

Microsurgery of the Retina With a Needle-Guided 193-nm Excimer Laser

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This article presents a method used to guide the beam from an argon fluoride excimer laser to make it suitable for microsurgical purposes and confine it to areas that can be varied in dimension from 1 μm to tens or hundreds of microns. This approach guides the excimer laser beam with an articulated mechanical arm and confines it with variable-diameter tapered tubes, possibly allowing the use of this laser in *in vitro* retinal surgery with endolaser techniques. Currently, because of the lack of a delivery and focusing system for the 193-nm argon fluoride beam and its absorption by biologic liquids, this laser is used exclusively in ophthalmology for topical applications, such as corneal sculpting. This new method resolves these problems in a unique way with impressive results. Specifically, it was shown that, with this needle-guided excimer laser, it is possible to remove retinal tissue accurately without detectable damage to surrounding cells. Applications of this new technique in retinal surgery are discussed. Invest Ophthalmol Vis Sci 33:2377-2381, 1992

In this article, we present our results of using a new method of tissue microsurgery that differ from the techniques currently available in terms of the mechanism of laser-tissue interaction and the damage zone induced by this interaction in the surrounding tissue.

Damage Zone in Laser Microsurgery

The interaction mechanisms that currently are associated with laser microsurgery are heat generation or dielectric breakdown of the tissue. As a result of heat diffusion into surrounding tissue and the shock waves created by dielectric breakdown, a damage zone is produced around the primary lesion with a width that is approximately the dimension of the laser beam used.¹⁻³ This means that the volume of damaged tissue is considerably larger than might be suspected from the size of the laser beam.

In such damage zones, it is important to differentiate between tissue and cell microsurgery. Thus, in cell microsurgery, where a submicron-sized lesion may be produced, the damage zone also may be of a

submicron size.⁴ However, these dimensions are achieved by using high-magnification microscope objectives, which generally are not applicable in the field of tissue microsurgery (eg, in the retina). Therefore, the minimal beam spots are on the order of several tens of microns, and the damage zones are of equivalent sizes.³ For delicate tissues like the retina, such a relatively large damage zone is the principal limiting factor that governs the type of laser applications that are possible. In summary, for laser tissue microsurgery, it is desirable to use a laser with a variable spot size from a few microns and a z-plane resolution of at least 1 μm , in which there would be no thermal or other spatially hard-to-confine damage to the surrounding tissue in all three dimensions.

Photochemical Laser-Tissue Interaction

To meet these requirements, it would be ideal for the laser to interact with biologic tissue photochemically rather than thermally or by dielectric breakdown. Biologic tissue is composed of molecules that are formed principally of carbon, nitrogen, and oxygen. The bonds that these molecules contain have energies of dissociation in an area that corresponds to the deep ultraviolet region of the electromagnetic spectrum. The argon fluoride excimer laser also emits such photonic energies. The wavelength of this laser is 193 nm, which is the shortest available laser wavelength that can be propagated in air. It has been shown^{5,6} that such a wavelength is absorbed by bio-

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polymeric molecules, which are raised by the radiation to a dissociative excited state. After they have reached this excited state, the molecules enter a photochemical pathway in which there is a direct breaking of molecular bonds. In principle, all the energy of the photon is used in this ablation process of breaking molecular bonds rather than to heat the tissue.

In practice, excimer lasers produce many emissions, some at longer and others at shorter wavelengths. The longer wavelengths (248 and 308 nm) result in ablation with increasing thermal effects and depth of penetration.⁶ In addition, these wavelengths are known to cause damage to genetic material; this does not occur with the 193-nm argon fluoride laser.⁷⁻⁹ Available laser wavelengths shorter than 193 nm can only be propagated in a vacuum. The 193-nm wavelength is the shortest for which optical elements are available for guiding the beam. Even though such optical elements exist, they are only sufficient to guide and focus the beam in a manner that is considerably less precise than that required for the microsurgical applications we envision. Specifically, lenses in this region have considerable aberrations, thus making their focal point too large for microsurgery. In addition to these optical limitations, there is another problem in using the argon fluoride excimer laser in biologic aqueous solutions. This is caused by the strong absorption of the 193-nm wavelength by such solutions.¹⁰

Because of these limitations, the principal application for the argon fluoride excimer laser in medicine is in the field of refractive corneal surgery.¹¹⁻¹³ The application of this laser to refractive surgery uses the lack of heating of the surrounding tissue and the shallow submicron ablation depth of the tissue with each pulse of the laser to remove layers of the cornea. This method does not require small beam diameters and does not allow for the possibility of endolaser applications on such tissues as the retina. In this article, we present a new method for excimer laser confinement and delivery that resolves the problems described by combining a long focal length lens with a rapidly tapered stainless-steel tube. This needle is long enough to permit the positioning of a tip in close proximity to the surface of the tissue that is to be ablated and thin enough to be introduced even into the eye. This combination makes it possible to achieve the energy flows exiting the aperture that are sufficient for ablation of biologic materials. Stability is produced by using an articulated mechanical arm controlled by the surgeon.

Materials and Methods

The special metal needles we used with an entrance diameter of 1 mm and rapidly tapered tip to dimen-

sions of either 120 or 250 μm were made by Nano-Med Instruments (Jerusalem, Israel; Fig. 1). With this approach, it is possible to confine the laser beam in the x - y plane to dimensions that can vary from tens to hundreds of microns. In addition to such x - y confinement, the high absorption coefficient of biologic materials allows for submicron accuracy in the z direction. If necessary, it is even possible to achieve micron-sized dimensions of the beam spot by introducing a properly tapered glass micropipette into the end of the needle. However, such a micropipette technique is not applicable to tissue microsurgery because the surgeon generally works with an up to 40 \times magnification binocular lens, which is only capable of distinguishing movements on the order of tens of microns. Therefore, the surgeon will not be able to hold the pipette tip in close proximity to the tissue surface without breaking it. With an appropriate microscope and micromanipulator, the micropipette technique is convenient for use in cellular surgical applications.¹⁴

In the first series of experiments, fresh bovine eyes were obtained from a local slaughterhouse immediately after the animals were killed. The eyes were prepared as eyecup preparations, which then were cut into pieces. The retina was covered with a thin layer of liquid, which would normally have prevented any ablation of the retinal tissue. To overcome this problem, we applied a low-pressure air stream that emerged from the tip of the tapered tube. This pushed the liquid out of the area that was to be irradiated. This stream of the air also removed from the irradiated region any blood or other effusions that would prevent effective ablation.

The tapered tube used in these experiments was a 35-mm long needle with an outer diameter of 1.1 mm (no. 18) and an aperture diameter of 120 μm . A Model 103 MSG, Lambda Physik (Gottingen, West Germany) argon fluoride excimer laser with a 193-nm wavelength was used. The laser beam was directed with a series of mirrors and focused by a 1000-mm focal-length fused silica lens. Various pulse energies and numbers of pulses were used until the optimal conditions were achieved to produce precise removal of the retinal layers in a manner that was identifiable by the histologic procedures employed.

To demonstrate the practicality of our method in eye surgery, *in vivo* experiments on rabbit eyes were undertaken. To deliver the laser beam reliably in these experiments, an articulated mechanical arm

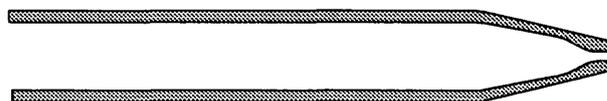


Fig. 1. A schematic representation of the tapered needle.

(NanoMed) was attached to the laser in conjunction with the tapered needle. Albino rabbits weighing 2.0–2.5 kg were anesthetized with intravenous thiopental. Extracapsular cataract extraction was done through a limbal incision. Pars plana vitrectomy also was done with fluid–air exchange. The retina remained covered with a thin liquid layer, which normally prevents ablation. An air stream (as described previously) was used to push the liquid out of the region of ablation (Fig. 2). A 35-mm long tapered needle with an outer diameter of 1.1 mm and tip diameter of 250 μm was inserted through a sclerotomy 3.0 mm from the limbus. A series of alterations were produced in the retina using various numbers of incident pulses with an energy flow of 0.5 J/cm^2 per pulse. All procedures involving animals conformed to the ARVO Resolution on the Use of Animals in Research.

Directly after the irradiation procedure for the pathologic studies, the eyes were fixed in buffered formaldehyde 4% for at least 48 hr. The tissue was processed and embedded in paraffin. Sections 5–6- μm thick were stained with hematoxylin and eosin.

Results

In Figure 3, we show a 10- μm depth hole in the bovine retina that was produced by applying 30 pulses of 193-nm radiation at a repetition rate of 100 Hz with an energy flow of 0.7 J/cm^2 /pulse and a spot size of 120 μm . Only part of the nerve fiber layer was ablated. With a series of 50 pulses and the same conditions, we achieved an ablation depth of 20 μm (Fig. 4). The whole nerve fiber layer up to the ganglion cell layer was ablated. With a reduction in the pressure of the air stream so that a thin layer of liquid remained

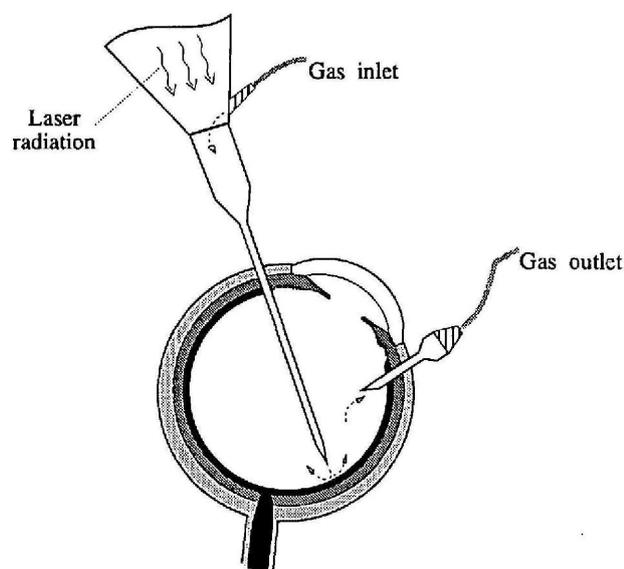


Fig. 2. A scheme of the experiment with an intact rabbit eye.

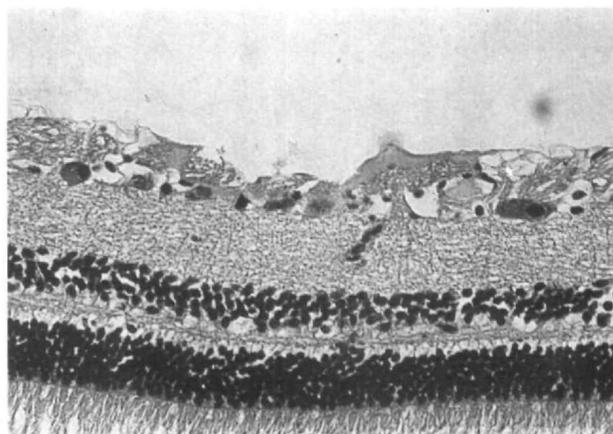


Fig. 3. A 10 μm depth hole in the bovine retina obtained by application of 30 pulses of 193 nm radiation at a repetition rate of 100 Hz and with an energy density of 0.7 J/cm^2 /pulse. The spot size is 120 μm .

over the area to be irradiated, it was possible to achieve an ablation depth (Fig. 5) of approximately one half of that obtained in Figure 4. However, by applying a laser with an energy flow of 1.2 J/cm^2 /pulse and increasing the air stream pressure, with two pulses, we achieved an ablation depth of 40 μm (Fig. 6). The nerve fiber layer, the ganglion cell layer, and the inner plexiform layer were ablated up to the inner nuclear layer. These results indicate the wide range in speed of tissue removal that can be achieved with high accuracy. Similar results were found in the *in vivo* experiments done on an intact rabbit eye, where pulses of 0.5 J/cm^2 /pulse were used with a repetition rate of 30 Hz (Fig. 7).

The form of the bottom of the lesion shown in Figures 4 and 6 indicated that the radiation emitted from the tip had a homogeneous energy distribution. Nuclear material was found to be more resistant to abla-

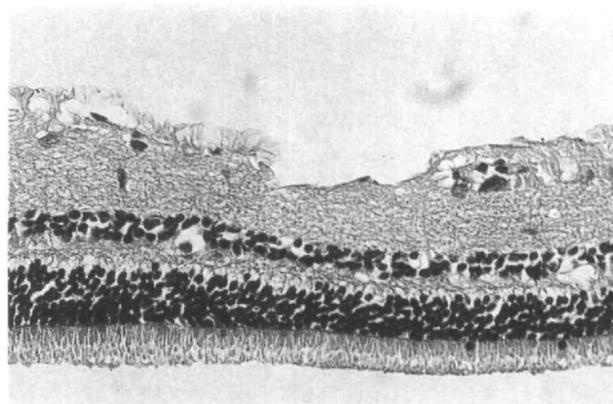


Fig. 4. A 20 μm bovine retinal hole produced under the same conditions as in Figure 3, except that 50 pulses of the laser were used.

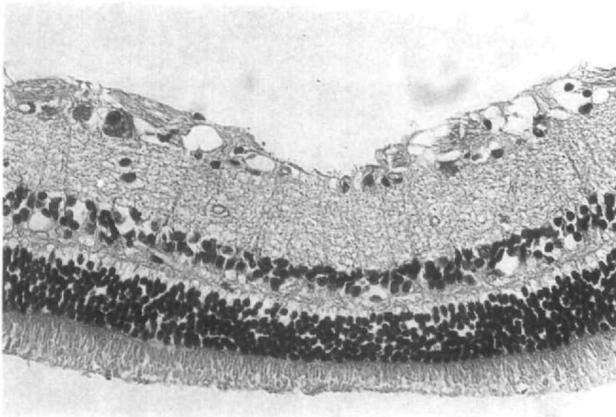


Fig. 5. A 10 μm depth hole produced under the same conditions as in Figure 4, except that a thin layer of water was allowed to remain over the retina to be ablated.

tion than the surrounding nervous tissue. The material in the immediate vicinity of the ablated region showed almost no effect of the procedure and the borders of the lesions were sharp and clean. These findings are interesting, especially when compared with the results of argon ion or low-energy neodymium: yttrium aluminum garnet (Nd:YAG) laser interactions with retinal tissue. The latter cause thermal coagulation. In higher energy ($>380 \text{ mJ/cm}^2$ on the retina) Q-switched Nd:YAG laser surgery, explosive detachment of the sensory retina from the retinal pigment epithelium occurs as a result of dielectric breakdown in the tissue.³

Discussion

It appears that, during wet tissue ablation, the pressure of the air stream that exits from the tip of the needle has great influence on the ablation depth. We found that, in some instances, reduction in the air

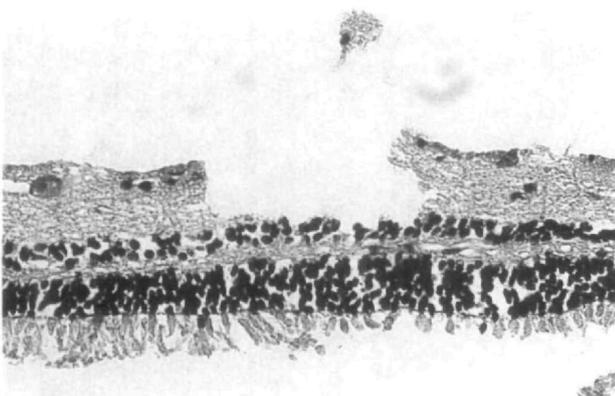


Fig. 6. A 40 μm depth hole in the bovine retina obtained by the application of two pulses of 193 nm radiation with an energy fluence of $1.2 \text{ J/cm}^2/\text{pulse}$.

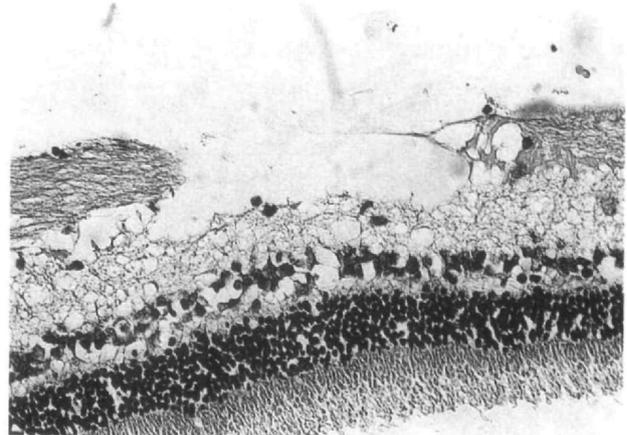


Fig. 7. A 30 μm depth hole obtained in the experiment with an intact rabbit eye by application of the laser pulses with an energy fluence of $0.5 \text{ J/cm}^2/\text{pulse}$ and repetition rate of 30 Hz.

stream decreased the ablation rate to zero. This agreed with previously published results that indicated no 193-nm ablation when bleeding occurred.¹⁰ Therefore, the air stream probably dries the surface of the retinal tissue, and this has a crucial effect on ablation depth. We also studied gels using our tapered-needle method and found that the depth of ablation is extremely sensitive to the air stream and inversely proportional to the gel concentration. These effects may explain the difference in the depth of ablation in our results compared with published data on corneal ablation using the 193-nm excimer laser.^{15,16} These experiments currently are being continued in our laboratory.

Numerous applications of this technique are possible in retinal surgery. For example, epiretinal membranes that cause shrinking of the retina and prevent reattachment during vitreoretinal surgery could be ablated by passing the tip over the affected area without damaging the retina. In such an application, the depth of ablation does not need to be measured because membrane destruction is the ultimate criterion for stopping the ablation. Furthermore, even a thin layer of liquid under such a membrane protects the underlying retina from radiation damage. Thus, this should be an effective procedure for freeing and reattaching the retina.

It also appears likely that a full-thickness retinectomy could be achieved. As seen from our results, such a cut would be sharp and accurate with little damage to the underlying layers. Currently, the method of choice in such instances is to use scissors. From our results, it is clear that needle-guided excimer lasers will be valuable tools in this procedure. In addition, in retinotomy, a sharp accurate hole could be produced to remove fluid behind the retina. Finally, other methods appear possible in ophthalmol-

ogy and other clinical disciplines using this new microsurgical technique.

In summary, in this article, we presented results that raise the exciting possibility of a new level of precision in laser microsurgery. Our approach addressed, not only the problem of producing a laser beam with high x-, y-, and z-plane precision, but also the problem of how to confine damage precisely to the region being illuminated. In essence, this technique produced, for the first time to our knowledge, a highly confined laser spot in all three dimensions in a fashion that is applicable to endomicrosurgical applications without thermal, light, or shock wave damage to the surrounding tissue.

Key words: laser surgery, microsurgery, photoablation, excimer laser, 193-nm radiation, retina surgery

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