions at mid-IR wavelengths. In this rather common process, heat is deposited in the tissue until it begins to vaporize through thermal dissociation of molecules. The forces for ejection of tissue can be rather large, caused as they are by a phase change from solid to gas, i.e., vaporization. Ablation through evaporization is a fundamentally explosive process and is therefore notoriously difficult to contain enough to achieve small and controlled tissue removal rates. Furthermore, because the ablation process involves elevated temperatures, the residual tissue surface is left generally near 100°C, the boiling temperature of water, which causes thermal denaturization of adjacent tissue layers, leading to large collateral damage zones. Thus, compared with a photodecomposition process induced by high energy UV photons, a photothermal process is likely to lead to inherently larger tissue effects. This expectation has largely been born out in experiments utilizing free-running or long pulse mid-IR radiation.[5,6]

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It has also been suggested that using shorter pulses could help reduce the extent of thermal damage seen in tissue with mid-IR lasers.[1,7,8] The hope was that energy deposition in a pulse that is short relative to the thermal relaxation time (about 2 μs in corneal tissue) would produce faster heating and less explosive evaporation of tissue. Indeed, experiments with Q-switched erbium-doped lasers such as Er: yttrium–aluminum–garnet (YAG) and Er:YSGG[7–9] indicated that with pulses that are about 100 ns long (as compared with hundreds of microseconds for the free-running laser), thermal damage zones could indeed be reduced to widths ranging from just under 2 μm to as high as 10 μm. To date, the shortest pulses—about 8 ns—were reported in investigations conducted with Raman-shifted Nd:YAG laser at 2.92 μm.[10] Yet, even in these experiments, 2–4 μm thermal damage layers were observed (with some zones as small as 1.5 μm seen in selected sample locations). Thus, while previous work indicated a degree of correlation between shorter pulsewidths and reduced damage zones, these experiments were not able, even at the shortest pulses, to demonstrate results which approach the submicron tissue effects typically achieved with excimer lasers. Thus, even though ablation rates on the order of 1 μm / pulse could be demonstrated, mid-IR lasers still caused larger than expected collateral thermal damage to remaining tissue. Clearly, imposing a condition that the pulses be shorter than the thermal relaxation time is not, in and of itself, sufficient to control the spatial extent of the interaction zone. While using short pulse radiation may help in confining the thermal energy to a smaller zone, a tissue removal process involving evaporation (even a gentler one) could still leave a residue of thermal damage larger than the amount of tissue actually removed. Consequently, it was not clear whether further reductions in pulse widths would ever result in the desired submicron levels of tissue damage, as long as a thermal process involving tissue vaporization is involved.

Prior experiments also left a legacy of considerable ambiguities regarding the preferred energy densities from mid-IR lasers. In particular, the extent of damage seen did not clearly correlate with either ablation thresholds or operating fluence levels. Thus, thresholds for ablation with nanosecond pulses are typically quoted as between 150 and 250 mJ / cm^2. Presumably lower thresholds would translate into lower operating fluences and hence less damage. This, however, was not always the case. For example, ablation thresholds of 250 mJ / cm^2 measured in experiments conducted using the shortest pulses of 8 ns[10] were among the highest, yet observed damage levels down to 1.5 μm were the lowest. On the other hand, the early studies with 50 ns pulses from a HF laser (which emits at several wavelengths between 2.7 and 3.0 μm)[2] reported very low thresholds of less than 100 mJ / cm^2 under certain conditions. Yet the tissue damage observed in the same work was quite high, with damage layers ranging up to 20 μm.

In general, analysis of results to date does not reveal a consistent pattern between pulse duration, fluence levels, and tissue damage zones. Such ambiguity in dosimetry puts even short pulsed mid-IR lasers at a clinical disadvantage when compared with excimer lasers, for which ablation rates in corneal tissue are fairly predictable and submicron collateral damage layers are consistently and readily achieved.

Recently we reported[11,12] on a series of animal studies in which radiation from short pulse mid-IR lasers was used to evaluate corneal ablation characteristics, tissue healing response, and refractive changes. Results obtained from these experiments, which included detailed investigations of tissue histology and morphology using light microscopy (LM), were found to be superior to any previously demonstrated in the IR—in terms of both the ablation precision and the minimal, submicron thermal damage levels observed. Tissue effects were, in fact, demonstrated to be equivalent, and in some cases even superior, to those achieved with the excimer—a major breakthrough for mid-IR lasers. Because of the near absence of any observable thermal effects we attributed the observed ablation features to a nonthermal, stress related mechanism known as photospallation. The use of short pulse laser radiation to induce ablation in absorbing tissue at relatively low fluence levels and with negligible thermal side effects has been suggested by a number of researchers including Dingus and Scammon[13] and by Dyer and Al-Dhahir[14] almost a decade ago. However, to our knowledge, the observations reported in Refs. 11 and 12 constitute the first direct evidence that a photospallation process may be involved in ablation of corneal tissue using nanosecond laser pulses from a practical mid-IR laser system. In the remainder of this paper we provide some preliminary analysis to support this conjecture, consistent with experimental observations. We also provide preliminary criteria for selecting optimal laser and system parameters to favor photospallation over other competing processes such as photothermal evaporation and photodisruption.

2 PHOTOSPALLATION—OVERVIEW AND QUALITATIVE ASPECTS

As an ablation mechanism, photospallation is the cleavage of tissue due to thermoelastic stress (or pressure) transients created following strong absorption of short pulse incident radiation. By nature, this constitutes a photomechanical process, distinctly different from either photothermal evaporization or photochemical decomposition. Yet, photospallation is also distinguished from photodisruption—another well known short pulse photomechanical ablation process—but one which occurs at higher intensities, even in the absence of any absorption of light. A front surface photospallation process in tissue was described in detail by Dingus and Scammon[13] who developed a simple model to estimate conditions for front surface spallation. Theoretical descriptions of the process were also given by Dyer and Al-Dhahir[14] who performed some of the first measurements of the ablative stress transients in corneal tissue. More detailed qualitative descriptions of spallation were given by Jacques[15] and by Dingus et al.[16] and numerical modeling provided more accurate quantitative descriptions of the process[17–19] in several types of biological tissue, including the cornea. Numerous experiments were also conducted by a number of different groups[20–24] which helped confirm some key predictions of these models. Most of the testing conducted to date used a variety of laser wavelengths and pulse durations to perform time resolved photoacoustic measurements of stress wave forms and/or the cavitation effects associated with the spallation process.

The basic steps associated with a front surface spallation process are qualitatively well understood and have been described by many previous papers. The front surface layer of the material is first heated by a laser beam over a time scale much shorter than the thermal relaxation time. Because the material is heated before it can thermally expand, rapid compression results, followed by dynamic expansion. A compressive (positive) stress pulse propagates away from the deposition region both toward the front surface and in the opposite direction. If this stress wave encounters a surface with lower acoustic impedance (e.g., a free surface) then a tensile (negative) stress wave will be reflected back into the material. The tensile pulse trails the compressive pulse into the material thus causing a characteristic bipolar stress pulse. If, at some depth, the tensile stress exceeds the tensile tissue strength, the material will begin to fracture at that depth. For sufficiently large tensile amplitudes entire layers or fragments of tissue will ''spall,'' i.e., be completely ''split off'' and ejected.

Even though both photospallation and photothermal evaporation are precipitated by absorption of light, the ablation process that follows is fundamentally different for the two processes. In terms of the potential effects on tissue, the more striking distinguishing characteristics between the two mechanisms are as follows:

a. Spallation can be much faster than a thermal process for which temporal confinement can never be perfect, even for short pulses. This is because thermal gradients are cumulative in time, i.e., they continue to develop well beyond the period over which heat was directly deposited into the material. By contrast, spallation occurs at a specific moment in time, namely, at the instant when tissue strength is exceeded. The process then stops abruptly and residual stresses are dissipated with little, if any, effect on tissue.

b. Photospallation is also better confined spatially than a purely thermal process, which acts over larger distances and whose effects are cumulative in space (and time). This is a direct consequence of the mechanical nature of the spallation process whereby tissue fragments are torn apart and immediately ejected, resulting in a ''clean break.''

c. For short enough pulse widths, tissue can spall before it evaporates, hence ablation can take place without any phase change in the material. This extends the ablation threshold to much lower fluences compared to a thermal process where a latent heat of vaporization must first be overcome for ablation to proceed. Because of the low fluences, less total heat is deposited into the target and the remaining tissue does not desiccate.

d. With photospallation, thermal effects are an undesirable by-product, not the primary ablation process. Since more of the radiation fluence is carried off in material fragments, the temperature in the residual target material may be low enough to avoid even mild denaturization, making this a truly ''cold'' ablation process.

Thus, photospallation-induced ablation has the potential to produce drastically smaller collateral thermal damage zones than a thermal process. This aspect makes the spallation mechanism a nearly ideal match to the exacting requirements of refractive surgery. Indeed, suggestions were often made in the literature as to the potential applicability of this process to corneal ablation.[13–17] The role of thermoelastic stress and pressure transients in short pulse ablation has also been verified in many different test media including liquids, biological tissue, and tissue-like materials.[14,21–24] However, prior to our work reported in Refs. 11 and 12 there were no reliable *in vivo* experimental data to support viability of such a mechanism in practical clinical settings, especially for actual refractive surgery. Furthermore, most previously suggested applications to corneal ablation[13–19] emphasized the desirability (and sometimes the necessity) of operating in the so-called ''stress confinement'' regime, achieved by using pulse durations that are short relative to a characteristic acoustic time, defined as the time it takes the speed of sound to traverse the penetration depth (or optical zone). For the strongly absorbed beams from mid-IR lasers, optical zones can be only a micron or so long. Hence, stress confinement can only be realized subnanosecond pulses. However, requiring picosecond pulse lasers of significant energy in the mid-IR is a condition that existing technology still cannot readily meet.

The experiments reported by Telfair et al.[11,12] were, to our knowledge, the first to reveal clear evidence of photospallation in corneal tissue *in vivo*, using a mid-IR beam to induce measurable refractive changes in animal eyes. Furthermore, these investigations were conducted with lasers that had nanosecond pulse durations, the usefulness of which was not well anticipated from most prior work in photoacoustic phenomena, as was stated above. In the next section, we show that previously developed theoretical models of spallation do, in fact, predict that a photospallation mechanism could be readily induced by nanosecond pulses from highly absorbed mid-IR lasers, under the right conditions.

3 PHOTOSPALLATION MECHANISM—THRESHOLD ESTIMATES

The magnitudes of the fluence thresholds required for front surface spallation can be estimated based on the simple one-dimensional model described by Dingus and Scammon.[13] In this model, the amplitude of the stress waves is calculated using the equations governing thermoelastic expansion of tissue, subjected to irradiance by highly absorbed, short pulse radiation. We applied the model to the particular case of corneal tissue exposed to mid-IR laser radiation with pulses in the nanosecond range and assuming a beam output typical of a refractive laser system operating in a scanning mode. In this case, a nominal beam spot size on the cornea would be on the order of 1–2 mm. With an absorption depth only about 1 μm deep, the exposure area is several orders of magnitude larger than the absorption depth and a one-dimensional approximation is appropriate. Further assuming that absorption follows the usual exponential Beer's law, the initial distribution of the compressive stress induced by thermoelastic expansion following a laser impulse in an absorbing tissue is given by the expression[13]

$$\sigma(z)=\Gamma\mu F_0 e^{-\mu z}, \tag{1}$$

where σ is the stress in J/cm^3, Γ is the (dimensionless) Grüneisen coefficient, F_0 is the incident beam's radiant exposure, or fluence, in J/cm^2, and μ is the tissue's absorption coefficient in cm^{-1}.

Upon reflection at the cornea/air interface, a bipolar stress wave results propagating in the $+z$ direction, with the magnitude of both positive and negative peak stress amplitudes achieving the same maximal value. The stress wave forms can be convolved with the time profile of the laser pulse to yield the stress response to the laser pulse as a function of time and propagation distance. In Ref. 13 analytic expressions were derived for these stress wave forms in the simplest case of a rectangular profile of duration t_p, equal to the nominal pulse duration. For this case, it was shown that the peak magnitude of the tensile stress developing in corneal tissue could be expressed as a simple function of the pulse duration of the form

$$\sigma_{\max}=A\Gamma\mu F_0/2, \tag{2}$$

where A is a ''stress dissipation factor'' given by the simple expression

$$A=(1-e^{-\tau})/\tau \tag{3}$$

with $\tau=t_p/t_0$ and $t_0=(\mu v_s)^{-1}$, v_s being the velocity of sound in the tissue. The factor t_0 is known as the ''stress relaxation time'' representing the time it takes for an acoustic wave to propagate over the absorption depth $1/\mu$ as alluded to earlier. Thus, the factor A decreases exponentially as a function of τ from a maximum value of unity. The two limiting cases of interest are

$$\sigma_{\max}\rightarrow(1/2)\Gamma\mu F_0, \qquad \tau\ll 1 \tag{4}$$

and

$$\sigma_{\max}\rightarrow(1/2)\Gamma F_0/(v_s t_p), \quad \tau\gg 1. \tag{5}$$

Equation (4) corresponds to the stress confinement region where the pulse duration is smaller than the characteristic time t_0. This is the case considered by authors of a number of previous studies[13–17,20–22] to be ideal for spallation in tissue since it allows maximum stress amplitudes to be achieved for the lowest fluence values. However, as was noted above, when the absorption is very high, as is the case of interest here, the value of t_0 becomes very small, leading to difficult—and possibly unnecessary—requirements for the laser pulse duration. By way of a more practical alternative, it is our contention that, given sufficiently large absorption, nanosecond pulses can give rise to peak stress amplitudes that are high enough to fracture the tissue, even with some stress dissipation. This effect may be further enhanced by the possibility that tensile stresses may develop during the pulse itself, reaching magnitudes that are sufficiently high to cause spallation in a shorter time scale than predicted by the model with a rectangular pulse input. This is a likely scenario when the optical zone is so small that the acoustic travel time across it is a fraction of the pulse duration, the later being quite small already. Expectations based on prior analysis where the possible onset of spallation during the pulse was not fully taken into account have to then be modified, making the analysis considerably more complicated. Nonetheless, it may still be possible to approximate the magnitude of peak stresses using Eq. (5), but with an effective pulse duration that is shorter than standard full width at half maximum (FWHM) values would imply. In particular, there are indications that, under certain conditions, the rise time of the pulse may be a more appropriate parameter to use in Eq. (5) than the FWHM effective pulse width (see below).

In the specific case of interest to us, mid-IR laser beams are used at or around the 2.94 μm wavelength matched to the peak of the water absorption curve. With an absorption coefficient near 10^4 cm^{-1}, the absorption depth is confined to an extremely small region of only about 1 μm in corneal tissue. Assuming that the speed of sound in cornea is about the same as it is in water, namely, v_s

$=1.5\,\text{ns}/\mu\text{m}$, we obtain $t_0 \approx 0.7\,\text{ns}$. Therefore, the stress confinement region of Eq. (4) can be achieved only for subnanosecond pulses. Nonetheless, from Eq. (5), we see that substantial stresses can still develop even for pulses that are in the nanosecond range, i.e., $t_p > t_0$, despite some stress dissipation. For example, when $t_p = 7$ ns ($\tau \approx 10$), we find that with fluence levels as low as 30 mJ/cm^2, $\sigma_{\max}$ can be as high as 20 bars, assuming $\Gamma = 0.15$, the same value used in Ref. 13. This may be sufficient to cause tissue spall in the cornea, which is thought to have spall strengths in the 2–20 bar range. Furthermore, in keeping with comments made above regarding the likelihood of spallation occurring during the pulse, it may be more appropriate to use a shorter effective pulse duration, resulting in still lower threshold fluences. We therefore believe that the values of the peak stress, and hence threshold fluences, as derived from Eq. (5) using typical FWHM pulse durations, should be considered an overestimate.

In addition, the value of 0.15 selected for the Grüneisen coefficient may be on the low side, since this factor is known to increase as a function of temperature.[13] This means that even lower fluences may be sufficient to generate the same high stress levels, depending on the extent to which the tissue temperature is elevated during pulse irradiation. This may help explain why, in the studies of Ref. 11, there was strong evidence for spallation even for an Er:YAG laser which had a nominal pulse duration of 70 ns. In this case, tissue was still consistently ablated at fluences down to just below 100 mJ/cm^2, indicating thresholds that are somewhat lower than this value (though still larger than the ones indicated for the shorter 7 ns pulses obtained from an optical parametric oscillator (OPO) at the same wavelength). A higher value of Γ for the longer pulse case could partially account for the relatively low threshold values inferred from the experiment. This is not too surprising, given that operating fluences were generally higher when the 70 ns pulse was used in the experiments. This could well cause a greater rise in the temperature of the heated tissue layer and hence higher effective value of Γ. A value for F_0 that is smaller than 100 mJ/cm^2 can be obtained from Eq. (5) for a 70-ns-long pulse assuming $\Gamma \approx 0.3$, i.e., about a factor of 2 larger than the value used for the 7 ns case. This estimate corresponds to a value of $\sigma_{\max}$ of about 10–15 bars, which may be sufficient for spallation in the cornea. The possibility of such relatively low threshold values may be more credible, given that the Er:YAG laser pulse shape has a relatively short rise time of only 20–25 ns, and this may be a better parameter to use in Eq. (5) than the FWHM effective pulse width, as was alluded to above.

As an example of the results expected from application of the model of Ref. 13 to the case where $\tau \approx 10$, Figure 1 shows stress wave forms plotted as a function of depth at several different times. Since with this model, only compressive (positive) stresses develop during the pulse, the times selected were longer than t_p. In this example, we assumed $\mu = 10^4\,\text{cm}^{-1}$, corresponding to a characteristic time $t_0 = 0.67$ ns. The pulse length is assumed to be about a factor of 10 larger, which is an adequate representation of the 2.94 μm output of the Nd:YAG/OPO of Ref. 11. As these plots indicate, conditions for relatively large negative stresses may be reached at times just a fraction of t_0 longer than the pulse duration (i.e., immediately upon termination of the pulse). Furthermore, large maximum amplitudes of the tensile stress can develop at depths that are smaller than the nominal absorption length (which is about 1 μm in this case). Gradients may also be sufficiently steep at certain locations to cause spallation, given the extremely small distances involved at such high absorption. Therefore, the simple model indicates feasibility of spallation at submicron depths. Of course, it should be realized that the results of the simple model should not be taken at face value, given the complex dynamics of the development and propagation of the stress waves, processes which may occur, in reality, during the laser pulse itself. In this respect, the model of Dingus and Scammon, with a rectangular step function laser profile as an input, is clearly an oversimplification.

One possible modification is the use of more realistic input pulse shapes. For example, a biexponential input pulse form was used by Dyer and Al-Dhahir,[14] whereby temporal pulse profiles are weighted towards the early part of the pulse, a shape similar to those produced by typical solid state (and excimer) lasers. Using similar biexponential profiles, characteristic bipolar stress wave forms can be most easily derived as a function of the delay time $t' = t - z/v_s$. Three such curves are depicted in Figure 2, representing pulses with different values of the rise time, but similar FWHM. These were all generated assuming the same absorption coefficient ($10^4\,\text{cm}^{-1}$) as before but a somewhat higher initial fluence of 0.05 J/cm^2. As Figure 2 shows, the maximum and minimum stress values are no longer symmetric, unlike the step function case of Figure 1. Thus, high values of compressive stress are reached very early in the pulse, whereas the smaller peak negative stress is reached later. Figure 2 shows, however, that sizable tensile stresses and sharp stress gradients can develop during the pulse. The magnitude of the peak tensile stress is seen to increase as the rise time decreases, as expected. For the case with the shortest rise time, the peak negative stress is reached at times $t > 5$ ns. Therefore the first spall plane can occur well before the end of the pulse at distances shorter than 1 μm from the surface.

The maximum stress amplitudes can be derived analytically using the expressions derived in Ref.

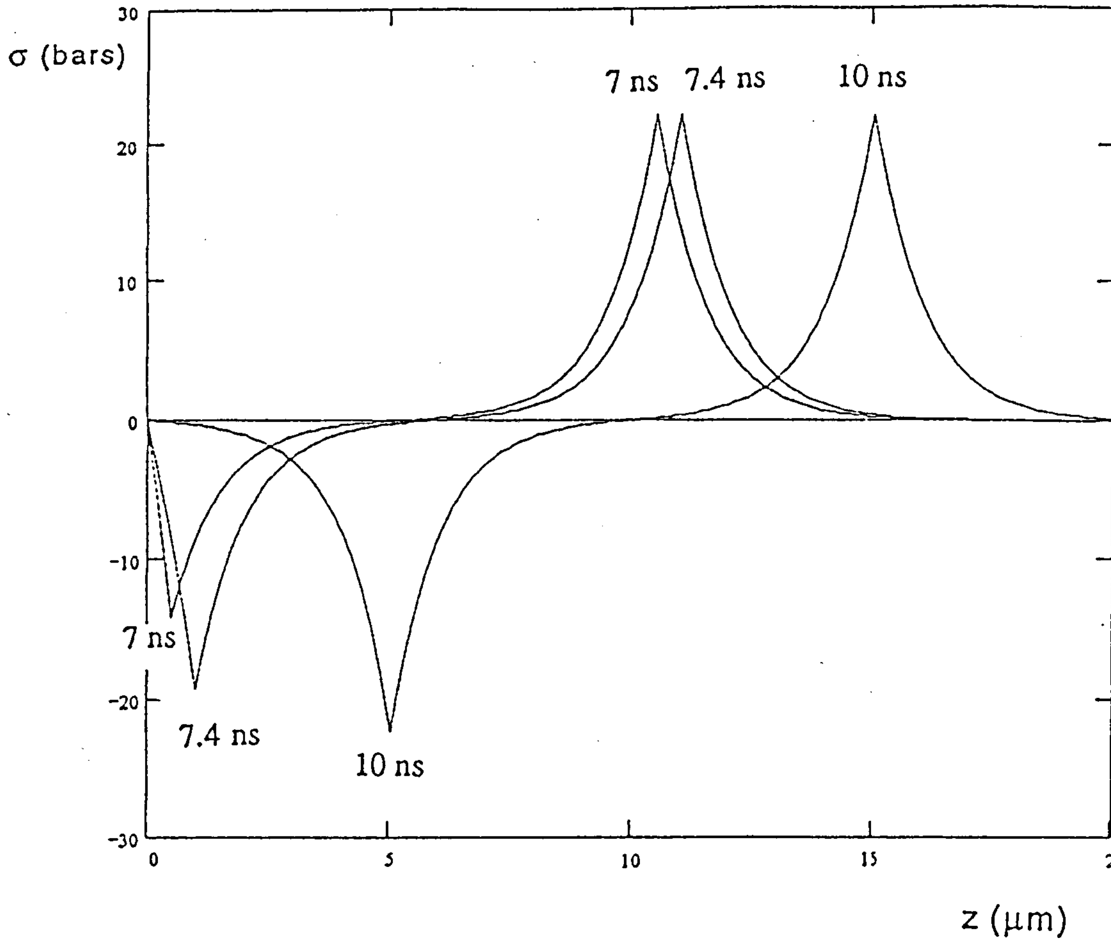

Fig. 1 Stress wave forms as a function of the distance z from the surface, calculated using the model of Ref. 13 for $\mu = 10^4\ \mathrm{cm}^{-1}$, $F_0 = 30\ \mathrm{mJ/cm}^2$, assuming a rectangular pulse input of duration $t_p = 10\ t_0$, where $t_0 = (\mu v_s)^{-1}$ with $v_s = 1.5\ \mu\mathrm{m/ns}$. The curves are shown at three different times just after the end of the pulse.

14. These are somewhat more complicated than the simple Eqs. (4) and (5) and are not given here. The key observation is that a one-dimensional model of spallation qualitatively indicates feasibility of achieving extremely small ablation depths with pulses of similar shape and duration to those used in the experiments of Ref. 11. It is hoped that additional computations, using more quantitative fits to the shape and magnitude of the real pulses, will be performed at a future time.

Regardless of the specifics of the input pulse profile, more precise estimates of spallation characteristics are difficult, given the large number of approximations involved in the computations used to derive the curves in both Figures 1 and 2. One simplification involves the assumption of a constant absorption coefficient. For example, in the case of the square pulse input, inspection of Eq. (5) reveals that the peak stress amplitude is independent of the absorption coefficient. Thus, the magnitude of the peak stress may be independent of wavelength when the condition $t_p \gg t_0$ holds. For pulse durations on the order of 10 ns, this would seem to imply that wavelengths in the range 2.7–3.1 μm ($\mu > 5000\ \mathrm{cm}^{-1}$) may all have an equivalent effect on tissue. Such a conclusion is not well supported by experiments. For example, it is well known that the absorption coefficient changes with temperature, a fact that was ignored in the derivations leading to Eq. (5) as well as in the computations of Figure 2. It was also suggested that variations in the absorption coefficient may be more significant for higher fluence levels, as was extensively discussed by Walsh and Cummings,[25] who postulated that the effective absorption peak of water shifts to shorter wavelengths as the temperature of the tissue rises. It is, however, not clear to what extent such a shift occurs at the lower fluences of interest in the present case, given the milder rise in temperature of the affected tissue. Further experimentation as well as more precise modeling would have to be undertaken to better elucidate the role of wavelength variation across the mid-IR for the lower fluences applicable to photospallation in the cornea.

Another issue which was not addressed thus far is the possibility of creating multiple spall layers.

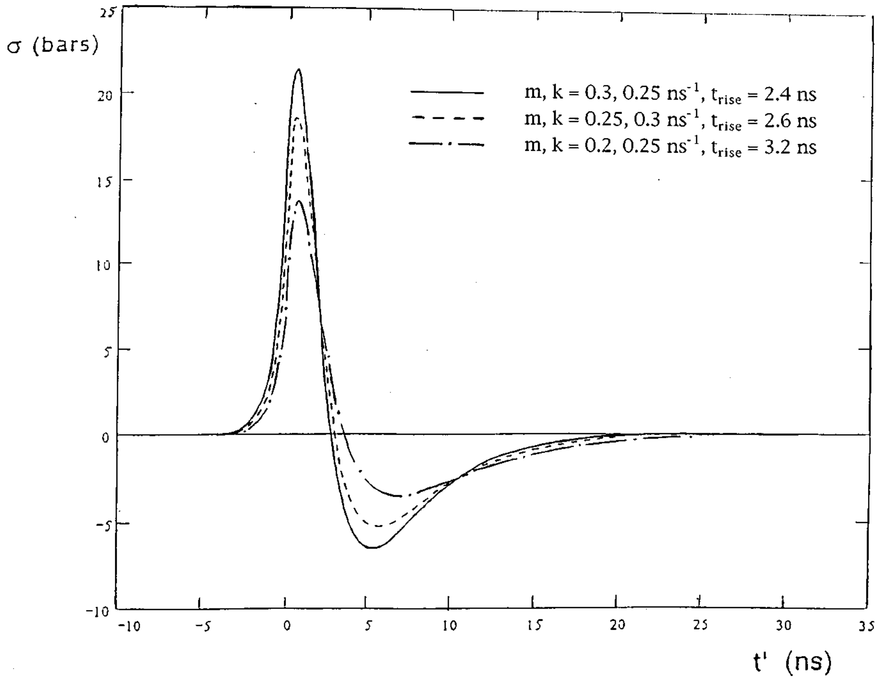

Fig. 2 Variation of the thermoelastic stress wave with the retarded time $t' = t - z/v_s$, with z the distance from the surface for different values of the pulse rise time. Following Ref. 14, the curves shown were generated using a temporal bi-exponential laser pulse of the form $f(t) = (1 - e^{-kt})e^{-mt}$ with k and m selected to produce the rise times indicated. For all the calculations, the absorption coefficient μ, was set to $10^4\ \mathrm{cm}^{-1}$, the incident fluence assumed was $F_0 = 50\ \mathrm{mJ/cm}^2$, and other parameters were as defined in the text. Note that a negative tensile peak starts developing near $t' > 3$ ns, peaking at about 5 ns for the bottom curve. Therefore spallation can occur at times $t > 3$ ns, i.e., well before the end of the pulse.

Secondary spalling can occur because the residual compressive stresses can be substantial even after spallation occurred. This residual wave continues to propagate towards the front surface, reflecting, possibly, a second time from the new boundary created when the first surface layer spalled away. If the reflected tensile stress is of sufficient magnitude to build up to the tissue spall strength again, then a secondary spall can occur, as was suggested in Ref. 13. This would effectively increase the ablation rate per pulse, and also introduce an element of unpredictability into the procedure. Such a scenario is likely at higher fluences (applicable, generally to longer pulses). The potential for multiple spalling may therefore provide another reason to keep the laser fluence to levels that do not exceed the threshold fluence by an overly large amount (the first reason being, of course, to minimize any undesirable thermal effects). To better analyze the possible effect of secondary spalling on the ablation rate also requires more precise numerical computations which are beyond the scope of the present work.

4 Summary and Conclusions

In this paper we have outlined a possible theoretical basis to explain observations reported in Refs. 11 and 12 for a short pulse mid-IR photorefractive ablation system, which revealed features far more similar to those achieved with excimer lasers in the UV than to any reported in previous work with mid-IR lasers. We believe that spallation is the only way to explain the striking outcome of submicron thermal damage observed with two mid-IR lasers of different pulse parameters. Photospallation is a nonthermal process requiring low levels of input energy because the ablation fluence thresholds are much lower than with a photothermal evaporation process. The resulting ablation rates can be very small—less than 1 μm/pulse—possibly resulting in tissue removal characteristics quite similar to what is obtained with excimer lasers. This mechanism ablates by peeling off very small layers without vaporization. Collateral thermal damage levels are therefore likely to be also very small because a large fraction of the deposited thermal and kinetic energy is carried away with the ejected mass, leaving the tissue "cool." This is consistent with the submicron thermal damage zones observed in the above experiments—less than 0.5 μm for the 7 ns pulses from the Nd:YAG/OPO laser. It is also consistent with observations of the dependence of the thermal necrosis zone on pulse duration, with damage layers extending further for the longer 70 ns pulses.

For this work we applied methods of analysis developed by a number of groups to the case of corneal ablation with highly absorbed, nanosecond pulse mid-IR laser beams. In fact, we found it remarkable that even with the simplest expressions, derived using numerous approximations, estimates obtained for fluence thresholds were in good qualitative agreement with observations, even for two pulse durations that differ by nearly an order of magnitude. The simple analysis supports our contention that in the mid-IR, photospallation can be effectively triggered in corneal tissue with nanosecond pulses and at energy levels that are within the current state of the art for practical lasers. In this, we differ from many prior discussions of photospallation and photoacoustic phenomena which emphasized the importance of using pulses that are shorter than the characteristic stress relaxation time. In the case of mid-IR beams tuned to the highest absorption, this would imply picosecond pulses. Yet we believe that, under the right conditions, consistent spallation can be obtained with longer (nanosecond) pulses, with the process being more effective for faster pulse rise times. In this case, maximum tensile stress values can be achieved early in the pulse and at very shallow depths, as desired. Furthermore, since spallation may well occur during the laser pulse, the ejected material can possibly provide some shielding from residual light photons that would otherwise deposit more heat in the medium. This last conjecture could explain the surprisingly small thermal damage levels seen in experiments with longer 70 ns pulses, where the temperature rise in the affected tissue appears to be lower than expected on the basis of operating fluence and absorption parameters alone. Direct experimental testing and further analysis is required, however, to confirm this last suggestion.

Regarding the issue of the optimal pulse width for a photospallation process in tissue, a few more comments are in order. In particular, we note that pulses that are longer than 1 ns or so may be preferred also from the viewpoint of avoiding potentially deleterious effects due to plasma formation and the attendant shock waves. The phenomenon of induced plasma, material breakdown, and photodisruption in ocular tissue due to intense laser radiation has been extensively studied in the past decade. The recent experiments and analysis by Vogel et al.[26] provide a particularly good review of the current understanding in this field, including numerous references and an extensive comparison of results obtained by various groups. In this, as well as in most other prior investigations, interest has been directed to ablation and cutting of tissue that is transparent to the incident radiation. In order to overcome thresholds for the photodisruption process, high power energy densities are required in this case, necessitating, typically, the use of tightly focused beams. However, there is evidence that in highly absorbing tissue, lower fluences may be sufficient to cause breakdown. In particular, as shown in the study by Oraevsky et al.,[27] energy densities of less than 200 $\mathrm{mJ/cm^2}$ could induce breakdown in highly absorbing collagen gels when power densities were just over $10^9\,\mathrm{W/cm^2}$. With the cornea even more absorbing in the infrared than the collagen gel used in the above work, there is the possibility that plasma can be formed even at lower energy densities. Since power densities on the order $10^9\,\mathrm{W/cm^2}$ can be readily reached with picosecond pulses, even for spot sizes as large as 1–2 mm, there is a finite probability of triggering plasma-mediated processes in the irradiated tissue. This could interfere with the desired spallation process and also generate shock waves, which may be injurious to tissue in locations remote from the surgical site. Use of longer nanosecond pulses thus eliminates the undesirable prospect of this additional collateral tissue damage mechanism.

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