Laser Energy Reaching the Posterior Pole During Transscleral Cyclophotocoagulation

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Objective: To measure scattered laser energy reaching the posterior pole during transscleral cyclophotocoagulation.

Methods: Transscleral cyclophotocoagulation was performed on 4 cadaver eyes with Nd:YAG noncontact, Nd:YAG contact, and diode contact lasers. Energy was measured with a photodiode through a 7-mm trephined hole in the posterior pole. Average percentage power, average power, and average energy transmission were calculated. American Conference of Governmental Industrial Hygienists (ACGIH) guidelines were used to calculate allowable energy exposures for each laser.

Results: All 3 lasers transmitted 3% to 5% of the power to the posterior pole. The average energy transmission was 240 to 260 mJ for all lasers. The contact lasers had an average power transmission of 120 mW. The noncontact Nd:YAG laser, with shorter pulse duration, had an average power transmission of 13 000 mW, significantly greater than that of the other lasers. The ACGIH guidelines for allowable energy exposures were 93 mJ for the noncontact Nd:YAG laser, 1300 mJ for the contact Nd:YAG laser, and 440 mJ for the contact diode laser.

Conclusions: Three percent to 5% of laser power delivered during cyclophotocoagulation reaches the posterior pole. Exposure energies may approach or exceed ACGIH guidelines. The clinical significance of these findings remains to be shown.

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T}RANSSCERAL laser cycloablative procedures have been shown to be an effective means of lowering the intraocular pressure for the high-risk glaucoma population in which more conventional operations have relatively poor success rates.1 Laser units are now commercially available for performing this surgery with either Nd:YAG (1064 nm) or semiconductor diode (810 nm) wavelengths. The utility of these procedures, however, is limited by the high incidence of postoperative visual loss; as a result, they are generally used only in patients with low visual potential.

The reported incidence of visual reduction with transscleral Nd:YAG cyclophotocoagulation ranges between one third and two thirds of treated patients. Trope and Steven2 treated 20 sighted eyes with a noncontact, slitlamp delivery Nd:YAG laser (Microruptor II, Lasag, Thun, Switzerland) and noted that 13 (65%) lost 1 line of vision, including 2 who lost light perception. Shields and Shields,3 using the same type of laser in 500 patients, found visual loss of 2 or more lines in 6 (22%).

The above reports include postoperative visual loss of any mechanism, since it is not possible to clearly distinguish between causes that are a direct result of the laser surgery and those related to underlying ocular disorders, including corneal and retinal diseases, which would contribute to the reduced vision in some cases.

The precise mechanisms by which the laser surgery may lead to reduced vision is unknown. To our knowledge, no studies have directly addressed this question, although reasonable theories of mechanism include maculopathy from inflammation, hypotony, and phthisis. We wondered whether another mechanism might be a toxic effect of light from scattered laser light reaching the macula. We report herein the results of a study designed to measure the laser energy reaching the posterior pole in human autopsy eyes treated with Nd:YAG or diode transscleral cyclophotocoagulation.

RESULTS

Average percentage power, average power, and average energy transmitted to the
**MATERIALS AND METHODS**

Two pairs of human autopsy eyes were obtained within 48 hours of death; donor age at death was greater than 70 years. Each eye was injected with balanced saline solution through clear cornea paracentesis until firm. A 7-mm Barron-Hessburg corneal trephine was placed so as to encompass the optic nerve and macular region, and that portion of sclera, optic nerve, choroid, and retina was removed. The vitreous was not removed. A 2-cm-square glass coverslip was glued over the opening with cyanoacrylate glue (Figure). The eye was again filled with balanced saline solution to achieve an intraocular pressure of approximately 25 mm Hg as judged by digital palpation. The eye was then secured on a specially designed Plexiglas mount that stabilized the eye during laser applications. Three commercially available lasers were used in the study. With each instrument, laser orientation and a range of settings were chosen to simulate the more commonly used clinical protocols. The Nd:YAG laser, Microruptor II (MR2), was used in the free-running, thermal mode, with a duration of 20 milliseconds, and a focal offset of 9 (3.6 mm in air) for noncontact cyclophotocoagulation at energies of 5 to 8 J, applied 1 to 2 mm posterior to the limbus. The laser was applied to the limbus with the eye pointed directly forward toward the laser, as is done clinically, so that the laser energy was directed parallel to the visual axis. With the use of a 600-μm quartz fiberoptic probe for contact Nd:YAG laser cyclophotocoagulation, a Microruptor III (MR3) was used at 1.6 to 3.2 W, with a 2-second duration, applied 1.4 mm posterior to the limbus, with the probe tip fully flush against the sclera in the identical orientation as is used clinically. The semiconductor diode laser, Oculight SLx (IRIS Medical Instruments Inc, Mountain View, Calif), was used with a 600-μm quartz fiberoptic probe (G-probe; IRIS Medical Instruments Inc) for contact cyclophotocoagulation at power levels of 1750 to 3000 mW for 2 seconds, 1.2 mm posterior to the limbus, with the probe edge aligned at the limbus and tip flush to the sclera in the same orientation as is used clinically. The angle of laser energy to the sclera is thus determined by the G-probe design. Each eye received 2 laser applications to each 180° of perilimbal area, except for 180° in the first eye, which was used for calibration of the power meter. Each laser instrument was used as the first treatment to 180° of one eye, and as a second treatment to another eye. Between 9 and 15 measurements were made with each laser at varied settings in each 180°.

A photodiode alternating-current power meter was designed and constructed to be sensitive to light from visible to far infrared wavelengths. The power meter was aligned directly behind the coverslip. The power meter was attached to a storage oscilloscope that displayed its response to laser irradiation. The power meter's initial output was directly proportional to the initial laser power (energy per time) reaching the detector. Before each laser application to the eye, the laser was fired directly into the power meter detector and the power meter response was recorded. All measurements with the MR2 were made with a 1.0 optical density filter in front of the detector to prevent damage to the detector. The laser was then applied to the cadaver eye by means of the same laser settings and the power meter response was again recorded. The 2 measurements were compared, and the percentage of the total laser power that was transmitted to the posterior pole was calculated.

The reason for the calculation of the percentage power transmission is as follows. The power meter displayed relative, not absolute, power measurements. This necessitated calibration of the measured power against a known power meter response was recorded. All measurements with the MR2 were made with a 1.0 optical density filter in front of the detector to prevent damage to the detector. The laser was then applied to the cadaver eye by means of the same laser settings and the power meter response was again recorded. The 2 measurements were compared, and the percentage of the total laser power that was transmitted to the posterior pole was calculated.

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![Cadaver eye preparation in which a 7-mm trephination was made encompassing the posterior pole including the optic nerve. A glass coverslip has been glued over this opening.](image)

**MR2 LASER**

Seventeen measurements were made at energy settings between 5 and 8 J. Average percentage power transmission was 3.8%, average power transmission was 13 000 mW, and average energy was 260 mJ. A statistically significant difference was seen between the average percentage power transmission at 5 to 7 J vs 7 to 8 J (4.3% and 3.3%, respectively; P<.001). No statistically significant difference was detected between measurements made with the 2 cadaver eyes.

**MR3 LASER**

Seventeen measurements were made at power settings between 1.6 and 3.2 W. Average percentage power transmission was 3.0%, average power transmission was 120 mW, and average energy was 240 mJ. A statistically significant difference was seen between the average percentage power transmission at low (1.6 and 1.8 W) vs high (3.2 W) power (5.6% and 4.5%, respectively; P<.001). No statistically significant difference was detected between measurements made in the 2 cadaver eyes.

**SLX LASER**

Eighteen measurements were made at power settings of either 1.75 or 3.00 W. Average percentage power transmission was 4.7%, average power transmission was 120 mW, and average energy was 220 mJ. No statistically signifi-
The ACGIH 1996 TLVs were not used, as suppression of movement is possible exposure values. The corrections suggested for the application of these values in ophthalmic procedures in the known laser energy settings, pulse duration, and the percentage transmission as measured by the power meter. This technique also eliminates the effects of any difference in power detector sensitivity at the different wavelengths of the different lasers. For the MR2, the protective filter was used both in calibration measurements and in measurements with the cadaver eyes and should therefore not have affected the calculated percentage of power reaching the posterior pole. Average percentage power, average power, and average energy transmitted per pulse are reported. Calculation of the threshold limit values (TLVs) was performed using the 1996 American Conference of Industrial Hygienists (ACGIH) Guidelines. The ACGIH issues guidelines for laser exposures each year.

The current applicable TLVs for diode (810 nm) equals 1.8 \(C_A \frac{t^{3/4}}{mJ/cm^2}\) and for Nd:YAG (1064 nm), 9 \(C_C \frac{mJ}{cm^2}\), where \(t\) indicates exposure duration in seconds; \(C_A\) and \(C_C\) correction factors for given wavelength ranges and exposure times. \(C_A = 1.0\) for \(\lambda = 810\) nm, \(C_A = 1.66\). \(C_C = 1.0\).

The TLV is given in energy per area, and may be multiplied by the pupil area (maximum pupil assumed to be 7 mm) to arrive at the maximum permissible energy exposure to the retina. For a 7-mm pupil, the area = \(\pi r^2 = \pi (3.5)^2 = 0.38\) cm².

The TLV is multiplied by another correction factor, \(C_E\), for exposures to large areas of the retina. The 7-mm area at the posterior pole over which measurements were made in this experiment qualifies as a large area. \(C_E = \alpha / (\alpha_{max} \times \alpha_{min})\) for \(\alpha > \alpha_{min}\), 100 mrad \(\alpha = \) the solid angle of retina exposed in radians. In this experiment, that corresponds to the 7-mm trephined area of measurement, where using the simplified model eye, with a distance of 17 mm from nodal point to retina, \(\alpha = 7\) mm/17 mm = 0.41. For exposures with time > 0.7 sec, \(\alpha_{min} = 2^\sqrt{3} mrad = 3.36\) mrad (for 2-second exposures). For exposures with time < 0.7 sec, \(\alpha_{min} = 1.5\) mrad (for 0.02-second exposures). This yields \(C_E = 504\) for 2-second exposures and 1130 for 0.02-second exposures.

An additional correction exists for repetitively pulsed exposures: \(TLV = (n - 1/4) \alpha / (TLV for single pulse)\), where \(n = \) number of pulses. Where \(n = 25\) (typical cyclophotocoagulation treatment), \(n - 1/4 = 0.447\).

The guidelines for energy exposure for these scenarios may thus be summarized: \(n - 1/4 \alpha / (TLV)\). Therefore, MR2, MR3, and SLx equals 93, 1300, and 440 mJ, respectively.

These calculations were based on 2-second exposures for the MR3 and diode lasers and 0.02-second exposures for the MR2; 25 applications of the laser during the cyclophotocoagulation; and a 7-mm pupil.

Data are reported to 2 significant figures. Measurements were compared between different energy settings for each laser, and between average values for all 3 lasers by means of the 2-tailed Student \(t\) test, assuming unequal variances.

<table>
<thead>
<tr>
<th>Laser*</th>
<th>Average % Power</th>
<th>Average Power, mW</th>
<th>Average Energy, mJ</th>
<th>Threshold Limit Value, mJ†</th>
</tr>
</thead>
<tbody>
<tr>
<td>MR2</td>
<td>3.8‡</td>
<td>13 000§</td>
<td>260</td>
<td>93</td>
</tr>
<tr>
<td>MR3</td>
<td>5.0</td>
<td>120</td>
<td>240</td>
<td>1300</td>
</tr>
<tr>
<td>SLx</td>
<td>4.7</td>
<td>120</td>
<td>240</td>
<td>440</td>
</tr>
</tbody>
</table>

*MR2 is the Microruptor II (Lasag, Thun, Switzerland); MR3, Microruptor III (Lasag); and SLx, Owlfight SLx, IRIS Medical Instruments Inc, Mountain View, Calif.

†Threshold limit value as calculated by American Conference of Governmental Industrial Hygienists guidelines assuming 2-second exposures for the MR3 and SLx, and a 20-millisecond exposure for the MR2.

‡MR2 had statistically significantly lower average percentage power transmitted than either of the other lasers.

§MR2 had statistically significantly higher average power transmitted than either of the other lasers.

This experiment was designed to assess the amount of laser energy reaching the posterior pole during Nd:YAG and diode laser cyclophotocoagulation. One hypothesis tested was that substantial light exposure to the posterior pole occurs during cyclophotocoagulation. A second hypothesis was that the shorter diode wavelength, with poorer scleral penetration and greater absorption by melanin than the longer Nd:YAG wavelength, might have a lower energy transmission to the posterior pole. Underlying both of
these hypotheses is the question as to whether light damage to the posterior pole may play some role in visual loss after cyclophotocoagulation.

The data support the first hypothesis, in that all 3 lasers transmitted large portions of laser energy to the posterior pole: between 3% and 5%. This corresponds to approximately 250 mJ of energy per pulse, depending on laser settings. The ACGIH and American National Standards Institute issue guidelines on laser exposures to the eye. These standards are based on a large body of data from animal experiments and accidental human exposures. The guidelines may be used to calculate the TLVs for specific situations (Table). The guidelines take into account laser wavelength, energy, duration and area of exposure, and number of exposures. The guidelines are designed to fall substantially below the levels at which tissue changes and damage will occur. The lower TLV for the MR2 is largely a function of its shorter pulse duration. These values indicate that all 3 laser systems, when used for cyclophotocoagulation, are close to the TLVs, and that the MR2 exceeds its TLV. As shown in the calculations, varying the number of applications within the clinically useful range has a relatively minor effect on the TLV. The analysis assumes a homogeneous distribution of scattered laser energy over the measurement area. Any lack of homogeneity, eg, a greater concentration of the energy in the center of the beam’s direction, would increase the potential toxic effect in this analysis. If the intensity of scattered laser energy varies markedly over the posterior pole, some areas of the retina may in fact be exposed to energies further in excess of the guidelines.

Our data do not support the second hypothesis regarding laser wavelength and energy transmission, in that there was no statistically significant difference in average laser energy reaching the posterior pole with the 3 instruments. The diode laser, along with the contact Nd:YAG technique, actually had a greater percentage power transmission than the noncontact Nd:YAG procedure. The lower percentage of power transmission with the MR2 may be caused in part by the angle of laser delivery; contact lasers with probes may aim more of the energy toward the posterior pole. Another possible explanation for the greater percentage transmission with the 2 contact cyclophotocoagulation techniques may be the thinning and increased transparency of the sclera that can result from pressure applied to the fiberoptic probe. Pressure on the sclera compacts the tissue, increasing the transmission of laser energy to the ciliary body and the rest of the inner eye, while Blanching the conjunctiva, thus reducing conjunctival burns.

The MR2 also had a significantly higher average power delivery to the posterior pole than the other 2 lasers. This is the result of the MR2’s very short pulse; the same energy is delivered as by the other lasers, but in 1/100th the time. In addition to direct effects of laser power, indirect effects from shock waves generated by the laser’s effects on tissue may also affect other ocular tissues.

This study has several limitations. First, a limited number of eyes were studied. It is possible that variations in pigmentation or anatomical dimensions might influence the laser’s penetration through the tissues. The initial measured laser energy transmission by the photodiode was assumed to remain constant throughout the laser exposure. It is possible that the energy transmitted may be increased or decreased by laser-induced tissue changes. However, no differences were seen between the measurements made during the first treatment of a limbal area and those of the second. Also, the energies used in this experiment were not in a range that would saturate melanin, the main source of absorption. Only a limited range of laser settings were tested that were believed to be clinically relevant. Within this range, though, some variations in energy transmission were seen. The percentage transmissions may be used to estimate transmissions for similar laser settings, and these may be compared with TLVs calculated from ACGIH guidelines for those specific exposures.

The clinical significance of the findings in this study is unknown. Even if the laser exposures to the macula are close to or in excess of the guidelines, it does not prove that this is the source of visual loss in any single patient or group of patients. The typical course of retinal phototoxic effects or thermal effects is early functional loss sometimes followed by late improvement. This is not the pattern of visual loss after cyclophotocoagulation. Additionally, even if the visual loss is in part attributable to the laser energy, its severity and incidence must be weighed against other considerations in any particular patient. Cyclophotocoagulation may still represent the best treatment alternative in some patients who have limited visual potential, who are unable to undergo surgery, or in whom tube shunts or trabeculectomies may not be feasible.

In summary, this study demonstrated that 3% to 5% of laser power delivered during cyclophotocoagulation reaches the posterior pole. Similar average energies are transmitted with the Nd:YAG in contact and noncontact modes, as well as with the diode laser in the contact mode. The power transmission to the posterior pole was higher in the noncontact Nd:YAG laser because of its shorter pulse length. The energy reaching the posterior pole may approach or exceed ACGIH guidelines. The clinical significance of these findings remains to be shown.

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