Lasers in Surgery and Medicine 11:569–579 (1991)

Studies in Fiber Guided Excimer Laser Surgery for Cutting and Drilling Bone and Meniscus

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Our experiments on transmitting high-power excimer laser pulses through optical fibers and our investigations on excimer laser ablation of hard tissue show the feasibility of using the excimer laser as an additional instrument in general and accident surgery involving minimal invasive surgery. By combining XeCl-excimer lasers and tapered fused silica fibers we obtained output fluences up to 32 J/cm^2 and ablation rates of 3 μm/pulse of hard tissue. This enables us to cut bone and cartilage in a period of time which is suitable for clinical operations. Various experiments were carried out on cadavers in order to optimize the parameters of the excimer laser and fibers: e.g., wavelength, pulse duration, energy, repetition rate, fiber core diameter. The surfaces of the cut tissue are comparable to cuts with conventional instruments. No carbonisation was observed. The temperature increase is below 40°C in the tissue surrounding the laser spot. The healing rate of an excimer laser cut is not slower than mechanical treatments; the quality is comparable.

Key words: **accident surgery, laser ablation, optical fiber, pulsed UV-laser**

INTRODUCTION

There is a large number of actual and potential excimer laser applications in medicine: e.g., angioplasty, ophthalmology, and dentistry. Great efforts have been made in cornea shaping in particular [1,2] and pulsed laser angioplasty [3,4]. In all cases the low penetration depth of ultraviolet radiation produces a locally restricted interaction. Only in the far-infrared, where the water absorption is high, can similar effects be seen. However, in the field of accident surgery and orthopedics there is so far little demand for excimer lasers [5–13] and for CO_2-lasers. The use of "flexible" (but actually rigid) light guides is not suitable for difficult interventions because the handling of mirrors, lenses, and light pipes is very poor. In accident surgery, however, no serious possibility of using any laser can be envisioned unless the problem of beam delivery is solved. While there is no chance to guide a CO_2-laser via non-toxic optical fibers, fused silica is transparent for the XeCl excimer laser radiation, and therefore flexible waveguides for excimer laser application can be made out of quartz.

Meniscectomy, for example, or removal of hypertrophic callus with mechanical instruments in post fractured areas near nerves or vessels is often very difficult and even dangerous using common mechanical instruments, e.g., pincers (Luer) and chisels. The combination of an effective fiber with the high-power excimer laser might be a new atraumatic tool for accident surgery using the interaction of short UV-laser beam

Accepted for publication August 15, 1991.

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pulses with solid material. Photoablation is the combination of breaking the bonds of solid material by high energy UV-photons (photodecomposition) and explosive removal of the material in small fragments by storing high intensity in a very thin area. This is caused by the short high-power laser pulses (a few gigawatts in some nanoseconds) and high absorption of organic material in the ultraviolett spectral range [14], which might even be improved by doping [15].

For optical fibers the available fluence at the end of the beam delivery system is limited by the attenuation and optical damage of fused silica. The average damage threshold of optical fibers is 20 J/cm^2 for XeCl-excimer laser (wavelength 308 nm) and 28-ns pulse duration [16], but it is hardly possible to transport more than 30 mJ through a 600-μm light guide without taking extraordinary care to the coupling parameters and fiber preparation. Therefore the ablation rate of bone does not exceed 0.2 μm/pulse [17], which is definitely not enough for successful surgery [5–9,12].

The undoubted advantages of the "athermal" treatment of materials by the UV-pulses of the excimer laser and the progress in fiber optics in the last years encouraged us to begin a new attempt at excimer laser surgery. To be able to replace some of the conventional tools in accident surgery and orthopedics we had to find a way to handle the laser beam conveniently and without problems for the surgeon as well as for the patient. It was necessary to improve the delivery system for transmitting high power in order to increase the ablation rate more than one order of magnitude. Finally, we must get good knowledge of the suitable laser parameters (energy, power, fluence, repetition rate, pulse duration, etc.) for the different kinds of application. It is our aim not only to study the interaction of the excimer laser beam with various tissue, but also to develop all tools that are necessary for clinical accident surgery.

MATERIALS AND METHODS

Fiberoptical Studies

First of all we have studied the high power excimer laser transmission of a large number of optical fibers at different wavelengths (193 nm, 248 nm, and 308 nm) and different pulse durations (28 ns, 60 ns, 300 ns) [16]. The attenuation of fused silica fibers is more than 2.0 dB/m for excimer lasers at 248 nm and 0.15 dB/m at 308 nm wavelength; it is almost nontransparent for 193 nm (approximately 100 dB/m). The transmission of high power energy is predominantly limited by the damage thresholds of the front and rear surfaces. These are very sensitive to any surface preparation. Every effort has to be made to get a homogeneous flat top spatial beam profile. Under optimized conditions the optical damage threshold decreases from 20 J/cm^2 at 308 nm, 8 J/cm^2 at 248 nm, to 6 J/cm^2 at 193 nm [16]. The damage threshold increases with the square root of the pulse duration. Via standard fiber, however, the transmitted energy is not enough for the aimed application of cutting and drilling bones and cartilage, while it works quite well in ophthalmology [18].

For this reason we used a new type of wave guide geometry especially developed for transmitting high-power laser beams [19,20]. In the front part of the fused silica fiber with a fluorine doped cladding, the cross section is gradually enlarged up to a diameter of 10 mm within a length of less than 100 mm (Fig. 1). Although the energy is kept constant, this lowers the fluence drastically and no surface damage was observed any more. Those tapered fibers were used for delivering peak power of 2 GW/cm^2 and more at 308 nm with various core diameters of 200 μm to 1,100 μm. We were able to transmit more than 250 mJ output energy (pulse width 28 ns) by a 1,000-μm fiber without damage to the front surface or the bulk (i.e., fluence of 32 J/cm^2); the coupling out was done in water. By splicing together tapered fibers and normal fused silica fibers the high-power excimer laser radiation can be guided over a long distance. Gold-coated fibers are available and are best for sterile applications.

Similar experiments were performed at shorter wavelengths (193 nm, 248 nm), where lower ablation thresholds and higher ablation rates are expected. However, the maximum laser fluence transmitted through tapered fibers is still too low for reasonable applications, because the absorption increases non-linearly at higher intensities for shorter wavelengths due to two photon absorption [16,21]. In addition, photodegradation (i.e., aging at high counts) of fused silica fibers has to be taken into account at high intensity–low wavelength radiation [16]. Finally the use of 248-nm radiation is not suitable for medical purposes, since mutagenic and cancerogenic side-effects are supposed.

The optical setup of our experiments is shown in Figure 2. The energy of the excimer laser was continuously varied by a stepping motor

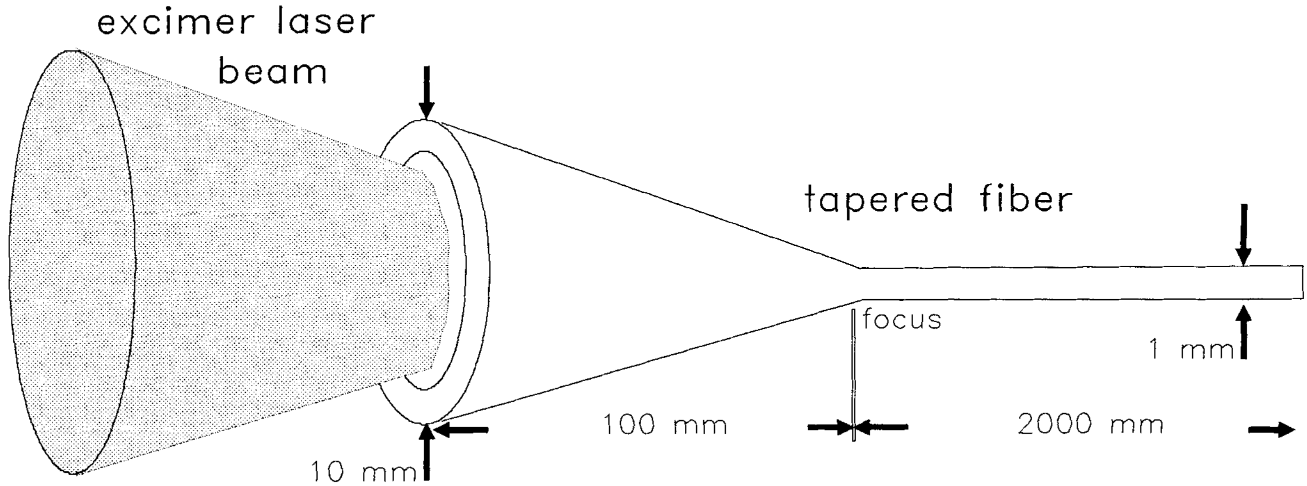

Fig. 1. Optical design of coupling an excimer laser beam into the tapered fiber.

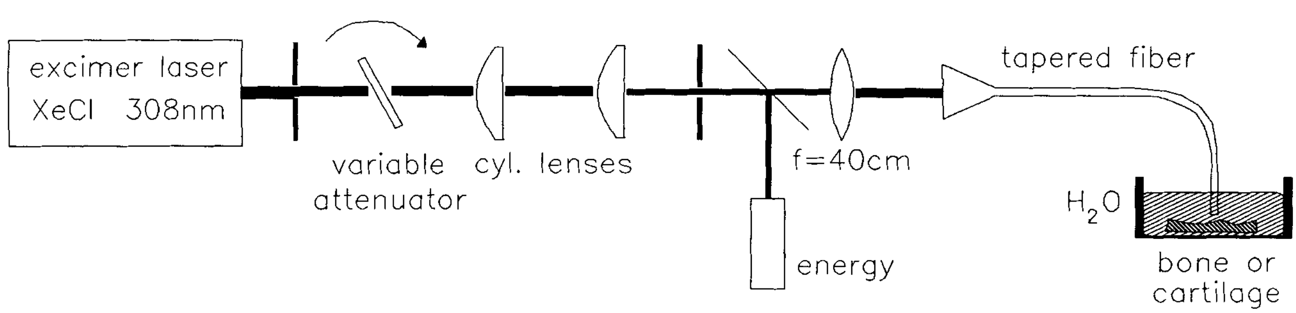

Fig. 2. Setup of our experiments on tissue ablation by an excimer laser guided through a tapered fiber.

driven attenuator (Laser-Laboratorium Göttingen) and the input energy was measured online. The temporal profile was checked by a fast photodiode (Hamamatsu R1193U-02) in combination with a digital storing oscilloscope. With a telescopical design of two cylindrical lenses the divergencies of the beam (25 mm × 12 mm and 1–2 mrad) were assimilated in both directions. The spatial beam profile could be measured simultaneously by a UV-beam profiler system (Laser-Laboratorium Göttingen). After passing a circular aperture the excimer laser beam was focussed into the tapered part of the fiber (i.e. behind the front surface: Fig. 1) by a spherical lens of about 400 mm focal length. The spot size of the beam inside the bulk material should be as large as the core diameter of the light guide.

Ablation of Hard Tissue and Cutting of Bone and Cartilage

In the last few years much progress has been made in cutting soft tissue with lasers, especially with the well-established CO_2-laser. But in accident surgery much of the injured tissue is quite hard and relatively water free: e.g., bone, cartilage, etc. The suitability of the excimer laser for cutting hard tissue has been shown recently for different wavelengths but without using an optical fiber [5–9,12]. In optimizing bone and cartilage ablation we tried different excimer laser systems (Lambda Physik EMG 1003i, LPX 210iCC, LPX 605iCC, EMG 602): wavelength 308 nm (XeCl); pulse length 28 ns, 60 ns, and 300 ns; laser output energy up to 400 mJ; repetition rate up to 250 Hz. The beam was guided to the operation site through various tapered fibers (Heraeus Quarzglas) with core diameter of 400 μm, 600 μm, and 1,000 μm and a overall length of about 2 m. Our experiments were carried out with an energy of 20–70 mJ per pulse, the fluence was 3–18 J/cm^2, the applied peak intensity 14–400 MW/cm^2. The quality of the rear surface of the fiber was checked before each treatment.

The studies were done on bone (rib) and menisci from just-slaughtered pork and beef. The prepared samples were placed under 10 mm of fresh water. During the laser ablation the fiber tip was always in contact with the tissue. The experiments on cutting bones were carried out under water by moving the sample on a mobile operating stage with constant velocity of 2–6 mm/s passing the fiber tip at a distance of 0.1 mm. The depth of holes and cuts was measured with an optical microscope (Zeiss Axioscope).

Thermal Effects During Excimer Laser Ablation of Tissue

During the excimer laser treatment the temperature rise of the adjacent tissue was measured with a miniaturized thermocouple made out of copper/constantan; the diameter of the wires was 0.1 mm (Omega). The thermocouple was placed inside the bone or cartilage at a known distance from the ablation site. For good thermal contact a thermal compound was used.

Healing Study

The first experiments to study the healing of bones drilled by the excimer laser were performed on rabbits. Six adult male rabbits were anesthetized with intramuscular injections of Rompun (5 mg/kg body weight) and Ketamin (50 mg/kg body weight). The right tibiae of each animal were surgically exposed. Sharp dissection was carried down through skin and fascia to bone with a scalpel blade. The muscle tissue was retracted circumferentially. With either the excimer laser (pulse width 300 ns, repetition rate 20 Hz, applied energy 50 mJ, fiber core diameter 1,090 μm, fluence 5.4 J/cm^2) or a mechanical drill bores were performed right through the tibiae. The laser borehole was placed at the lateral surface of the tibia. The medial surface of the tibia was drilled with a mechanical bore. The animals were sacrificed after 2 and 4 weeks postoperatively. For 10 days the bone specimens were fixed in 4% formalin, then washed in water. No decalcification can be seen. The specimens were dehydrated in an ascended alcoholic line, embedded in methylmetacrylate, sectioned at 8 μm, and stained after Goldner [29].

RESULTS

Ablation of Hard Tissue

Figures 3–5 show the etch depth of holes in bone tissue drilled by different fiber guided excimer lasers with pulse durations of 28 ns, 60 ns, and 300 ns. A tapered fiber with 1,000-μm core diameter was used; the irradiation time was 60 s. The repetition rate rises from 20 Hz to 50 Hz, the applied energy varies from 20 mJ to 50 mJ at the output of the fiber; i.e., the fluence increases from 2.1 J/cm^2 to 5.4 J/cm^2. Since no standard bone material was used but as-grown beef, the results are affected by the inhomogeneity of the tissue. An additional error is caused by the measurement of the depth of the holes; it can be estimated to within 30%, but not more than 100 μm. The reported values are real data, with no averaging.

Up to an energy of 40 mJ and for pulse rates up to 30 Hz the bone ablation of excimer lasers for pulse duration of 28 ns and 300 ns are comparable, with a slightly higher ablation rate of the short-pulse laser (Figs. 3, 4). The etch depth obtained with the long pulse laser (300 ns) increases with energy rising to 50 mJ and a repetition rate of 50 Hz (Fig. 4). The surface of the fibers showed no damage, even at higher energy and pulse rate. The holes' diameter were in the range of the diameter of the used light guide, e.g., 1,000 μm.

While it was no problem to transmit 50 mJ with 50 Hz for ablation with the long-pulse laser radiation, we were not able to ablate hard tissue with any higher repetition rate and energy by the short-pulse laser. Usually the distal end of the fiber was damaged by shock waves or similar mechanical effects when the ablation was done with energy above 40 mJ and repetition rates larger than 30 Hz. Note, however, that no damage of the

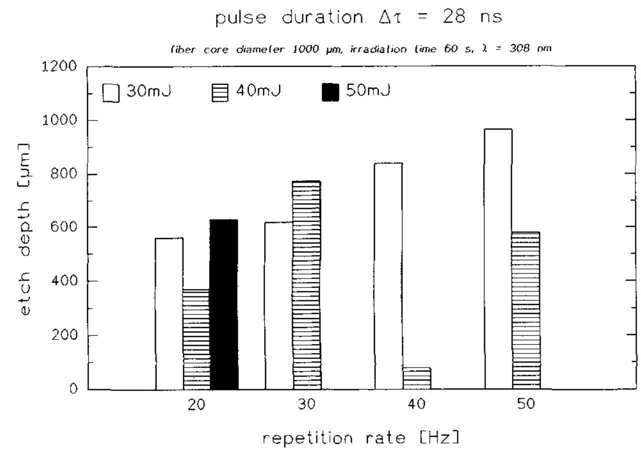

Fig. 3. Etch depth of holes drilled in bone with the 28-ns XeCl-excimer laser (wavelength 308 nm) transmitted by a tapered fiber with 1,000-μm core diameter. The exposure time was 60 s, *the repetition rate increased from 20 Hz to 50 Hz*, the applied energy was varied from 30 mJ to 50 mJ.

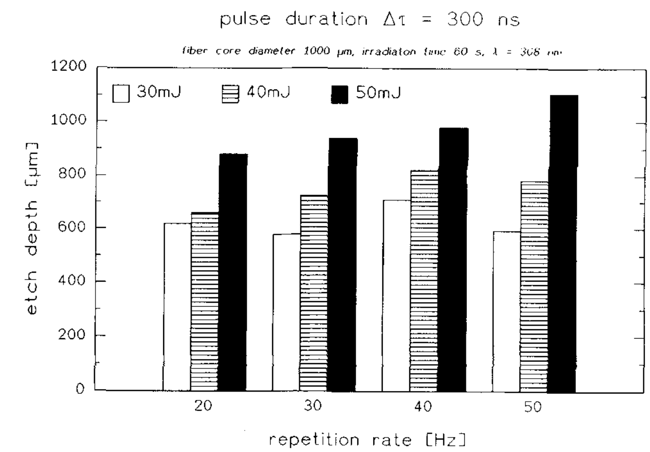

Fig. 4. Etch depth of holes drilled in bone with the 300-ns XeCl-excimer laser (wavelength 308 nm) transported through a tapered fiber with 1,000 μm core diameter. The *exposure time was 60 s, the repetition rate increased from 20 Hz to 50 Hz*, the applied energy was varied from 30 mJ to 50 mJ.

fiber is observed without ablation of hard tissue; even soft tissue ablation didn't cause any damage. Therefore, excimer laser radiation of about 30-ns pulse length is not suitable for fiber guided ablation of hard tissue.

Up to now, excimer lasers with pulse durations of more than 150 ns have rarely been available, have had a limited output energy, and have required good maintenance. In order to combine the advantages of photoablation with high energy and fiber durability, we used an excimer laser with 60 ns pulse length. Figure 5 shows the ablation depth obtained in bones with 60 ns pulses for different repetition rates and applied energies. Clearly, in the range from 30 mJ to 50 mJ considerably deeper bores were achieved in comparison to Figure 3 and Figure 4. At 60 s exposure time, for example, the maximum bore depth effected by 50 mJ output energy was about 3.5 mm, using a repetition rate of 40 Hz (Fig. 5); this corresponds to an ablation rate of 1.5 μm/pulse. The fiber was usually damaged at a higher energy and repetition rate (e.g., 50 mJ and 50 Hz). For medium rates and energies the 60-ns pulse was found to be best for hard tissue ablation and fiber protection. The ablation is much more effective with the 60-ns pulses compared to 300-ns pulses, and damage to the fiber does not occur as frequently as with 28-ns pulses.

The ablation rate of bone as a function of applied fluence is plotted in Figure 6 for different repetition rates. The 300-ns series was used because more data were available which also cover the high-fluence range. In principle the results are similar for the other lasers [11]. Below 40 Hz no saturation of the ablation rate of hard tissue was observed up to 20 J/cm^2; more than 3 μm/pulse were possible. This means a drilling speed of 0.08 mm/s. Obviously, higher repetition rates lead to lower ablation depth per pulse. This is caused by the long duration of the ablation process, which is not immediately finished when the excimer laser pulse has stopped after 300 ns. At about 1 ms to 2 ms after the laser pulse a kind of bubble starts developing from the tissue surface and blows up for about 200 μs before it collapses.

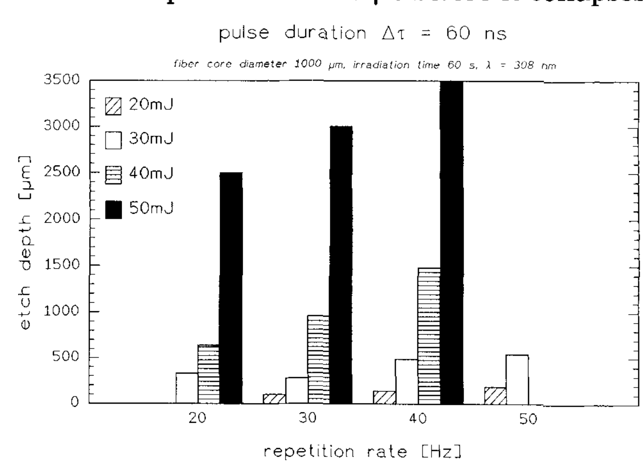

Fig. 5. Etch depth of holes drilled in bone with the 60-ns XeCl-excimer laser (wavelength 308 nm) guided through a tapered fiber with 1,000 μm core diameter. The exposure time was 60 s, the repetition rate increased from 20 Hz to 50 Hz, the applied energy was varied from 20 mJ to 50 mJ.

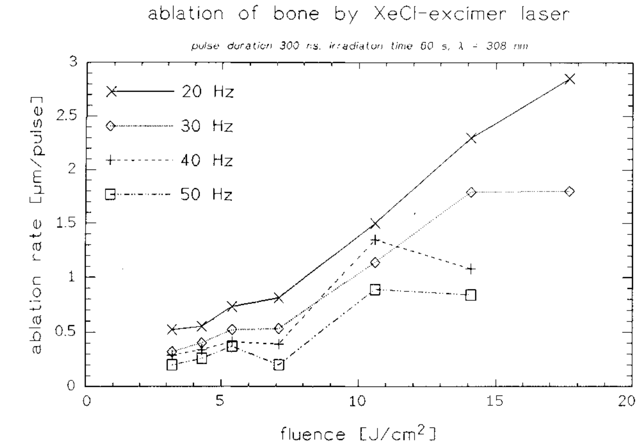

Fig. 6. Example of the ablation rate as a function of the fluence for different repetition rates (20 Hz up to 50 Hz). The experiments were performed on bone with the 300-ns XeCl-excimer laser (wavelength 308 nm) and 60-s exposure time.

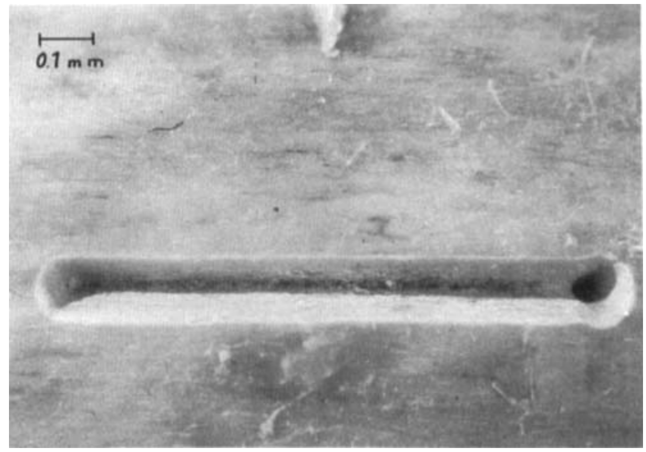

Fig. 8. Bone cut by an excimer laser beam. The edges are sharp, the walls are clean; no carbonisation can be observed.

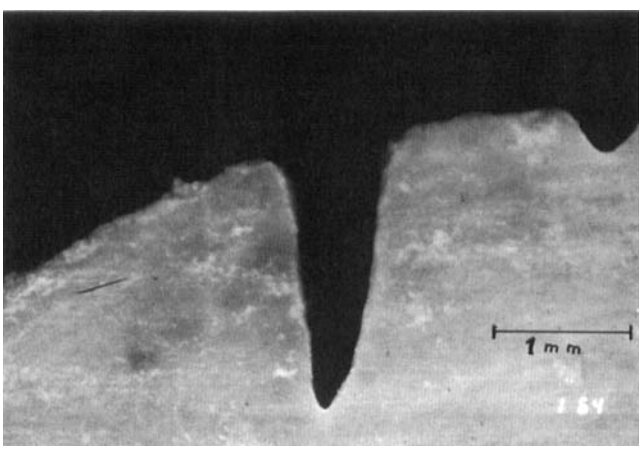

Fig. 7. Profile of cuts in bone carried out with the excimer laser transmitted through a tapered fiber with 600-µm core diameter.

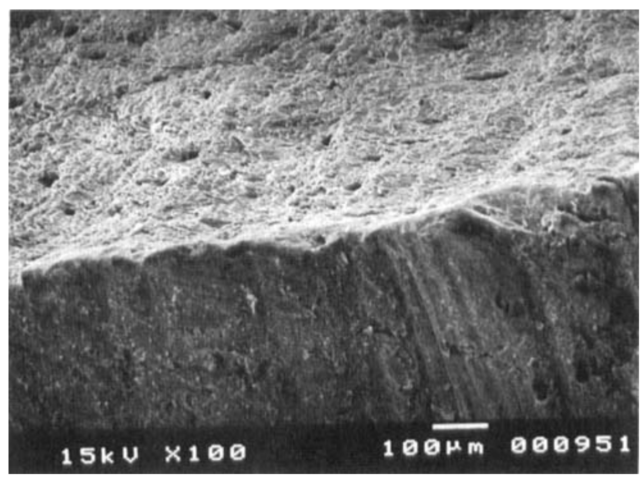

Fig. 9. Scanning electron microscope photograph of a bone tissue. The upper part was cut by the excimer laser: clean surface and no lubrication of the tissue. The haversian canals have been opened smoothly. The lower part shows the cutting by a conventional saw.

But it can take much longer before the ablated particle are completely blown away [22,23].

Cutting Bone and Cartilage

We were able to cut bone with the laser beam guided through the optical fiber. The ablation rate exceeds 2 µm/pulse. Of course, in this case no problems with the ablated material even in deep slits have to be mentioned, since the fragments can disperse at both sides. The cutting profile (Fig. 7) is quite good, and comparable to saw cuttings, except for the conical bottom. This is typical [6,11] when moving a circular laser beam with flat top intensity profile along one direction while ablating the material. Using a ring structure spatial beam profile with lower intensity in the central part, an almost rectangular cut is attainable [24]. Cuts of more than 4.5 mm depth show clean, sharp, and smooth edges, without any colour alternation or carbonisation (Fig. 8). The clean removal of bone is the main advantage of the excimer laser [5–13]. Figure 9, taken by a scanning electron microscope (SEM), shows in the upper part of the surface of a bone cut by the laser beam, and in the lower part cut by a conventional saw. The saw cut is characterized as a surface with clear streaky marks of the saw. In contrast, the laser cut shows no marks of this kind: its surface

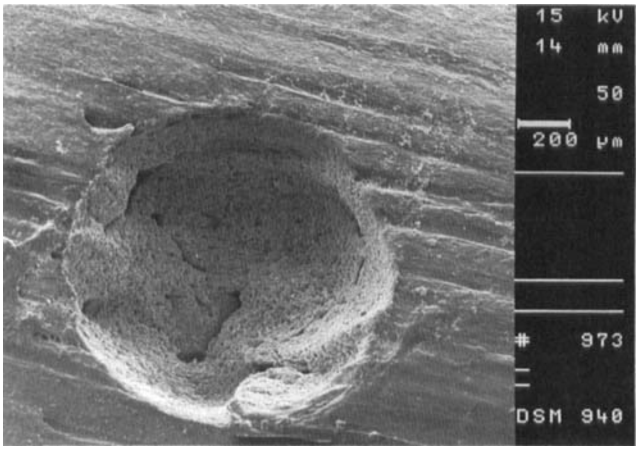

Fig. 10. Scanning electron microscope photograph of a laser borehole in bone tissue. The edges were burrless without melting on. The bottom shows no strange features which could be assigned to a thermal interaction or chemical reaction.

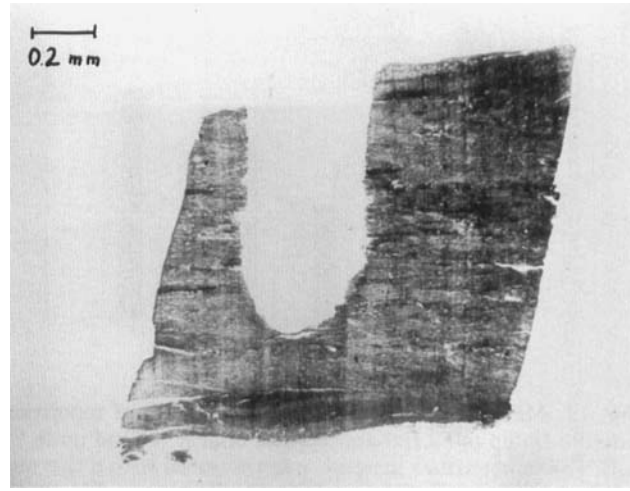

Fig. 11. Histologic photograph of excimer laser borehole in bone tissue. These specimens were embedded in methacrylate, sectioned at 8 µm and stained with toluidine blue, ×25. Parameters: energy 50 mJ, repetition rate 20 Hz, exposition 60 s, fiber core diameter 1,000 µm. A little necrotic zone (25–40 µm) resulted. No carbonisation zone. Depth about 1,000 µm.

is completely smooth. No thermal interaction, lubrications, or weldings of bone tissue are visible. No products of any chemical reaction can be observed. The numerous opened haversian canals show sharp borders and no damaged lamellar structures, since the laser treatment is hardly mechanical. Also, the edges of bores (Fig. 10) are smooth and burrless without damage to the adjacent tissue. Inside the crater walls are even and some bone canals are opened. No mushroom-like features were observed which could be assigned to thermal interaction or chemical reaction [12].

Histological studies of bone of fresh-slaughtered cows show a zone of only 25–40 µm of cells destroyed by the irradiation (Fig. 11). This may be caused by thermal effects if energy is below the ablation threshold [6]. Our results agree with the values reported by several other groups [6,7,8,25]. The thermal damage, however, decreases for shorter wavelength [12]. No carbonisation is seen at all. The crater shows a conical profile with sharp edges. The bone material is always completely ablated within the bores or cuts. As already reported by Lustmann et al. [12], the quality of the cutting surface is not affected by changing the fluence or the number of pulses. It depends neither on the repetition rate nor the pulse duration.

The drilling speed in meniscus was up to 6 mm/s. The recently reported results of other groups [9] are more than three orders of magnitude below the values taken from our experiments [10,11]. We were able to cut a meniscus of a pig (area of 4 mm × 13 mm) in 110 s. This operation velocity is reasonable for clinical operations. In complicated cases, such as knee-joint injuries of the rear meniscus, the intervention can be performed using fiber guided laser technology, which is much less traumatic than the conventionally used mechanical cutting techniques.

Thermal Effects During Excimer Laser Ablation of Tissue

In Figure 12 the increase of temperature during the lasering of the meniscus is plotted as a function of the repetition rate and applied energy. The experiments were carried out with the 60-ns excimer laser, the ablation was done under fresh water at 20°C, and the temperature was measured at a distance of 1 mm from the laser spot. Using medium parameters for the energy (e.g., 40 mJ) and especially for the pulse rate (e.g., 40 Hz), the achieved temperature difference is less than 40°C. The temperature is proportional to the inverse distance, and the gradient is about 0.3 mm^{-1} (Fig. 13). At a distance of 2 or 3 mm the temperature increased only 5°C. The necrotic zone is less than 50 µm, as shown by the histological studies (Fig. 10). This is comparable to the diameter of single cell [6,12].

In rather dry tissue, as bones for example, the temperature rise, however, was quite large.

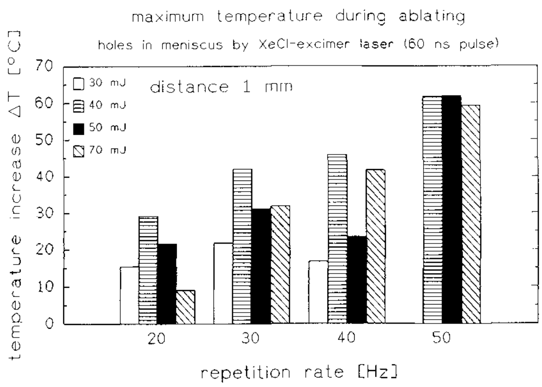

Fig. 12. Maximum temperature as a function of repetition rate (20 Hz up to 50 Hz) and applied energy (30 mJ up to 70 mJ). The temperature increase was measured with a thermocouple in a distance of 1 mm from the laser spot. The 60-ns excimer laser was drilling holes in meniscus via a tapered fiber of 1,000-μm core diameter. The experiments were performed in water of 20°C.

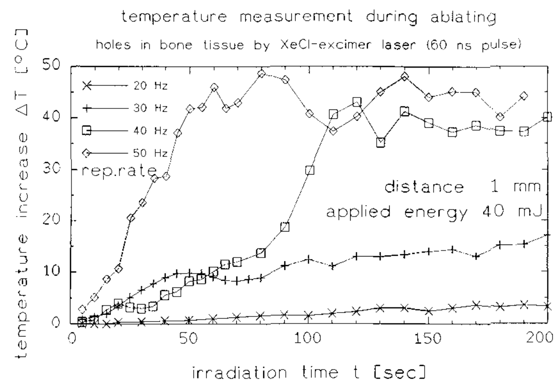

Fig. 14. The temperature elevation is a function of the irradiation time, depending on the repetition rate. The 60-ns excimer laser was ablating holes in bone tissue with an applied energy of 40 mJ guided through a 1,000-μm fiber. The experiments were performed under water at 20°C.

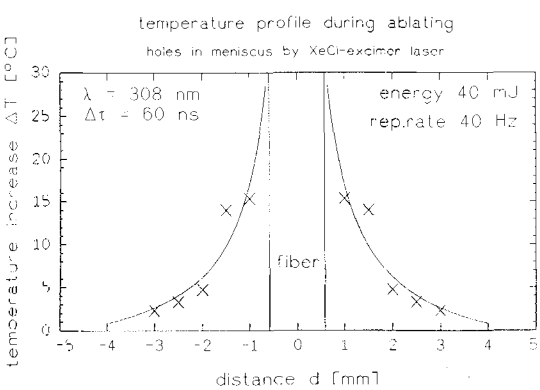

Fig. 13. The temperature profile in the area of the fiber tip, when an XeCl-excimer laser beam (wavelength 308 nm) of 40 mJ pulse energy, 40 Hz repetition rate, and 60 ns pulse duration is ablating meniscus.

For high energy (50 mJ) and high repetition rate (50 Hz) more than 200°C were measured at the tip of the fiber, but decreasing rapidly in the surroundings. The excimer laser ablation is not totally athermal. However, it is well known that working with mechanical tools (drill, saw) can also lead to temperature rises of 80°C to 100°C in bone [26]. In Figure 14 the time-dependent temperature increase is plotted for different values of the repetition rate, to which the absolute rise is very sensitive; nevertheless a saturation is always seen after 300 pulses. It has to be pointed out that the laser experiments have to be carried out under water. Bores drilled in air show carbonized surroundings depending on the applied energy and repetition rate.

The error in the temperature measurement is less than 1°C. Since the results are taken by on-line measurements during the drilling, however, no equilibrium state was attained. To demonstrate the fluctuations no averaging of the results was done in Figures 12 and 14. A comparison of different measurement series gives error bars of 5–10°C, but this is due to the inhomogeneity of the bone.

Healing Study

A detailed analysis of the healing process after excimer laser treatment is not yet possible. We wish however to briefly report our impressions after having treated six rabbits simultaneously with excimer laser and mechanical instruments. For years the general problem of the laser osteotomy was the delayed healing time compared to the mechanical osteotomy. This was related to a extremely high temperature during the laser intervention. By using a liquid medium for application of the excimer lasers beam the histological results show the negligible influence of the small temperature increase (40°C) for bone healing.

At 2 weeks postoperatively the bores show similar healing trends after laser application (Fig. 15) just like after mechanical drilling (Fig. 16). In the canal drilled by the excimer laser beam no carbonaceous material is revealed. The cell formations come into direct contact with the bore edges. No defects or vacancies can be observed, just like the mechanical treatment. In both samples a bridge-like shape of the new bone tissue can partly be observed as in immature bone tissue. The drilling hole is partly filled with fibrous tis-

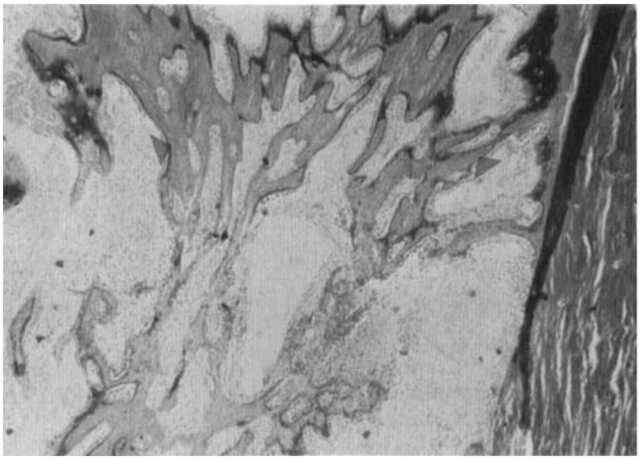

Fig. 15. Photomicrograph of excimer laser borehole in rabbit tibia. Goldner, ×15. Two weeks postoperatively. The dark grey islands show new bone tissue, which has a bridgelike shape, as in immature bone (arrows). The cell formations come into direct contact. No significant difference can be observed compared to the mechanical case (Fig. 16). No thermal necrosis is seen.

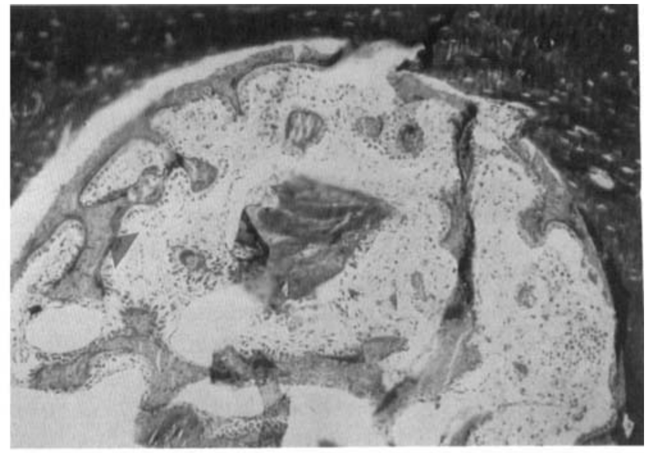

Fig. 16. Photomicrograph of mechanical bore in rabbit tibia. Goldner, ×15. Two weeks postoperatively. The dark grey islands show newly formed bone tissue (arrow).

sue and osteoid elements. Dark grey islands in Figures 15 and 16 show mineralized bone tissue. The amount of fibrous tissue as well as osteoid formation is similar. After four weeks postoperatively the laser bores as well as the mechanical bores have been well filled with osteoid and mineralized bone tissue. No delay of the healing process can be observed. The laser treatment leads to no irregularity in the osteoid formation.

Summing up, there is no significant difference in healing rate and quality between the laser bore and the mechanical bore observed in this first investigation. After only 2 weeks new bone forms on both the endosteale and the periosteale surfaces. The results of the experiments performed on animal tissue indicate that laser surgery has taken a large step toward partially replacing mechanical instruments for bone drilling and cutting. However, more extensive studies have to be carried out in order to confirm these preliminary results.

DISCUSSION

For the first time high-power excimer laser pulses of more than 30 J/cm^2 with 308 nm wavelength could be transmitted through a light guide by the use of tapered fused silica fibers. This is a breakthrough in fiber guided excimer laser surgery. Our results are comparable to recently reported experiments using a non-guided excimer laser beam [6,7,12]. With energies in excess of 50 mJ at the output of a 1,000-μm core fiber, precise cutting of bone, cartilage, and other hard tissue is possible in a suitable time frame for clinical surgery. The surface of the cut tissue is comparable to that cut with conventional instruments. No carbonisation was observed; the temperature increase of the tissue in the surroundings of the laser spot is below 40°C. There is no delay in the healing process.

Tapered fused silica fibers can only be used for transmitting high-power excimer laser radiation if the light emerges into water, especially when ablating hard tissue. Otherwise the distal side of the fiber will be damaged and carbonisation is observed at higher repetition rates. However, this is not a restriction for most applications in accident surgery. Since longer pulse lengths decrease the damage threshold of the fiber, a medium pulse duration is found to be most suitable for good ablation. The repetition rate must not exceed 40 Hz in order to avoid unwanted heating of the tissue without increasing the ablation speed. The applied fluence, however, could be raised since no saturation of the ablation rate has yet been observed. For this purpose, however, the quality of the excimer laser beam has to be improved or the laser output energy has to be extended. In each case improvements at the fiber output are necessary to avoid any damage. A better mechanical stability of silica fibers with a large diameter of about 1 millimeter would be desirable.

The next step will be to improve the handling of the fiber guide, and to find a way to reduce any remaining fiber damage produced by mechanical effects, in order to apply more than 50 mJ with high repetition rates. Perhaps we have to work in a non-contact way with those high-power transmitting fibers. Nevertheless, the results of our studies on excimer laser ablation of hard tissue show the feasibility of using the excimer laser as an instrument for precisely calculable operations, not only for microsurgery but for many special manipulations in narrow areas, e.g., resection of the rear meniscus. The operating speed obtained in meniscus ablation (6 mm/s) is already reasonable. The laser-fiber technology is much less traumatic than conventional cutting techniques. It is a big advantage that UV-laser radiation is applicable in aqueous solution, in contrast to CO_2-lasers. The way to a minimal invasive accident surgery has to improve endoscopic surgery needing a liquid operation medium.

In addition we are working on a diagnostic system to distinguish between different kinds of tissue during the ablation. The ablated tissue can be identified either by simple fluorescence spectroscopy or by using an additional dye laser for resonance fluorescence spectroscopy of a specific material [27,28].

ACKNOWLEDGMENTS

We would like to thank Dr. U. Grzesik, Dr. K.F. Klein, H. Fabian, and Dr. J. Kesper; this work was supported by Heraeus Quarzglas (Hanau, Germany) and Lambda Physik (Göttingen, Germany). The authors also wish to thank Prof. G. Delling, Department of Osteopathology, and Prof. W. Lierse, Department of Neurosurgery of the University of Hamburg, for their cooperation of the histological and scanning electron microscopical slides.

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