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Q-Switched CTE:YAG (2.69 μm) Laser Ablation: Basic Investigations on Soft (Corneal) and Hard (Dental) Tissues

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Ablative infrared lasers either show poor transmission in optical fibers (Er:YAG: 2.94 μm; ErCr:YSGG: 2.79 μm) or are characterized by potential relevant thermal side effects (Ho:YAG: 2.1 μm). The CTE:YAG laser (Cr,Tm,Er doted YAG) emits radiation at a wavelength of 2.69 μm. Efficiently high optical fiber transmission is accomplished (attenuation: < 8db/m for Low-Hydroxy-Fused-Silica (LHFS): 0.3 ppm). Since the laser can easily be run in the Q-switch mode (pulse duration: 0.5–2.5 μs) thermal side effects of tissue interaction were expected to be low. Laser tissue interaction was studied on soft (porcine and human cornea), as well as on hard (human dental) tissue. Histological and micromorphological examinations were performed by light microscopy and scanning electron microscopy. It was found that ablation rates in corneal tissue increased from 5 to 90 μm/pulse with increasing laser fluences (5.5–20 J/cm^2). Collateral thermal damage reached as far as 20 ± 5 μm, and was higher (up to 50 μm) when craters where processed in the contact mode using LHFS-optical fibers. In comparison to soft tissue ablation, hard dental tissue ablation showed very little increase of ablation rate (1–3 μm/pulse) when higher fluences were applied. In dental tissue processing, the ablative effect was accompanied by a luminescence, indicating the presence of plasma. We conclude that the presented CTE:YAG laser can be considered as an effective tool for a variety of laser surgical applications where high power optical fiber delivery is required and where strong thermal side effects are not desired.

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Key words: **midinfrared-laser, photoablation, cornea, dental tissue**

INTRODUCTION

The surgical goal in laser ablation is the processing of tissue with a minimum of collateral thermal or mechanical damage. The quality of the laser ablative process can be described by the ratio of the total volume of tissue removal and the expansion of collateral tissue damage.

Midinfrared-Lasers are under investigation as an alternative to Ultraviolet-Excimer-Lasers for photoablative surgery of the 1980s [2,3,11,14,17,20,21]. Due to an absorption maximum at a wavelength of ~ 3 μm, tissue water functions as a chromophore when irradiated by Er:YAG (2.94 μm) or ErCr:YSGG (2.79 μm) laser radiation [5,23]. These two lasers, characterized by high ablative efficiency and rather low thermal collateral damage, have been proposed for extracorporal applications such as ocular or dental surgery [1,6,12,22].

As with short wavelength (193 nm) UV-laser radiation, however, transmission of Er:YAG and

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ErCr:YSGG laser radiation in conventional quartz optical fibers is very poor. Optical fiber delivered laser surgical applications are still limited by technical problems of adequate beam delivery.

The Ho:YAG laser emitting at 2.1 μm, therefore, has been proposed as an alternative tool since adequate energy transmission in conventional optical fibers is feasible. The character of tissue interaction in Holmium Laser applications, however, is strongly determined by potentially relevant thermal effects [13,11,15]. Our goal was to optimize laser parameters for optical fiber delivered photoablative laser procedures. Tissue interaction should be comparable to what is known from so-called 3 μm lasers. Furthermore, clinically efficient transmission for conventional quartz optical fibers was desired.

MATERIALS AND METHODS

The laser used in this study is a solid state Cr^{3+}, Tm^{3+}, Er^{3+} dotated YAG laser (CTE:YAG) that emits radiation at a wavelength of 2.69 μm. The dimensions of the laser are: 30 cm × 8 cm × 9 cm. The length of the resonator is 20 cm. The laser is pumped by two flashlamps. The system is internally watercooled.

The prototype used has a rotating prism as a Q-switch. The resulting pulse duration is 0.5–2.5 μs, depending on the pumping energy. The maximum energy is 50 mJ/pulse (~ 10^5W). At a focus diameter of 200 μm, the maximal fluence reaches 150 J/cm^2. Repetition rates are 1–10 Hz. A Q-switched ErCr:YSGG laser emitting at 2.79 μm with a pulse duration of 200 ns was used for comparative analysis.

Energy transmission in low-OH^--fused-silica (Ceram Optec—LHFS, 0.3 ppm) optical fibers was tested for diameters of 200 and 600 μm. Energy was measured with a pyroelectrical Joulemeter (Gen-Tech, ED 200).

Ablation rates in soft (human and porcine cornea) and hard (human teeth) tissues were determined as a function of applied fluence. In the ablation rate measurement setup, the laser beam was delivered through air and focussed (f = 20 cm) on the target material. A diaphragm was used to select a sharp-edge irradiation zone of 200 μm diameter. Applied fluences ranged from 5 to 20 J/cm^2 in corneal tissues and 20–60 J/cm^2 in dental tissues.

For histological and micromorphologic examinations, each experiment was performed twice. One sample was determined for light microscopy (standard HE-staining), the other for scanning electron microscopy.

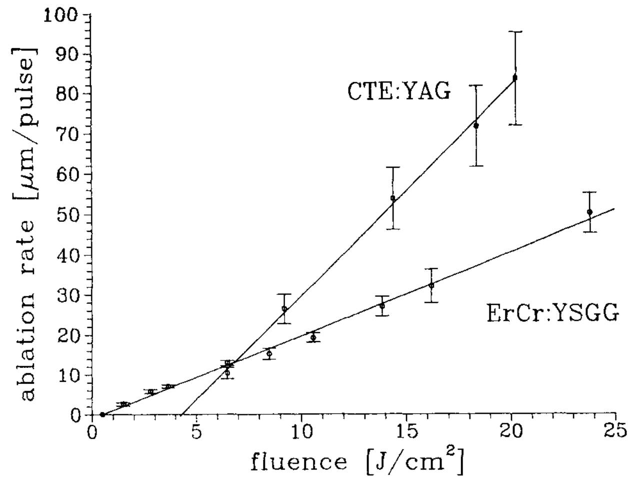

Fig. 1. The ablation rate (cornea) in μm/pulse as a function of applied laser fluence (J/cm^2) for CTE:YAG and ErCr:YSGG laser radiation. The ablation rate for Q-switched (1 μs) CTE: YAG laser radiation (2.69 μm) increased as a function of applied fluences from 5 μm/pulse (5.5 J/cm^2) to ~ 90 μm/pulse (18 J/cm^2). In comparison, ablation rates of Q-switched (200 ns) ErCr-YSGG laser (2.79 μm) radiation in corneal tissue ranged from 2 μm/pulse (3 J/cm^2) to 50 μm/pulse (24 J/cm^2).

RESULTS

CTE:YAG laser energy (2.69 μm) transmission in LHFS (OH^-–content: 0.3 ppm) optical fibers shows an attenuation of < 8 db/m. Maximal fluence at a laser output energy of 50 mJ/pulse in an optical fiber of 200 μm diameter and 50 cm lengths reaches 64 J/cm^2.

Absorption depth of 2.69 μm laser radiation in corneal tissue is 12.5 μm [5,23]. At clinically relevant fluences, the ablation rate per pulse increased as a function of applied fluences from 5 μm/pulse (5.5 J/cm^2) to ~ 90 μm/pulse (18 J/cm^2). The ablation threshold could be extrapolated to be ~ 4–5 J/cm^2. Below threshold fluences, only a whitening of the tissue surface (cornea) was noted, indicating that tissue water had evaporated leaving the protein matrix behind.

In comparison, ablation rates of Q-switched (200 ns) ErCr-YSGG laser (2.79 μm) radiation in corneal tissue ranged from 2 μm/pulse (3 J/cm^2) to 50 μm/pulse (24 J/cm^2) (Fig. 1).

Light-microscopical analysis of craters created in porcine corneal tissue, demonstrates the quality of tissue processing of the two different

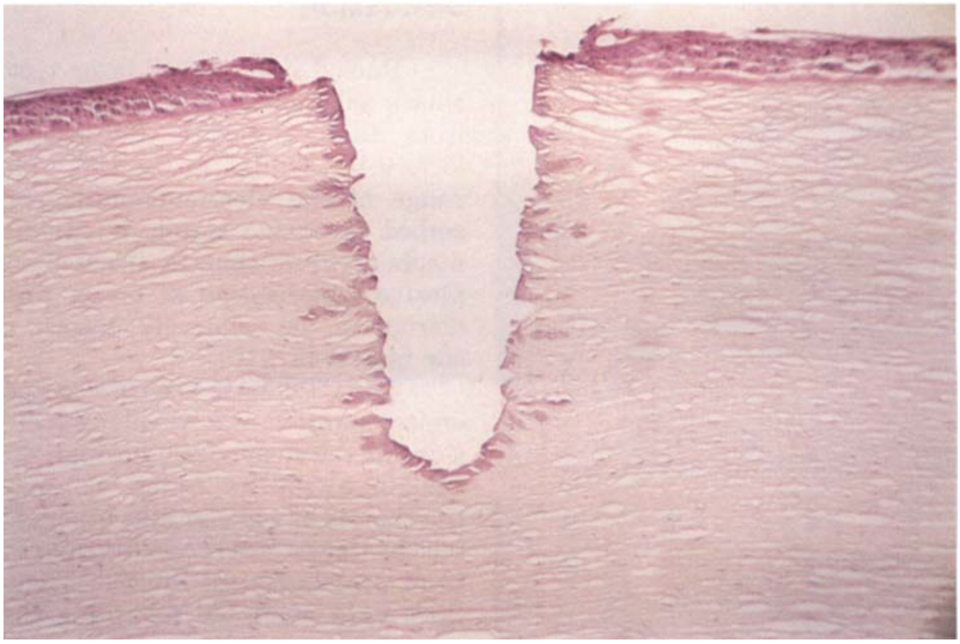

Fig. 2. Light-microscopical analysis of a crater (width: 250 μm) in porcine corneal tissue, processed with a Q-switched CTE:YAG laser. Applied fluence was 11 J/cm^2. Repetition rate was 1 Hz (20 pulses). Visible collateral damage extends 20 ± 5 μm.

lasers (Figs. 2, 3). The zone of collateral thermal damage was 8 ± 2 μm for the ErCr:YSGG laser (6 J/cm^2), and 20 ± 5 μm for the CTE:YAG laser (11 J/cm^2). In the fluence range of 5–20 J/cm^2, there was no significant difference in the expansion of collateral thermal alteration for both lasers. At higher fluences, deterioration of adjacent tissue appeared to be due to mechanical effects. The explosive character of vapour and fragmental ejection lead to a radial injection of rather large gas bubbles in between the collagen lamellae of the corneal tissue.

In comparison to soft tissue ablation, ablation rates (CTE:YAG) in dental tissue showed only little variance. At lower fluences (20 J/cm^2), ~ 1 μm/pulse of enamel material was removed. Tissue interaction was accompanied by luminescence at the interaction zone, indicating the induction of plasma. Due to plasma ignition, an increase of ablation rate beyond 2–3 μm/pulse was not possible in enamel, even at fluences as high

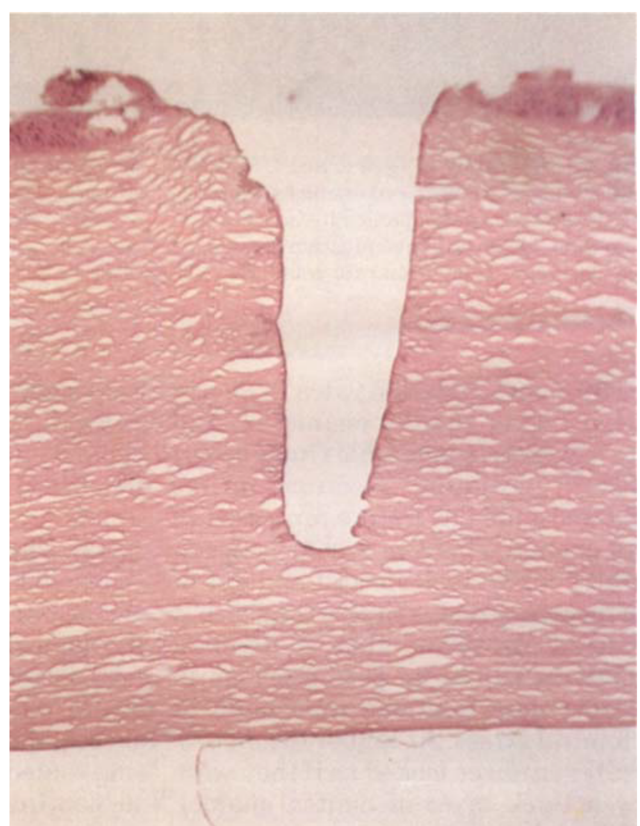

Fig. 3. Light-microscopical analysis of a crater (width: 250 μm) in porcine corneal tissue, processed with a Q-switched ErCr:YSGG laser. Applied fluence was 6 J/cm^2. Repetition rate was 1 Hz (50 pulses). Visible collateral damage extends 8 ± 2 μm.

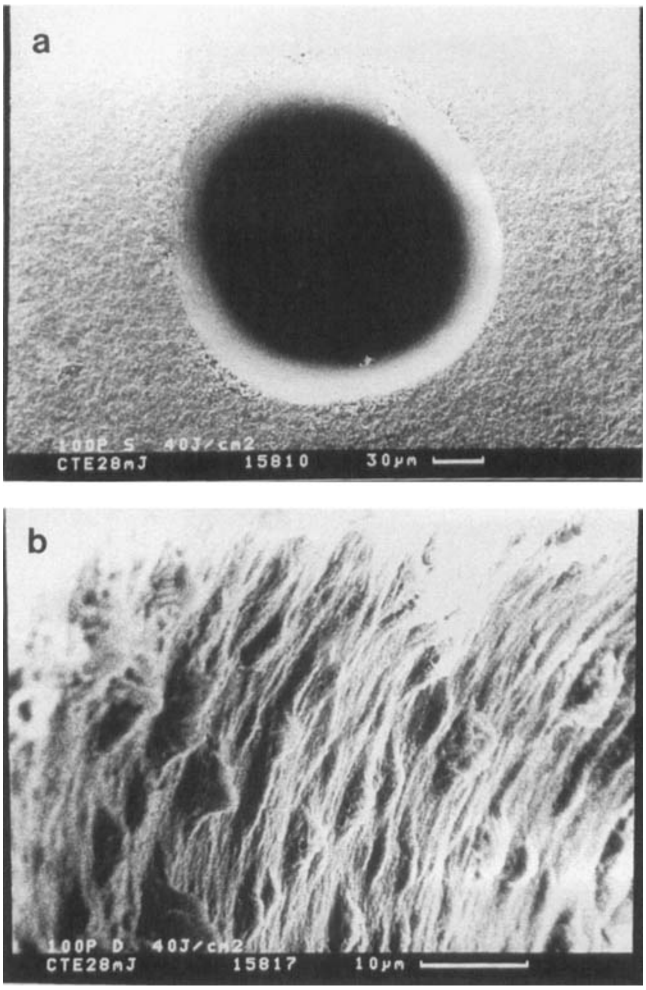

Fig. 4. SEM study of Q-switched CTE:YAG laser processed crater in dental tissue—**a)** enamel surface; **b)** dentine. Tissue interaction was accompanied by luminescence at the interaction zone, indicating the induction of plasma. Applied fluence was 40 J/cm^2. Repetition rate was 1 Hz (100 pulses).

as 60 J/cm^2. Ablation rates in dentine were somewhat higher than in enamel (2–5 μm/pulse).

When considerable fluences were applied (40 J/cm^2), scanning electron microscopy (SEM) showed that the craters processed in human dental enamel where very sharp edged (Fig. 4a). The crater walls were smooth and did not show signs of mechanic disturbances (Fig. 4b). It must be considered, however, that a detailed histological analysis has to be performed in order to detect deeper deteriorations that could be due to mechanical stress. At higher (60 J/cm^2) fluences, the crater surfaces looked as if they were being sealed by a thick layer of melted enamel and dentine material (Fig. 5).

DISCUSSION

Pulsed midinfrared-laser tissue interaction shows substantial differences to what is known from high intensity UV-laser photoablation [3,7,10,16]. Unlike in the UV, light in the 3 μm range of the electromagnetic spectrum is absorbed by tissue water, functioning as a chromophor, rather than by tissue matrix [5,23]. Explosive vaporization of tissue water leads to a disruptive, yet relatively smooth, removal of tissue matrix [3,7,10].

The CTE:YAG laser presented in this study emits radiation at a wavelength of 2.69 μm. Considering the steepness of the high water absorption peak in the spectral range ~ 3 μm, it appears that the choices of wavelengths other than those occupied by the Er:YAG (2.94 μm) and the ErCr:YSGG (2.74 μm) laser are limited. There is, however, little data available regarding the character of water absorption in the electromagnetic spectral range of 2.4–3.0 microns [5,22]. It must be taken into consideration that all data released is given for moderate temperatures (30°C) and atmospheric pressure, disregarding the dynamics of midinfrared laser photoablation. However, it is likely that the spectroscopic properties of water may deviate, due to extremely different ambient conditions during the dynamic ablative process [1,10].

During the interaction of the laser pulse with the irradiated tissue, a pressure front in the kilobar regime is built up at the ablation zone. The process that generates the pressure front is present as long as the laser irradiates the tissue (2.5 μs) and proceeds corresponding to the ablation rate (5.5–90 μm) into the irradiated tissue probe [1,10].

Since Q-switched laser pulses are extremely short (ns–μs) it is rather difficult to perform real-time temperature measurements. However, it is well known that with increasing pressures, the temperature necessary to lead to water vaporization increases to a critical temperature of 374°C. It can, therefore, also be assumed that due to the presence of pressure peaks in the kilobar regime, tissue temperatures have to be far above 100°C, at least during the incidence of the laser pulse.

Corresponding to the increase of temperature and pressure, Vodop'yanov [18] documented a spectral shift of the 3 μm water absorption peak into lower wavelengths accompanied by a decreasing absorption maximum (Fig. 6). This phenomenon is due to a decreasing influence of the hydrogen bonding in water [18].

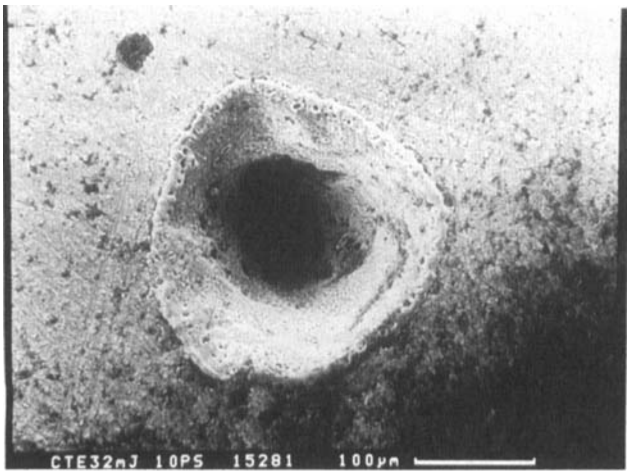

Fig. 5. SEM study of Q-switched CTE:YAG laser processed crater in dental enamel. Applied fluence was 60 J/cm^2. Repetition rate was 1 Hz (10 pulses). The crater surface is sealed by a thick layer of melted enamel and dentine material.

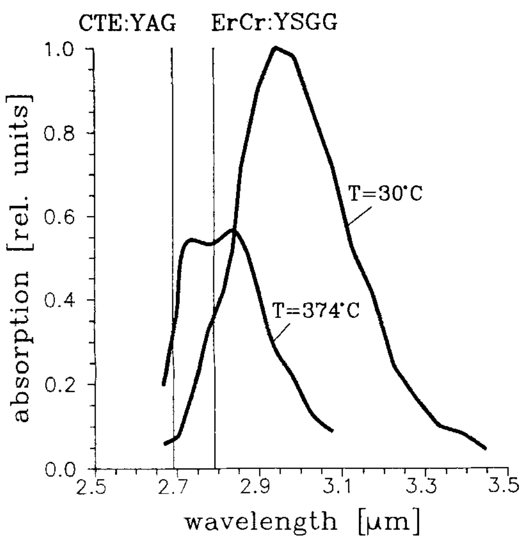

Fig. 6. Frequency shift of the water absorption peak at high temperatures according to the results of Vodop'yanov [18]. During ablation with Q-switched midinfrared lasers (CTE: YAG, ErCr:YSGG), high pressure of several hundred bars is build up within the interacting tissue volume [1,10]. The boiling temperature of interacting tissue water, therefore, is far above 100°C. The strong rise in absorption at 2.69 µm at higher temperatures is a possible explanation for the surprisingly small collateral thermal tissue damage created by the CTE:YAG laser. The effective penetration depth during laser interaction is probably considerably lower than at room temperature.

Such a behavior could explain why tissue effects of lasers emitting at different wavelengths in this rather critical portion of the spectrum are of such similarity. Comparing craters in corneal tissue created by a Q-switched ErCr:YSGG (wavelength: 2.74 µm) laser with those created by a Q-switched CTE:YAG (wavelength: 2.69 µm) laser, the difference in collateral tissue alteration is less enhanced than expected from the difference of the calculated absorption depth of both lasers (2.1 µm for ErCr:YSGG and 12.5 µm for CTE: YAG) [5].

In application of CO_2 laser radiation, the induced collateral thermal damage is nearly twice than in CTE:YAG laser application, although there are almost identical penetration depths in water (12.6 µm) and comparable (2 µs) pulse durations [19]. These facts support the assumption that the spectral behavior of water is influenced by the ambient conditions in the midinfrared spectral range.

Ren et al. [13] compared different laser wavelengths in the midinfrared with respect to the pulse duration. The collateral damage in Er:YAG (wavelength: 2.94 µm) and ErCr:YSGG laser processed corneal craters is almost equal ($<$ 10 µm) when Q-switched pulses of 100 ns duration are applied. In the free-running spiking mode (200 µs), the difference of wavelengths is depictable from a more enhanced collateral tissue damage ($\sim$ 15 µm for Er:YAG and $\sim$ 40 µm for ErCr:YSGG), which reflects the difference of absorption depths of both laser wavelengths [13]. It seems that the interaction time, and herewith the intensity of the incident laser pulse, strongly determines pressure and temperature within the interacting tissue volume influencing the absorption pattern of exposed tissue water.

Significantly higher fluences (20 J/cm^2) are required in order to achieve ablation in dental tissues. However, ablation rates are more efficient then in ArF excimer laser photoablation, where even at highest fluences the rate per pulse is below 0.5 µm [4]. The ablation rate data obtained with the CTE:YAG laser correspond to the findings of Hibst et al. [6] who have worked with a free-running Er:YAG laser. Enamel and dentine consist mainly of mineralogic material with strong chemical bond energies. The presence of plasma illumination during dental tissue processing with high fluences suggests that thermal ionization occurs.

There are a variety of different optical fibers presently available that are able to adequately transport midinfrared radiation. Only low hydroxy fused silica (LHFS) optical fibers, however, fullfil the clinical requirements of flexibility, nonsolubility, nontoxicity, and low costs. Transmission of ErCr:YSGG laser radiation (2.79 μm) in a 200 μm LHFS optical fiber (0.3 ppm, CeramOptec) of 50 cm lengths is as low as 1 to 2% compared to 40% energy transmission when CTE:YAG (2.69 μm) laser radiation is applied. Even less energy ($< 1\%$) is transmitted when an Er:YAG laser (2.94 μm) is used. In no way Er:YAG or ErCr:YSGG laser radiation can be transmitted in LHFS optical fibers with fluences sufficient for tissue-ablation when the laser pulses are Q-switched. The reason is that high fluences lead to interaction with the fiber material itself.

Laser pulses in the prototype used in this study are characterized by multimode output and pulse to pulse energy fluctuations. Certainly, with an electro-optical Q-switch modulator, the pulse characteristic of the CTE:YAG laser could be improved. Unlike the ErCr:YSGG and the Er:YAG laser, electro-optical Q-switching is easily available for CTE:YAG laser radiation due to better transmission of optical components of the pulse modulators. Furthermore, it has been shown that ultrashort pulses in the nanosecond range can generally improve the quality of tissue processing for midinfrared lasers [17].

In conclusion, it is emphasized that the laser parameter, conceptualized with the presented CTE:YAG laser, are appropriate for an experimental approach to a variety of intracorporal laser surgical applications. Both the solid-state technology and the lack of potentially hazardous radiation complement the CTE:YAG laser for medical applications. Although there is a slight increase in thermal damage range compared to Er:YAG and Er:YSGG laser application, the CTE:YAG laser appears to be a good compromise between optimal tissue penetration and optimal fiber transmission.

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