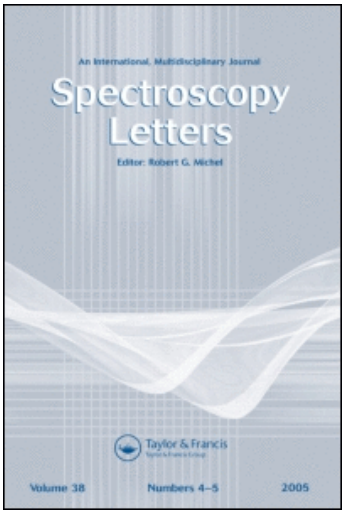

Spectroscopy Letters
Publication details, including instructions for authors and subscription information:
http://www.informaworld.com/smpp/title~content=t713597299

Non-Thermal Laser Ablation Model for Micro-Surgical Applications

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To cite this Article Sentrayan, K. , Thorpe Jr., A. and Trouth, C. O.(1998) 'Non-Thermal Laser Ablation Model for Micro-Surgical Applications', Spectroscopy Letters, 31: 3, 559 — 572
To link to this Article: DOI: 10.1080/00387019808002749
URL: http://dx.doi.org/10.1080/00387019808002749

SPECTROSCOPY LETTERS, 31(3), 559-572 (1998)

NON-THERMAL LASER ABLATION MODEL FOR MICRO-SURGICAL APPLICATIONS

Key Words: Ablation Depth, Polymethyl Methacrylate, Polyimide, Thermal Relaxation, Photochemical, Photothermal

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ABSTRACT

We have developed a non-thermal laser ablation model which may reduce thermal damage to neighboring structures. Based on this model, the three critical parameters for a well controlled non-thermal microsurgery are (1) the laser wavelength with its photon energy matching closely the bond dissociation energy, (2) the energy fluence must be above threshold to avoid thermal process due to non-radiative relaxation from the excited electronic states to vibrational, (3) ultra short laser pulses (few fs) to completely eliminate thermal and direct biomolecular reactions. In this model the UV laser photon dissociates the molecular bonds which leads to the splitting of longer polymer chains into small fragments. The excess energy if any may appear as kinetic energy in the polymer-fragments. The extreme rapidity of the bond breaking process reduces heat conduction. The model establishes a relationship between ablation depth per pulse, the absorption coefficient, the incident laser energy fluence, and the threshold energy fluence. The ablation depths per pulse were calculated for the polymers Polymethyl methacrylate (PMMA) and polyimide for various commercially available UV lasers. It has been found that the minimum ablations depth occurs at 193 nm for both PMMA and polyimide. This assures a well defined

incision with minimal thermal damage to the surrounding structures at this wavelength. There exists a definite threshold energy fluence for non-thermal ablation for any given biomolecule and below the threshold the non-radiative relaxation process may cause thermal ablation. New ultra fast lasers (few femtoseconds) (fs) will completely eliminate thermal diffusion as well as direct biomolecular reactions.

1. INTRODUCTION

In order to use the laser technology in surgery, it is very important to understand the problems associated with laser-tissue interaction. Interaction between lasers and biological tissue can be studied by relating the physical parameters of the biological tissue, such as absorption, scattering, reflection, density, thermal conductivity, and heat capacity to the parameters of the laser, namely wavelength, power, energy density, and pulse duration. Based on the time of interaction, effective power density and wavelength of lasers, three types of laser-tissue interaction can be distinguished, and these can be categorized as those having photochemical, photothermal and photo non-thermal effects. [1,2] Photochemical effects are caused by absorption of radiation from lasers with low power densities and their interaction with tissues for extremely long times, but without producing primary heating of tissue. Photodynamic therapy (PDT) is an example of photochemical effect and has been quite effective in the retention of a photsensitizer in malignant tumors and subsequently destruction of this tumor by irradiation with a laser of suitable wavelength and dose [3]. The PDT treatment in clinical trials in the area of dermatology [4,5] and gastroenterology [6] is reported.

Photothermal effects are caused by a laser of high power density interacting with tissue for a short period of time. Since the primary surgical applications of lasers are based on the conversion of laser radiation into heat, photothermal effects are widely utilized in surgery for tissue removal and tissue welding. [7-11] The degree and extent of the thermal effect depends on

the optical and thermal properties of the tissue, the geometry and energy density of the incident beam.

When the laser radiation power density exceeds 10^7 W/cm^2 the strong electric filed of the laser dissociates the materials involved. The non-thermal breaking of the intermolecular bonds due to the absorption of a high density of photons is called photoablation, which can be utilized to produce precise (smaller than 50μm) non-necrotic cuts. [12] When the laser power density exceeds 10^{10} W/cm^2 for a nanosecond-wide pulses and 10^{12} W/cm^2 for picosecond-wide pulses (with corresponding electric fields in the range 10^6-10^7 V/cm), laser-tissue interaction produces plasma by spontaneous ionization due to free electrons and ionized atoms. After a certain degree of ionization has been reached, the plasma undergoes sudden expansion accompanied by a mechanical (acoustic) shock waves. The shock waves either ruptures the tissue structure (photo-disruption) or disintegrates the target material (photo-fragmentation). The plasma induced ablation of brain tissue with pico-second laser pulses has been reported [13]. The non-thermal interactions, namely photoablation, photo-disruption/photo-fragmentation, have the following distinct advantages: (1) Unlike thermal models, the non-thermal model enables us to calculate precisely the applied radiation energy; (2) by properly selecting the laser wavelength, with its photon energy matching closely the bond dissociation energy, and choosing ultra-short pulses - with the pulse duration shorter than the thermal relaxation time of the tissue - well controlled non-thermal laser surgery can be performed with minimal damage to the neighboring healthy tissue structures.

Our objective in this study is to develop a theoretical model based on non-thermal photochemical ablation that will be directly applicable and can be tested for the processes that might be encountered in microsurgery. Non-thermal etching of polymers with high energy UV lasers in the area of material processing is well documented. [14-17] The principal features of the non-thermal etching due to UV laser pulse (distinct from visible of IR laser pulse) are the control

that can be exercised over the depth of the etching by controlling the number of pulses and the energy fluence of the laser and the absence of detectable thermal damage to the substrate. This allows the possibility of extending the process to biological tissue and subsequent development of a non-thermal photochemical ablation model for microsurgical applications. In this model, the high energy UV photons break the bonds of the biomolecules. If the intensity is large enough, the molecule will be clipped into small fragments before the laser energy has time to heat the tissue. If these fragments are small enough, they can simply fly away, carrying excess energy which was not used in bond breaking. The escape velocity of the fragments can be controlled by tuning the laser wavelength.

2. NON-THERMAL LASER ABLATION MODEL

2. 1. Principle

The basic principle involved in non-thermal ablation is that the laser photon energy is absorbed by tissue biomolecules and excites the electronic states that lie above the dissociation energy of the molecules. The dissociation of molecular bonds leads to the splitting of longer polymer chains into small fragments. Numerous bond breaks cause an increase of pressure inside the laser irradiated tissue volume, which causes the molecular fragments to escape from the tissue volume, which causes the molecular fragments to escape from the tissue. The quantum yield q for molecular bond dissociation is given by:

$$q = \exp\left[-(\hbar\omega_d / \hbar\omega_L)\right] \tag{1}$$

where $\hbar\omega_d$ is the bond dissociation energy and $\hbar\omega_L$ is the energy of a single laser photon. The quantum yields of molecular bond dissociation for various molecular bonds of biomolecules at different laser wavelengths are given in Table 1. It is clear that any appreciable quantum yield of bond dissociation in biomolecules can only be expected at laser wavelengths shorter than 200 nm.

TABLE 1
Calculated Quantum Yield for Biomolecular Bands at UV Laser Wavelengths

	Wavelength (nm)								
Bond	355	308	282	266	223	193	172	146	126
Structure	Quantum Yield								
C-C	0.36	0.41	0.44	0.46	0.52	0.57	0.60	0.65	0.69
C-H	0.29	0.34	0.38	0.40	0.46	0.51	0.55	0.60	0.65
CH-H	0.29	0.34	0.34	0.37	0.39	0.45	0.51	0.60	0.64
CH_2-⬡	0.28	0.33	0.36	0.39	0.45	0.47	0.51	0.56	0.61
C≡N	0.08	0.11	0.13	0.15	0.20	0.55	0.59	0.64	0.68
C=O	0.13	0.17	0.19	0.21	0.27	0.56	0.60	0.65	0.69
CH_2-H	0.25	0.30	0.33	0.36	0.42	0.65	0.72	0.82	0.91
CH=CH	0.12	0.16	0.18	0.20	0.26	0.33	0.37	0.43	0.48
O-⬡	0.25	0.30	0.33	0.36	0.42	0.47	0.57	0.59	0.61
N-C	0.43	0.48	0.51	0.53	0.55	0.61	0.68	0.69	0.75
O-H	0.16	0.30	0.33	0.36	0.42	0.45	0.49	0.54	0.59

2.2. Excess Energy

Any excess energy $\Delta E = \hbar(\omega_c - \omega_L)$ may appear as kinetic energy in the irradiated fragments. The extreme rapidity of the bond breaking process eliminates head conduction. If m is the average mass of the ejected fragments, then the escape velocity v_e is given by:

$$V_e = [2\,(\Delta E)/m]^{1/2} = [2\,\hbar\,(\omega_d - \omega_L)/m]^{1/2} \qquad (2)$$

The escape velocity can be varied by tuning the laser wavelength. When the laser energy exactly matches the bond dissociative energy, the escape velocity, v_e, will be zero (i.e., $v_e = 0$).

2.3. Ablation Depth/Pulse and Beer-Lambert Law

Assuming that the ablation is a two-step process in which the laser absorption is followed by material ablation, one can use Beer-Lambert Law [15] to establish a relationship between ablation depth/pulse [d_t], the absorption coefficient (α), the incident laser energy fluence (F_o), and the threshold energy fluence (F_T) and can be written as:

$$d_f = (1/\alpha) \ln (F_o / F_T) \quad (3)$$

It is interesting to note that Eqn.(1) does not contain any time dependent quantities. If the ablation proceeds layer by layer, then the depth of ablation and the laser intensity must carry time dependent components. In addition, if one considers the "bond shielding effect" the incident laser intensity also varies as a function of spatial coordinates during bond breaking.

2.4. Laser Pulse Duration and Thermal Damage

Another important parameter to be considered for non-thermal ablation is the pulse duration of the laser. In order to have minimum thermal damage due to heat diffusion, the pulse duration of the laser must be shorter than the thermal relaxation time (t_T) of the tissue and is given by:

$$t_T = d^2/\chi \quad (4)$$

where d is the optical penetration depth and χ is the ratio between the medium's ability to conduct heat and its storage capacity and is given by:

$$\chi = k/\rho c \quad (5)$$

where k is the thermal conductivity, ρ is the specific gravity and c is the specific heat per unit mass. The thermal relaxation times for various tissues at different laser wavelengths are calculated using Eqns. (4) and (5) and are given in Table 2. Though the relaxation time for bulk tissues is in the microsecond (10^{-6}s) range, the relaxation time among various vibrational modes is only a fraction of a nanosecond (10^{-9}s). The direct biomolecular reactions in which the breaking of old bond and the formation of new bond proceed over a time period of 10-100 fs [18].

3. RESULTS AND DISCUSSION

We have used Eqn.(3) to calculate the ablation depths per pulse for the polymers polymethyl methacrylate (PMMA) and polyimide at different laser wavelengths. The polymers (PMMA) and polyimide are selected in this investigation due to their dramatic dissimilarity in their

TABLE 2
Thermal Relaxation Time for Biological Tissues at Different UV Laser Wavelengths

	Wavelength (nm)				
Tissue	223	244	266	308	355
	Thermal Relaxation Time (μs)				
Liver	0.545	3.00	5.10	82.0	250
Aorta	0.552	3.10	5.20	84.0	260
Myocardium	0.557	3.10	5.15	83.5	258
Skin	0.416	2.34	3.90	62.4	250

optical properties and are well suited to evaluate the performance of the model. The model of ablation for the cornea is likely to be similar to that of PMMA [19].

PMMA is considered as a weak absorber of UV compared to polyimide and has its first absorption maximum close to the ArF excimer laser at 193 nm. The absorption of the polymer PMMA drops by one order of magnitude at 248 nm (KrF laser) and at 308 nm (XeCl laser) it is almost transparent. The variation of ablation depth with laser intensity for the polymer PMMA at different laser wavelengths, namely 193 nm, 223 nm, 244 nm, and 266 nm, is given in Fig. 1. The slope of the straight line curves (ablation depths vs. ln I) gives inverse of absorption coefficient at a particular laser wavelength. The intersection points are interpreted as threshold intensity for ablation.

The variation of ablation with laser intensity for PMMA at different laser wavelengths, namely 282 nm, 308 nm, 351 nm, and 355 nm, is given in Fig. 2. It is evident from Fig. 1 and Fig. 2 that the minimum ablation depth/pulse occurs at 193 nm which corresponds to maximum absorption at that laser wavelength. In general, the ablation depths increases with increase in laser wavelengths. This is due to the fact that an inverse relationship exists between absorption coefficient and laser wavelengths for PMMA.

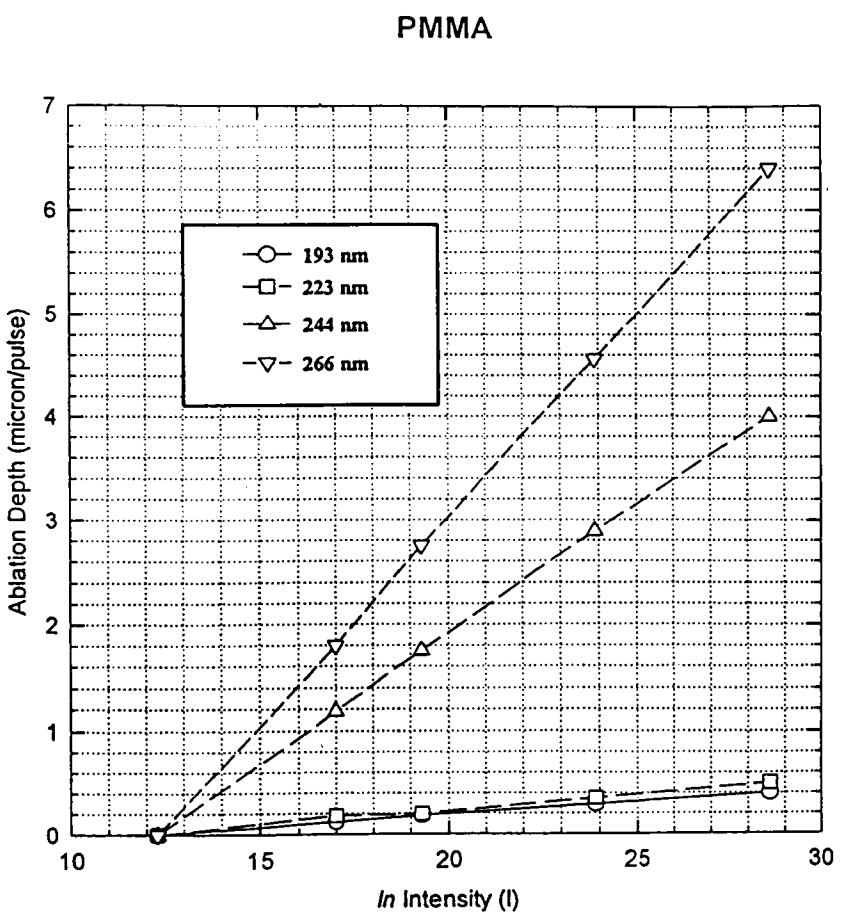

FIG. 1. Variation of ablation depth with laser intensity for PMMA at different laser wavelengths namely, 193 nm 223 nm, 244 nm, and 266 nm.

The variation of ablation depths with laser intensity for Polyimide at different laser wavelengths, namely 193 nm, 223 nm, 244 nm, and 266 nm is shown in Fig. 3 and for the laser wavelengths 351 nm and 355 nm it is shown in Fig. 4. Due to a strongly delocalized distribution of electronic states, polyimides exhibit larger absorption coefficients in UV than that of PMMA [20]. The threshold intensity of ablation is much smaller for polyimide than that of PMMA for

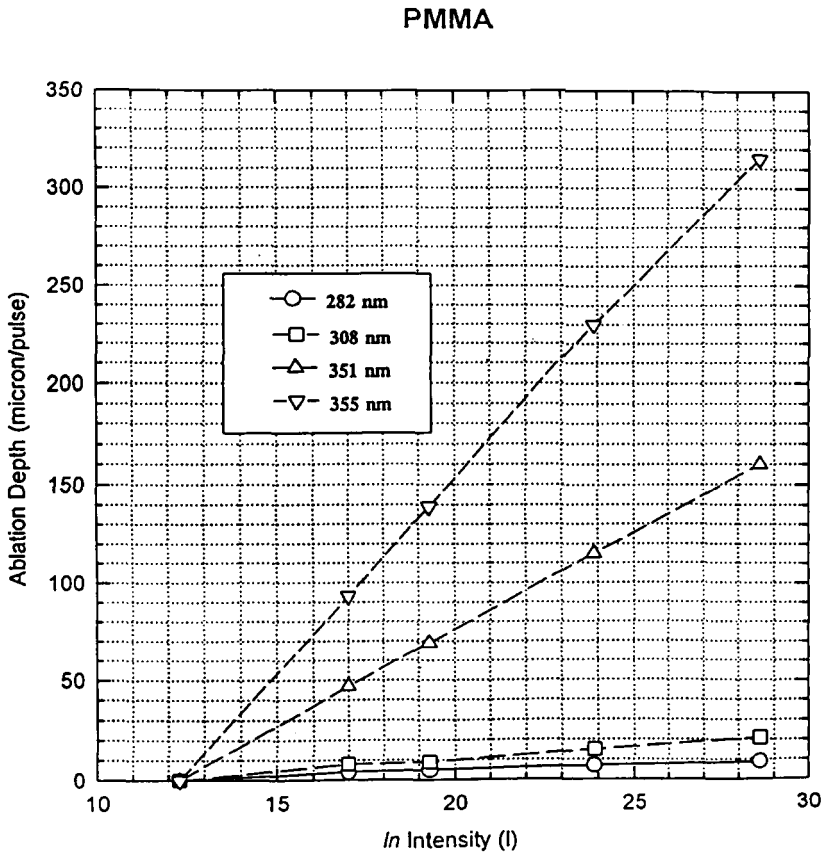

FIG. 2. Variation of ablation depth with laser intensity for PMMA at different laser wavelengths namely, 282 nm 308 nm, 351 nm, and 355 nm.

all the given laser wavelengths used in this investigation. This is due to the stronger absorption of polyimide compared to PMMA. At a fixed laser intensity, the ablation depth is minimum at 193 nm compared to other wavelengths, and is found similar to that of the polymer PMMA. For any given wavelength, at a fixed laser intensity, ablation depth is smaller in polyimide compared to that of PMMA. The minimum ablation depth at 193 nm ensures a well defined cutting with minimal thermal damage to the surrounding structures at this laser wavelength. It has been shown

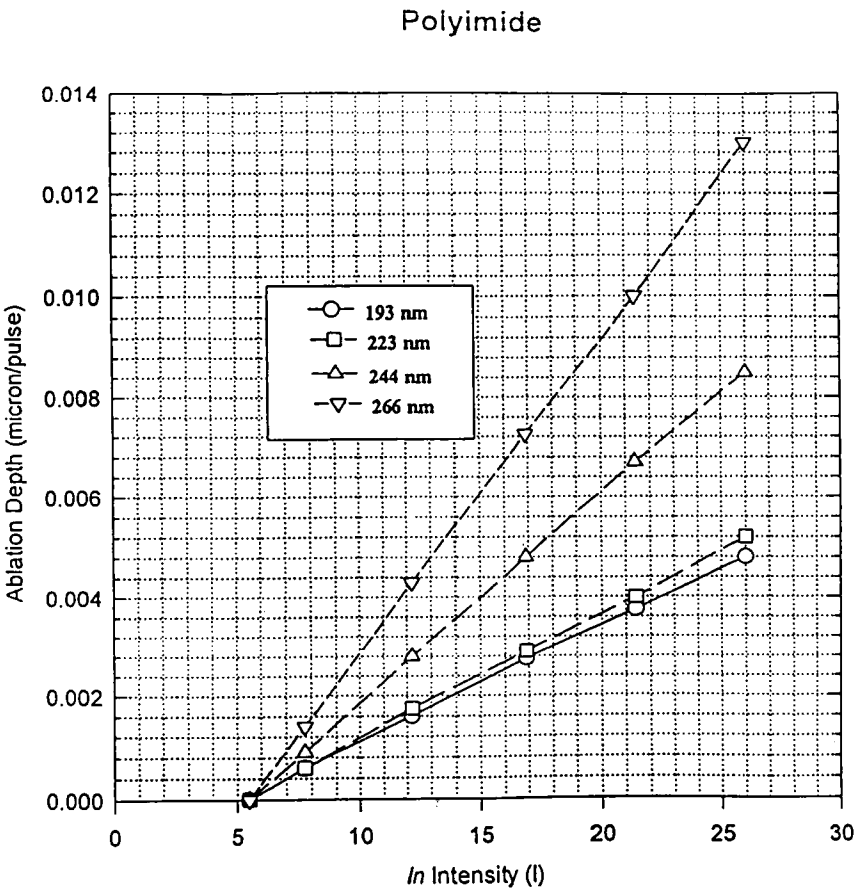

FIG. 3. Variation of ablation depth with laser intensity for Polyimide at different laser wavelengths namely, 193 nm 223 nm, 244 nm, and 266 nm.

that the wall of the aorta from a cadaver exposed to 193 nm laser produces a clean cut whereas with green light at 532 nm the same sample exhibits distorted cut, dehydration, charring, and possibly melting [21]. The UV photon at 193 nm excites the electronic states which leads to bond dissociation, whereas the laser photon at visible wavelengths excites the vibrational modes of the molecule either directly or indirectly via a low lying electronic state. This is basically a thermal

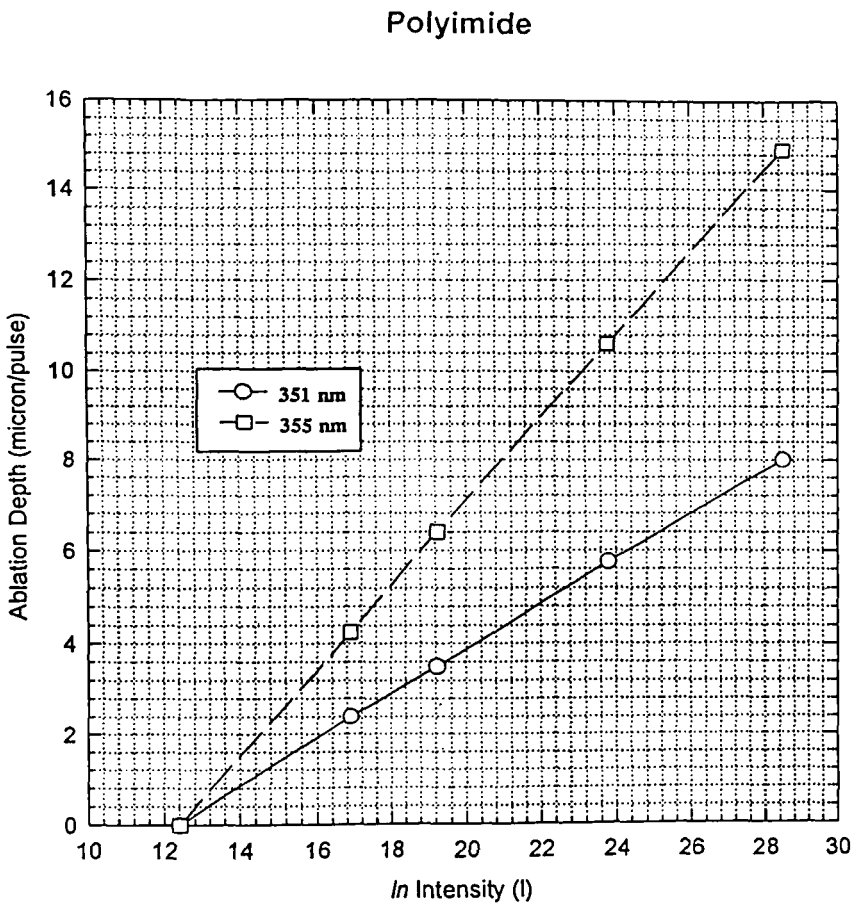

FIG. 4. Variation of ablation depth with laser intensity for Polyimide at two laser wavelengths namely, 351 nm, and 355 nm.

process. But, the exact wavelength regimes which separates the non-thermal and thermal process for individual tissues are not yet known. Below non-thermal ablation threshold, non-radiative relaxation processes from excited electronic states may excite the vibrational modes which will heat the sample. The photo-fragmentation can also be exothermic.

CONCLUSION

We have developed a non-thermal laser ablation model which may reduce thermal damage to neighboring structures. In this model, bimolecular chains are broken up and macromolecule are clipped into fragments during laser radiation. Any excess energy appears as kinetic energy of photo-fragments. The model has been tested for the biopolymers PMMA and polyimide with various lasers and can be extended to various biological tissues. It has been found that the ablation depth/pulse is minimum at 193 nm both for PMMA and polyimide as compared to any other laser wavelength in this study. The minimum ablation depth at 193 nm ensures a well defined cutting with minimal thermal damage to the surrounding structures at this wavelength. There exists a minimum threshold energy fluence for non-thermal ablation and below this threshold, non-radiative relaxation processes from the excited electronic states may lead to thermal ablation. The selection of an ultra short laser pulse with pulse duration shorter than the thermal relaxation time will eliminate thermal diffusion to surrounding structures. By properly selecting the laser wavelength, energy fluence, and pulse duration for an individual tissue, a well controlled non-thermal surgery can by performed. It is expected that the new femto-second lasers will completely eliminate tissue damage due to thermal diffusion.

ACKNOWLEDGMENTS

The authors would like to acknowledge the financial support of the Naval Research Grant #N00014-94-1-0523 and the technical assistance of C. Haridas, Ms. Mona Patel, Moogi-Ayanna Mogengege, and Ms. Donna McGregor in the preparation of the manuscript.

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Date Received:September 24, 1997
Date Accepted:November 6, 1997