Ablation of Vitreous Tissue with Erbium:YAG Laser

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PURPOSE. Using a noncontact erbium (Er):yttrium—aluminum—garnet (YAG) laser, ablation of vitreous was compared to distilled water in vitro.

METHODS. The porcine vitreous body and distilled water were ablated in vitro at different pulse lengths and pulse energies. Selected pulse energies were 25, 35, 45, 75, and 100 mJ (pulse rate: 1 Hz; laser beam diameter at the surface of the sample: 2 mm). Pulse lengths were at 140 ± 3 μsec, 190 ± 4 μsec, and 240 ± 5 μsec. The loss of weight in vitreous tissue and distilled water was measured using precision scales and corrected for evaporation, respectively. The Mann-Whitney U test was used to assess the significance of differences in ablation rates of water and vitreous. P < 0.05 was considered statistically significant.

RESULTS. Reproducible and constant ablation rates were found in both vitreous and distilled water in each of 10 consecutive series of 50 laser pulses at constant laser parameters. Ablation rates per pulse (μg/μsec) of vitreous tissue were as follows: 3.0 μg to 45.8 μg (140 μsec), 10.4 μg to 53.8 μg (190 μsec), and 17.9 μg to 24.2 μg (240 μsec). The ablation rates exhibited a linear correlation with increasing pulse energies and also with decreasing pulse lengths. Considering the pulse lengths of 190 μsec and 240 μsec with all pulse energies tested, the ablation rates of distilled water were significantly higher (P < 0.05) than ablation of vitreous tissue. The ablation rates at a pulse length of 140 μsec were not significantly different. The differences per pulse were as follows: 0.5 μg to 2.1 μg (140 μsec), 1.9 μg to 6.0 μg (190 μsec), and 3.5 μg to 8.7 μg (240 μsec).

CONCLUSIONS. Vitreous ablation is possible using Er:YAG laser. The ablation characteristics of vitreous have proved to be similar but not equal to that of water. (Invest Ophthalmol Vis Sci. 1999; 40:1025-1032)

Infrared lasers, such as neodymium (Nd):ytrrium—aluminum—garnet (YAG), carbon dioxide (CO₂), holmium (Ho):YAG, and, more recently, erbium (Er):YAG, have characteristics that suggest a valuable role in surgery. These lasers transect and ablate tissue based on the absorption of energy by tissue water with consecutive vaporization.1-4 Er:YAG laser radiation of 2.94-μm has an extremely short penetration depth in water (~1 μm) and, hence, allows tissue ablation with micrometer precision. Er:YAG lasers have been successfully used in dentistry,5,6 orthopedics,7 and neurosurgery.8,9 In ophthalmic surgery, Er:YAG lasers have been investigated in corneal10-13, cataract,14,15-17 and glaucoma14 surgery. In retinal and vitreous surgery, Er:YAG laser radiation delivered by different devices to the intraocular space offers the possibility of tractionless cutting of retina and membranes that are difficult to cut mechanically.14,18,19 Wolbarsht20 is among the first to point out the advantages and possibilities of laser-induced removal of the vitreous body. He discussed this proposal based on the CO₂ laser and the hydrogen fluoride laser and stated that the smallest thermal damage is observed by choosing a pulse duration of 1 μsec. Based on this work, further research was performed.21 In the technical field, the common Er:YAG pulse duration is within the range of several hundred microseconds, which has proved to be suitable in ophthalmic surgery for phacoemulsification and vitrectomy.22 Theoretical models of laser-induced vaporization also were developed. The first models were mainly based on thermal confinement (i.e., heat should not diffuse out from the irradiated volume of tissue), whereas more recent models included dynamic effects like the formation of cavitation bubbles.23 With respect to laser vitrectomy, exact data of vitreous ablation by Er:YAG laser energy are essential to evaluate optimal laser parameters, to determine the ablation enthalpy, and to compare vitreous ablation with that of distilled water.

MATERIALS AND METHODS

Er:YAG Laser

The flash lamp-pumped Er:YAG laser used for these experiments was developed at the Department of Physics of the University of Kaiserslautern (Kaiserslautern, Germany).24 The system was used in free-running mode with an emission wavelength of 2.94 μm. The repetition rate of the pulses was 1 Hz and remained unchanged during the measurements. An elec-
FIGURE 1. Time course of laser-light intensity of three single laser pulses measured at different pulse lengths and different pulse energies. The shape of the curves could be reproduced at constant laser parameters. For each laser parameter varied (pulse length, pulse energy), the pulse lengths of 140 ± 3 μsec, 190 ± 4 μsec, and 240 ± 5 μsec were calculated from 50 consecutive laser pulses before ablation. a.u., arbitrary units; τ, laser pulse duration (full width half maximum).

Electronic trigger was used to maintain constant repetition rates. Pulse energies of 25, 35, 45, 75, and 100 mJ were selected. Because of the limited maximal voltage in the electric system, the pulse energy of short laser pulses (140 μsec) was restricted to 45 mJ at maximum. To control fluctuation of the pulse energies, the laser output was monitored by a pyroelectric detector (Coherent Components Auburn, CA) before starting and immediately after finishing each series of measurements at fixed laser parameters.

The length of laser pulses was selected by combining them with the capacities of the discharge unit: 100 μF, 200 μF, and 400 μF to 140 ± 3 μsec, 190 ± 4 μsec, and 240 ± 5 μsec, respectively. The pulse length was measured by an indium-arsenide photodiode (type J12D-M204-R01M; EG&G Company, London, UK). The configuration of the laser pulse was detected by a digital multichannel storage oscilloscope (TDS 540 A; Tektronix, Wilsonville, OR). Figure 1 shows three examples of single laser pulses at different pulse lengths used for the experiments. The shape of the curves could be reproduced at constant laser parameters. For each laser parameter varied (pulse length, pulse energy), the pulse lengths of 140 ± 3 μsec, 190 ± 4 μsec, and 240 ± 5 μsec were calculated before ablation from 50 consecutive laser pulses. Also, the pulse lengths were measured after finishing the ablation measurements to ensure that there was no drift in the pulse lengths.

Experimental Setup
A highly reflective mirror set at a 45° angle to the laser beam was used to direct the laser beam perpendicularly into the center of the surface of the water and vitreous (Fig. 2). The distance between the output mirror and the surface of the fluid in the test tube was 60 cm. The divergence of the laser beam was 0.0018 rad. Before starting and immediately after finishing each series of laser pulses, the diameter of the laser beam was adjusted to 2 mm at the distance of the tissue sample.

To avoid diffraction, as may happen with an aperture, adjustment was done by producing bleaching effects on blackened photopaper with maximum sensitivity in the infrared range of the spectrum (Agfa classic paper; Agfa Gevaert N.V., Mortsel, Belgium) and measuring the diameter of the spots. This method has been tested for accuracy by adjustment of a round aperture of 2-mm diameter concentric to the Er:YAG laser beam. Then, the photopaper was fixed close to the back plane of the aperture. For all laser energies and pulse lengths tested, a round bleaching effect of 2-mm diameter with a distinct margin was detected on the photopaper.

Preparation and Ablation of Vitreous Tissue
Fresh porcine eyes were obtained from an abattoir and immediately stored in physiological electrolyte solution at a temper-
Vitreous Ablation with Erbium:YAG Laser

The initial hypothesis (H₀) to be determined was that there were no significant differences between ablation of water and vitreous. Data were expressed as mean ± SD. The Mann-Whitney U test was used to assess the significance of differences in ablation rates of water and vitreous. P < 0.05 was considered statistically significant.

**RESULTS**

Reproducible and constant ablation rates were found in both vitreous and distilled water in each of 10 consecutive series of 50 laser pulses at constant laser parameters (Fig. 3). In most measurements, SD was within 15% of the total amount of ablated weight (Tables 1, 2). Ablation rates per pulse (μg/μsec) ranged from 3.0 μg to 53.8 μg in vitreous and from 11.7 μg to 55.7 μg in distilled water (Tables 1, 2). Considering the pulse lengths of 190 μsec and 240 μsec of all pulse energies tested, the ablation rates of distilled water were significantly higher (P < 0.05) than the ablation rates of vitreous tissue (Fig. 4, Tables 1, 2). The ablation rates at a pulse length of 140 μsec were not significantly different. Considering a single laser pulse, the differences of weight loss between ablation of water and vitreous were as follows: 0.5 μg to 2.1 μg (140 μsec), 1.9 μg to 6.0 μg (190 μsec), and 3.5 μg to 8.7 μg (240 μsec). The amount of tissue ablation both in vitreous and distilled water at a single laser pulse exhibited a linear increase with increasing pulse energies (Fig. 4) and also with decreasing pulse lengths (Fig. 5).

**DISCUSSION**

Thermal models are commonly used to describe tissue ablation by application of infrared laser light. Investigations on the Er:YAG laser used for vitreoretinal surgery aimed at development of a model for laser-generated bubble expansion and a model for predicting intravitreal transient thermal phenomena based on intravitreal temperature measurements. Heat accumulation in vitreous tissue and water with application of consecutive laser pulses was theoretically avoided in the present investigation by using a low repetition rate of pulses (1 Hz) and...
pulse lengths far exceeding the thermal relaxation time of water $\tau_{\text{therm}}$ of approximately 1 $\mu$s at the absorption peak near 3 $\mu$m. Hence, heat diffusion away from the area of tissue exposed to laser radiation may explain constant ablation rates detected when laser parameters are kept unchanged (Fig. 3). Using a CO$_2$ laser and an Er:YAG laser, Zweig et al. measured a linear increase of ablation crater depth in gelatin (density, 0.66 g/cm$^3$ and 0.84 g/cm$^3$, respectively) with number of pulses. Gijsbers et al. ablating colored polyacrylamide gel sticks with an Ar$^+$ continuous laser and Hibst investigating Er:YAG laser ablation of hydrated agarose, porcine skin, and porcine corneal tissue reported similar results. In the present study, the range of ablation measured in vitreous tissue was 3.0 $\mu$g to 53.8 $\mu$g per pulse for energies ranging from 25 mJ to 100 mJ (Table 1). The data show that significant amounts of vitreous tissue can be ablated by Er:YAG laser. Ablation rates increased linearly with increasing pulse energies (Fig. 4) and also with decreasing pulse lengths (Fig. 5). A variety of methodological differences exist between studies measuring tissue ablation (e.g., the profile of laser light emission and the type of material ablated). However, after adjustment for different diameters of the laser beam, the amount of ablated material calculated by simple approximation from the data of these studies is, with the exception of the study of Ren et al., within the magnitude of our results. In all studies a linear correlation with increasing energy of laser pulses was reported for corresponding depths of the ablation crater. For ablation of skin tissue, Hibst hypothesized that the linear correlation found may be based on the effect that laser radiation is not significantly attenuated by ablation material. Hibst ablated porcine corneal tissue with a normal spiking Er:YAG laser (fluence: 1.27–3.25 J/cm$^2$; pulse length: 130 $\mu$s). The crater

**TABLE 1. Net Ablation Rates of Vitreous Measured at Different Pulse Lengths and Pulse Energies**

<table>
<thead>
<tr>
<th>Pulse Energy (mJ)</th>
<th>50 Pulses</th>
<th>1 Pulse</th>
<th>50 Pulses</th>
<th>1 Pulse</th>
<th>50 Pulses</th>
<th>1 Pulse</th>
</tr>
</thead>
<tbody>
<tr>
<td>25</td>
<td>896 ± 101</td>
<td>17.9 ± 2.0</td>
<td>522 ± 60</td>
<td>10.4 ± 1.2</td>
<td>150 ± 40</td>
<td>3.0 ± 0.8</td>
</tr>
<tr>
<td>35</td>
<td>1090 ± 75</td>
<td>21.8 ± 1.5</td>
<td>768 ± 23</td>
<td>15.4 ± 4.7</td>
<td>528 ± 30</td>
<td>10.6 ± 0.6</td>
</tr>
<tr>
<td>45</td>
<td>1211 ± 85</td>
<td>24.2 ± 1.7</td>
<td>1035 ± 131</td>
<td>20.7 ± 2.6</td>
<td>821 ± 359</td>
<td>16.4 ± 7.2</td>
</tr>
<tr>
<td>75</td>
<td>1789 ± 115</td>
<td>35.8 ± 2.3</td>
<td>1557 ± 125</td>
<td>31.1 ± 2.5</td>
<td></td>
<td></td>
</tr>
<tr>
<td>100</td>
<td>2690 ± 135</td>
<td>53.8 ± 2.7</td>
<td>2289 ± 187</td>
<td>45.8 ± 3.7</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

The amount of weight ablated is given as average ± SD, expressed as micrograms, for 50 and 1 laser pulses, calculated from 10 series of 50 laser pulses applied at constant laser parameters.
### Table 2. Net Ablation Rates of Distilled Water Measured at Different Pulse Lengths and Pulse Energies

<table>
<thead>
<tr>
<th>Pulse Energy (mJ)</th>
<th>Pulse Length (ms)</th>
<th>50 Pulses</th>
<th>1 Pulse</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>140</td>
<td>25</td>
<td>35</td>
</tr>
<tr>
<td></td>
<td></td>
<td>998 ± 81</td>
<td>20.0 ± 1.6</td>
</tr>
<tr>
<td></td>
<td>190</td>
<td>722 ± 49</td>
<td>14.4 ± 1.0</td>
</tr>
<tr>
<td></td>
<td></td>
<td>586 ± 35</td>
<td>11.7 ± 0.7</td>
</tr>
<tr>
<td></td>
<td>240</td>
<td>2021 ± 121</td>
<td>40.4 ± 2.4</td>
</tr>
<tr>
<td></td>
<td></td>
<td>2463 ± 31</td>
<td>49.3 ± 0.6</td>
</tr>
</tbody>
</table>

The amount of weight ablated is given as average ± SD, expressed as micrograms, for 50 and 1 laser pulses, calculated from 10 series of 50 laser pulses applied at constant laser parameters.

The depth was between 7 µm and 30 µm. Zweig et al.²⁹ used an Er:YAG laser for drilling holes in gelatin targets (density: 0.84 g/cm³; calculated fluence: 128-442 J/cm²). Galitsis et al.¹⁵ drilled holes into cataractous human crystalline lenses (fluence, 4.5–19.5 J/cm²; pulse length, 250 µsec). Ren et al.¹³ ablated corneal tissue with a normal spiking Er:YAG laser in 10 calf eyes. Approximation of ablated weight of tissue was not possible based on the data given in the text. A survey of further studies evaluating laser ablation of different tissues is given by Hibst. The amount of weight ablated is given as average ± SD, expressed as micrograms, for 50 and 1 laser pulses, calculated from 10 series of 50 laser pulses applied at constant laser parameters.

Figure 6 shows the laser energy calculated for ablation of 1 g of vitreous tissue at different pulse energies and pulse lengths in comparison to evaporation enthalpy of water (energy needed for heating water up to 100°C and consecutive evaporation). Increasing the energy per laser pulse and shortening the pulse length reduces the overall energy required for ablation of 1 g of vitreous tissue. Hibst found the energy needed for Er:YAG laser ablation of skin being lower than the evaporation enthalpy of water and hypothesized that besides vapor tissue particles containing fluid may be expelled from the target. On the basis of this hypothesis, one can assume...
from our investigations that rapid localized heat accumulation during ablation with short pulses (e.g., 140 μsec) and high energies per pulse may induce disruptive effects (Fig. 7). In contrast, more energy is required for ablation of a certain amount of vitreous by using pulses of longer duration (e.g., 240 μsec) or lower energies per pulse. Heat accumulation may be
less rapid and the thermal transition zone larger than pulses of shorter duration or higher energy. Tissue is mainly ablated by vaporization. Disruption of tissue particles may occur less frequently.

Comparative measurement of ablation of distilled water has been performed to prove consistency of data. The water content of human vitreous tissue lies between 98.5% and 99.7%. Hence, results of ablation of distilled water were expected to be similar to those of vitreous tissue ablation. Actually, ablation rates of distilled water exhibited a linear increase with increasing pulse energies and also with decreasing pulse lengths. Furthermore, as found in vitreous tissue, the ablation rates of water were constant in consecutive series of 10 measurements (50 laser pulses each series). High power densities and disruptive effects similar to those of vitreous tissue may have increased the range of ablation rates measured at 140 μsec (pulse energy, 35 mJ and 45 mJ) and, hence, may have affected the calculated SD and statistical analysis performed in comparison with vitreous tissue. Considering the pulse lengths of 190 μsec and 240 μsec with all pulse energies tested, the ablation rates of distilled water were significantly higher than ablation rates of vitreous tissue (Fig. 4, Tables 1, 2). Evaporation enthalpy of soluble vitreous proteins (5 kJ/g) exceeds the amount necessary for evaporation of water (2.6 kJ/g). Besides, the ultrastructural organization of the vitreous may affect parameters of thermal tissue interaction (e.g., absorption coefficient, heat conductivity, and heat capacity). The water of the vitreous is stabilized by an ordered meshwork of very fine collagen fibrils that are tied together in loosely parallel bundles or sprays by bridges of sulfated glycosaminoglycan. The glycosaminoglycans of the vitreous (hyaluronan and chondroitin sulfate) aggregate with themselves and with each other in solution. The protein cores of the proteoglycans are attached to collagen fibrils.

Pulse energies lower than 25 mJ were not used because prior pilot measurements did not reveal a loss of weight for a pulse energy of 15 mJ (pulse length, 240 μsec). A significant loss of weight by ablation may exist that is, however, within the SD of measurements. Another possible explanation is a steep decrease of ablation with pulses below a certain threshold energy. Investigation of laser pulse energies lower than 25 mJ may require a different methodological approach and, thus, was not part of this study. Pulse energies were also limited to 100 mJ because considerable visual and acoustic signs of shock waves were observed for higher energy levels (e.g., 125 mJ), and, thus, such energies are inappropriate for use in intraocular surgery because they may lead to severe injury of ocular tissue. However, future quantitative measurements of shock waves and in vivo investigation, including histology, may provide detailed data on this topic.

To the best of our knowledge this is the first time laser ablation of vitreous tissue has been quantified. Noncontact ablation was chosen instead of placing a laser fiber tip beneath the surface of material to avoid recondensation interfering with measurement of ablation. Ablation characteristics of the vitreous have proven to be similar but not equal to those of water. The distinct physical and chemical properties may explain the significantly lower ablation rate of vitreous tissue than distilled water. Further investigation is required to measure ablation using fibers in contact with tissue.

Acknowledgment

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References


