Comparison of Different Focusing Fiber Tips for Improved Oral Diode Laser Surgery

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Background and Objectives: State of the art for use of the fiber guided diode laser in dental therapy is the application of bare fibers. A novel concept with delivery fiber and exchangeable fiber tips enables the use of tips with special and optimized geometries for various applications. The aim of this study is the comparison of different focusing fiber tips for enhanced cutting efficacy in oral surgery.
Material and Methods: For this purpose various designs of tip geometry were investigated and optimized by ray tracing simulations. Two applicators, one with a sphere, and another one with a taper, were realized and tested on porcine gingiva (diode laser, 940 nm, 5 W/cw; 7 W/modulated). The cutting depth and quality were determined by light microscope. Histological sections of the cuts were prepared by a cryo-microtome and microscopically analyzed to determine the cut depths and thermal damage zones.
Results: The simulations show that, using a sphere as fiber tip, an intensity increase of up to a factor of 16.2 in air, and 13.2 in water compared to a bare 200 μm fiber can be achieved. Although offering high focusing factor in water, the cutting quality of the sphere was rather poor. This is probably caused by a derogation of the focusing quality due to contamination during cutting and light scattering. Much better results were achieved with conically shaped fiber tips. Compared to bare fibers they exhibit improved handling properties with no hooking, more regular and deeper cuts (5 W/cw: 2,393 ± 468 μm, compared to the cleaved bare fiber 5 W/cw: 711 ± 268 μm). The thermal damage zones of the cuts are comparable for the various tips and fibers.
Conclusions: In conclusion the results of our study show that cutting quality and efficiency of diode laser on soft tissue can be significantly improved using conically shaped fiber tips. Lasers Surg. Med.

Key words: diode laser; dentistry; oral surgery; fiber tips; soft tissue; cutting efficacy

INTRODUCTION

The diode laser is a well established tool in dental therapy [1–3]. The most important applications are surgical procedures, e.g. frenectomy or uncovering of implants (recommended laser parameters for these indications are 4–7 W laser power and continuous wave as temporal mode) or vestibuloplastic (up to 15 W; temporal mode: modulated, 20 Hz) [4,5]. Further applications are the bacteria reduction in root canals or periodontal pockets as part of an endodontic or periodontal treatment. For these applications average laser power is 2–3 W, usually modulated up to 50 Hz) [6,7]. Another important application is dental bleaching associated with fluoride [8].

The used laser wavelengths are in the near infrared spectrum, 800–980 nm, irrespective of the dental application. The laser light is coupled into a delivery fiber (typical fiber diameter: 200–400 μm) and either the delivery fiber itself or the laser light is coupled into fiber tips, both typically are used in contact mode. To prevent deep light penetration into the tissue with uncontrolled heating the fibers are typically initiated before use. For initiation black articulating paper is used for instance. After treatment the fibers are cut or the fiber tips are exchanged. Due to intraoral lack of space the geometry (length, diameter, (variable) bending of distal fiber end) of the used handpieces has an essential influence on the handling and on the accessibility, especially in the rear intraoral region.

The aim of this study is the investigation and optimization of fiber tip geometries for improved cutting quality in oral surgery. This is part of a novel concept of a handpiece with exchangeable fiber tips for a dental diode laser allowing not only the use of new sterile fiber tips for each patient but also tips with special and optimized geometries for various applications. This could include side or circular firing fibers for root canal or periodontal treatment.

The objective of the actual study is to increase the cutting quality and depth by increasing the laser irradiance at the tip compared to the bare fiber.

At first various potentially suitable tip geometries were investigated by optical ray tracing. Then the most promising tips were realized and tested on porcine mucosa with regard to the resulting cutting efficacy, quality, and thermal effects.

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MATERIALS AND METHODS

Part 1: Simulation and Optimization of Several Fiber Tip Geometries

The simulations of the fiber tip geometries were done using commercial optical raytracing software for design of optical systems (Zemax, version November 10, 2008, Radiant Zemax Europe Ltd., Stamsted, United Kingdom). The software was used in the "non-sequential ray tracing mode," which means that the optical elements such as planes can be hit several times by the same optical ray. The investigated geometries were:

- bare fiber (Ø = 200 μm) with plane output end (as benchmark for comparison),
- hemispherical tip end (A),
- conically shaped fiber tip with flat end (taper; B),
- taper with hemispherical end (C), and
- sphere (not really a fiber tip; D): the optical design corresponds to a telecentric set-up, consisting of a plane convex lens and a sphere, which images the output end of the delivery fiber onto the tissue surface.

Figure 1 shows, as an example, the simulation layout of the fiber tip with plane output end. To simulate the typical laser beam distribution at the output end of the delivery fiber (silica, diameter 200 μm, NA = 0.2) the rays of a homogeneous circular light source (diameter 1 mm, beam divergence NA = 0.4) hit a circular aperture (inner diameter 200 μm) and afterwards the fiber input end. The distance between light source and aperture was chosen in a way that the NA of both in coupled light and fiber output end correspond to the typical NA = 0.22 of a silica fiber. The delivery fiber output end was imaged (image scale 1:1) onto the fiber tip input end by a matched lens for coupling the laser light (power 1 W) into the tip after deflection by a mirror. The mirror allowed the design of an angled handpiece and has no influence on the beam distribution inside and after the fiber tip.

The intensity distribution within and behind the output plane of the fiber tip (Fig. 1, lower left) and the intensity profile as well as the maximum intensity behind the fiber tip (Fig. 1, lower right) were determined by a detector, which could be positioned in different distances z to the output end (resolution $\Delta z = 10$ μm). Based upon these simulations the transmission efficiency of the fiber tip as well as the focus position, diameter, and length (FWHM) were calculated. Also the beam diameter $\varnothing_{0.1\times \mathrm{IMax}}$ (diameter, where the intensity has decreased to 0.1^* maximum intensity) in a distance $z = 2$ mm behind the focus was determined and the divergency α was calculated:

$$\alpha = \arctan\left(\frac{\varnothing}{2z}\right)$$

To get a measure of the focusing property of each design, the "focusing factor" F_o was calculated:

$$F_o = \frac{\text{maximum intensity of actual fiber tip}}{\text{maximum intensity of bare fiber}}$$

In a second step several geometries were varied by maximizing the focusing factor F_o. The following parameters were varied for optimization:

- hemispherical tip end A: radius of the hemisphere,

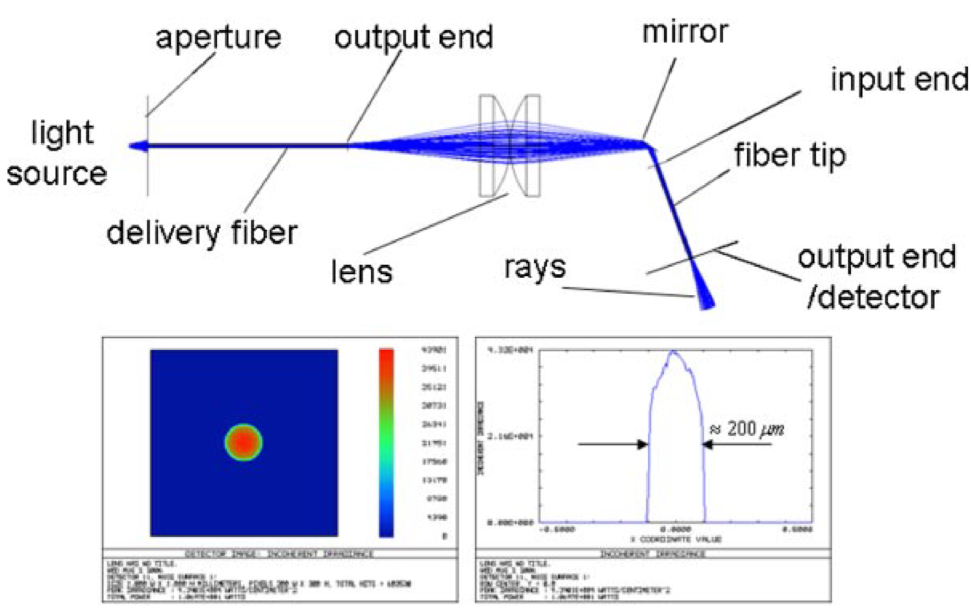

Fig. 1. **Above**: simulation layout of the fiber tip with plane output end (the blue lines are the simulated light rays); the output plane of the handpiece corresponds to the output end of the fiber tip; **lower left**: simulated laser intensity distribution in the output end; **lower right**: intensity profile.

- conically shaped fiber tip with flat end (taper) B: total length and diameter of the cylindrical part; length, cone angle and output end diameter of the conically shaped part,
- taper with hemispherical end C: total length and diameter of the cylindrical part; length, cone angle and output end diameter of the conically shaped part, and
- sphere D: focal length of the lens; diameter/focal length of the sphere.

In order to consider both situations of no or slight contact of the tip to the tissue, as well as close contact, air and water were chosen as surrounding medium.

Part 2: Development, Realization, and Test of the Most Promising Tips

Because the hemispherical tips A and C revealed severe disadvantages (see Results and Discussion Section) and promising simulation results were obtained for the sphere D and for the taper B it was decided to realize and test one laboratory hand piece with a sphere as contact tip and another one laser hand piece with a conically shaped tip.

In order to provide conditions close to clinical practice, for all investigations a commercial dental diode laser system (wavelength 970 nm, Sirona Dental Systems GmbH, Bensheim, Germany) with 200 and 300 μm silica delivery fibers was used as light source. Also fibers and tips were "initiated" by usage on soft tissue for about 10 seconds before performing the investigations.

Investigations with spheres as contact tip. Figure 2 shows the developed and produced laboratory handpiece with plane convex lens (sapphire, $f = 3$ mm, Victor Kyburz AG, Safnern, Switzerland) and a sphere (BK7 glass, diameter 1 mm, Kugel Pompel GmbH&CoKG, Vienna, Austria). The sphere was pressed into a stainless steal tube.

The cutting tests were performed on fresh porcine skin (provided on the same day from slaughterhouse), which was fixed on a cork scaler. Because of the observed obvious poor cut quality the cutting tests were not repeated on porcine oral mucosa, which was used for the subsequent tests. The laboratory handpiece was manually moved in contact to the tissue surface, with a velocity of about 2 mm/second. For comparison cuts were made with the standard laser handpiece and 300 μm bare fiber (manufacture recommendation, cutted output end). Laser parameters were: 2, 3, 4, and 6 W (display values); continuous wave (cw).

The cuts were photographed by the help of a dental operation microscope with an implemented digital camera (OPMI Pico;, Carl-Zeiss Meditec AG, Jena, Germany; magnification 2.5×). Because of an obvious poor cut quality no measurement of cut geometry and no histological sections were performed. Before and after treatment the laser power was measured at the delivery fiber output end and behind the sphere using a power meter (Laserstar,; Thermal Head 3A; Ophir Optronics Solutions Ltd., Jerusalem, Israel). Then the handpiece transmission and the loss of transmission due to contamination of the sphere was calculated by dividing both values.

Investigations with conically shaped tips (taper). The tapers were produced by grinding and polishing (diamond pads; 3M Deutschland GmbH, Neuss, Germany) 800 μm cylindrical sapphire tips (Victor Kyburz AG, Safnern, Switzerland). The taper corresponds to the simulation results for the optimized geometry (conical length 10 mm, cone angle 5°, input diameter 1 mm, output diameter 200 μm). To test the tapers under realistic conditions, a handpiece was developed, which widely conforms to the specifications for clinical use. The laser light of the diode laser system was directly coupled from the delivery fiber (the fiber diameter of 200 μm corresponds to the simulation results) into the exchangeable fiber tip without additional optics. Figure 3 shows the grinded and polished taper, which is fixed in a stainless steal tube and fitted into the new handpiece.

In order to standardize the cutting tests with the new handpiece and sapphire taper, fresh porcine oral mucosa (see above) was fixed on a computer-controlled motion unit and the laser hand piece was manually positioned on the tissue surface in contact. To compensate for tremor, forearm, and hand were leaned on a table. By this, the tip

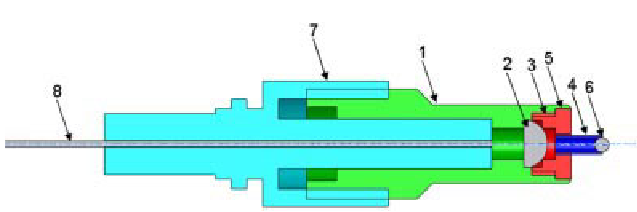

Fig. 2. First laboratory laser handpiece with a sphere as contact "fiber tip" (1: main body; 2: lens, $f = 3$ mm; 3: tip holder; 4: tube; 5: fit; 6: sphere (Ø = 1 mm); 7: SMA connector; 8: delivery fiber).

Fig. 3. New laser handpiece with conically shaped fiber tip, adapted to the diode laser (the radially out coupled light of the red pilot beam is visible); small figure: Grinded and polished sapphire taper ($l = 12$ mm, $Ø_{in} = 800$ μm, $Ø_{out} = 200$ μm).

could be flexibly adjusted to the tissue surface, while the cutting speed was precisely controlled by the sample movement. For comparison cuts were made with a standard laser handpiece and 300 μm bare fiber with polished and cut ends. The polishing of the fiber ends was done manually with SiC grinding paper (grid 1200, 2500, 400; Buehler, Duesseldorf, Germany). In total four different cuts with speed of 3 mm/second were made (preliminary tests have shown that at 3 mm/second, and at the used laser parameters a readily evaluatable depth of cut is achieved). The used laser parameters, 5 W/continuous and 7 W/1 kHz modulated, correspond to typical recommendations in the literature for oral surgery [4,5]. During the cutting tests the handpiece transmission before and after treatment was determined (see above).

The cuts were photographed by the help of the dental operation microscope. Afterwards histological sections were prepared using a cryo-microtome. The histological sections were stained with Hematoxylin–Eosin (HE) and then microscopically analyzed (Axiophot; Carl-Zeiss AG; adapted digital camera: ProGres C12, Jenoptik AG, Jena, Germany). HE stain is well-established method in histology and is used to visualize tissue morphology. It stains cell nuclei in blue while cytoplasm and other structures appear in red [9].

Although the achievable quality of the sections is demonstrated well by the above tests, it is difficult to obtain reproducible quantitative results, when the tissue is moved and the handpiece is fixed. In contrast the manual control leeds to a minor cut quality (the cuts appear crooked) but allows the compensation for variations of tissue height and for fiber tissue adhesion. For this reason an additional experimental set-up was performed with tissue fixed on a cork scaler and manually guided handpiece. In total four different cuts with a predetermined length of 10 mm and treatment time of 15 seconds were made (in contact mode, laser parameters: 5 W/continuous and 7 W/1 kHz). The analysis of the cuts corresponds to the prior described procedure. For each cut three histological sections at about 1 mm distance were made and the cut depths were measured by the light microscope using the software implemented in the digital camera. Statistics *t*-test, significances ($P < 0.05$, $P < 0.1$)) were performed (Excel 2000, Microsoft Deutschland GmbH, Unterschleißheim, Germany).

RESULTS

Part 1: Simulation and Optimization of Several Fiber Tip Geometries

In Figure 4 the simulation results for the investigated tip designs are depicted (surrounding medium: air). Table 1 contains the transmission, the focus position and diameter, the divergency, and the focusing factor of each design.

The intensity distribution within the output plane and therewith the focusing factor of the tip strongly depend on the tip geometry. In sum following results were obtained:

- The plane tip end shows a homogeneous intensity distribution across the output end. The divergence of the emitted light is smaller compared to the other simulated geometries.
- The hemispherical tip end acts as a convex lens, which focuses the out-coupled laser light at a distance of 120 μm behind the output end (focus diameter 54 μm). 47.7% of the light power is totally reflected at the hemisphere and leaves the input end towards the deflection mirror. This is the reason for the reduced transmission (49.6%) compared to the bare fiber.
- The conic shape of the taper also results in focusing of the laser light (focusing factor $F_o = 4.53$). The divergency of the emitted light (29.1°) is the largest.
- The combination of taper and hemispherical output surface leads to nearly the same focusing effect as the standard fiber with hemispherical end. Also here a part of the light is totally reflected at the hemisphere and leads to a reduced transmission compared to the taper with plane end.
- The smallest focus diameter (31.4 μm) and best focusing factor ($F_o = 16.2$) was achieved by imaging the fiber end onto the tissue surface using a sphere (BK7, diameter 1 mm) in combination with a second lens ($f = 3$ mm).
- When the medium surrounding tip is water, the transmission increases in all cases, but especially for the geometries with hemispherical tip end. For these geometries the focusing factor decreases.

Part 2: Development, Realization, and Test of the Most Promising Tips

Investigations with spheres as contact tip. In Figure 5 microscopic photographs of the cuts are depicted, which were made with the sphere as contact tip and with bare fiber at various laser powers. Using the bare fiber adhesion and therewith hooking of the fiber is observed, which always leads to irregular cuts. The cuts made with the sphere are more homogeneous and no hooking is observed. However, they show an obvious wider zone of ablation and thermal injury. During cutting both the bare fiber end and the sphere become dirty and hot. While usage, the measured light transmission of the handpiece with sphere decreased from 40% to 18%, and the sphere had obvious disruptions.

Investigations with conically shaped tips (taper). The measured transmission of the developed handpiece with taper before cutting is 80% and matches the simulation value (79%). After treatment the measured transmissions are 56% for the taper, 46% for the bare fiber with cut ends, and 60% with polished ends.

The cuts made with computer controlled movement of the tissue (3 mm/second) and manually positioned hand piece appear more uniform without hooking of the sapphire taper compared to the bare fiber (see Fig. 6).

To get a more realistic comparison of the bare fibers and the conically shaped tip the following results are

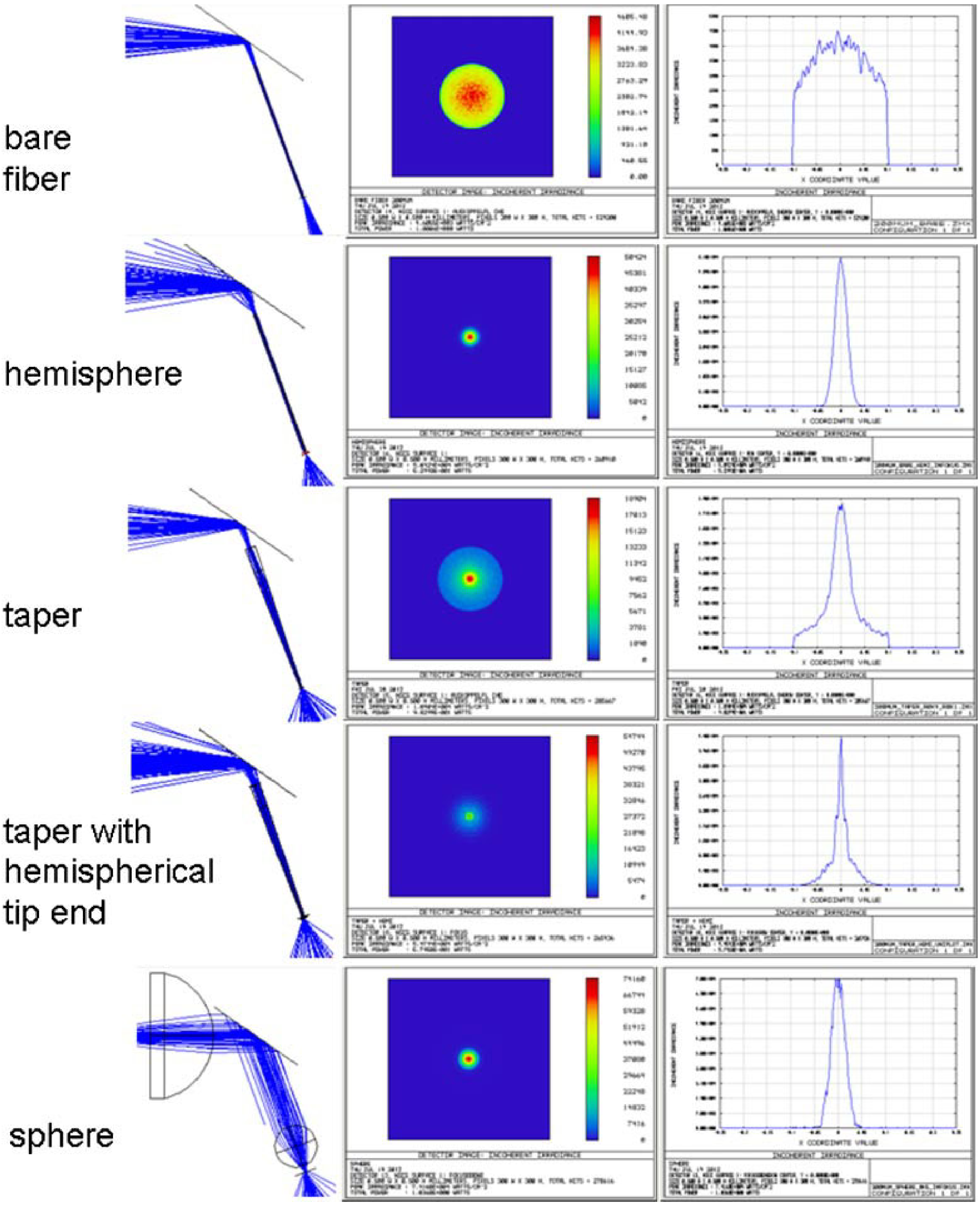

Fig. 4. Optical ray-tracing of the investigated tips for oral laser surgery (surrounding medium: air): from top: plane tip end, hemispherical tip end, taper, taper with hemispherical tip end, sphere in combination with lens (f = 3 mm); from left: layout, laser light intensity distribution, and intensity profile within the focus plane.

TABLE 1. Transmission, Focus Position, Diameter, and Length, Divergency, and Focusing Factor of Simulated Tip Designs

	Transmission	Focus position (μm)	Focus diameter (μm)	Divergency (°)	Max. intensity (W/m^2)	Focusing factor
In air						
Bare fiber	0.935	0	200	8.53	0.43	1
Hemispherical tip end (A)	0.496	120	54	22.6	5.0	11.6
Taper with flat end (B)	0.917	0	39.5	29.1	1.95	4.53
Taper with hemispherical end (C)	0.536	40	36.5	22.5	4.44	10.3
Sphere (D)	0.97	158	31.4	25.64	6.97	16.2
In water						
Bare fiber	0.967	0	200	6.56	0.434	1
Hemispherical tip end (A)	0.873	473	74.5	22.6	1.478	3.41
Taper with flat end (B)	0.94	0	40.5	29.1	1.80	2.30
Taper with hemispherical end (C)	0.796	55	12.6	22.5	5.12	11.80
Sphere (D)	1.1	220	35.5	25.64	5.72	13.18

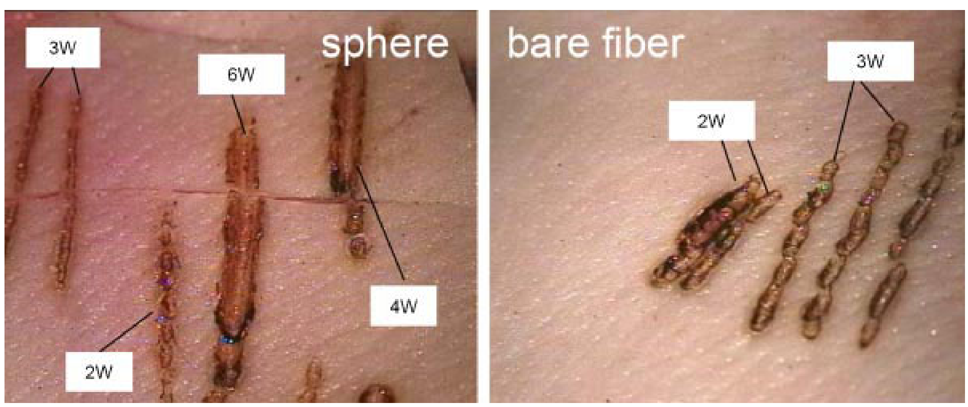

Fig. 5. Cuts on porcine skin with various laser powers; **left**: laboratory laser handpiece with 1 mm glass sphere (a cut with a scalpel perpendicular to the laser cuts was made to see the resulting cut geometry and coagulation zone); **right**: bare fiber.

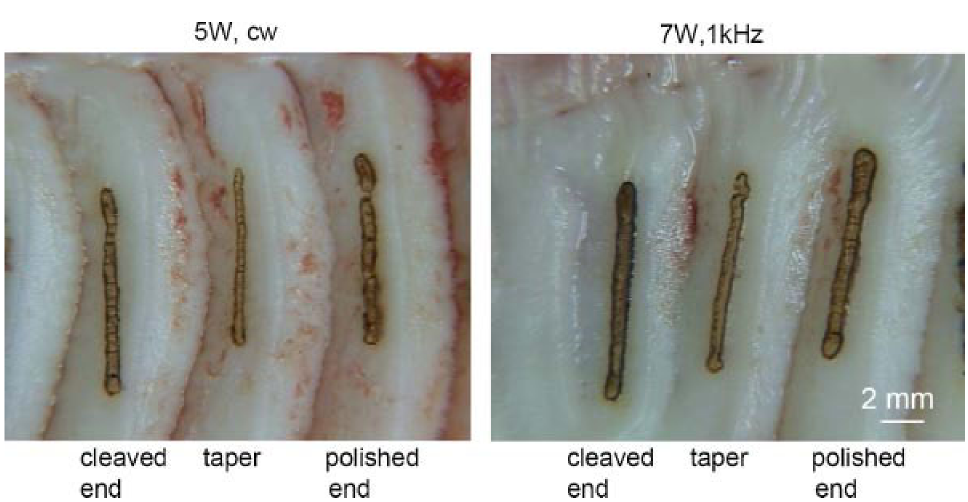

Fig. 6. Test cuts on porcine oral mucosa, made with the taper and with 300 μm bare fiber, cleaved, or polished end (movement velocity of the tissue $v = 3$ mm/second).

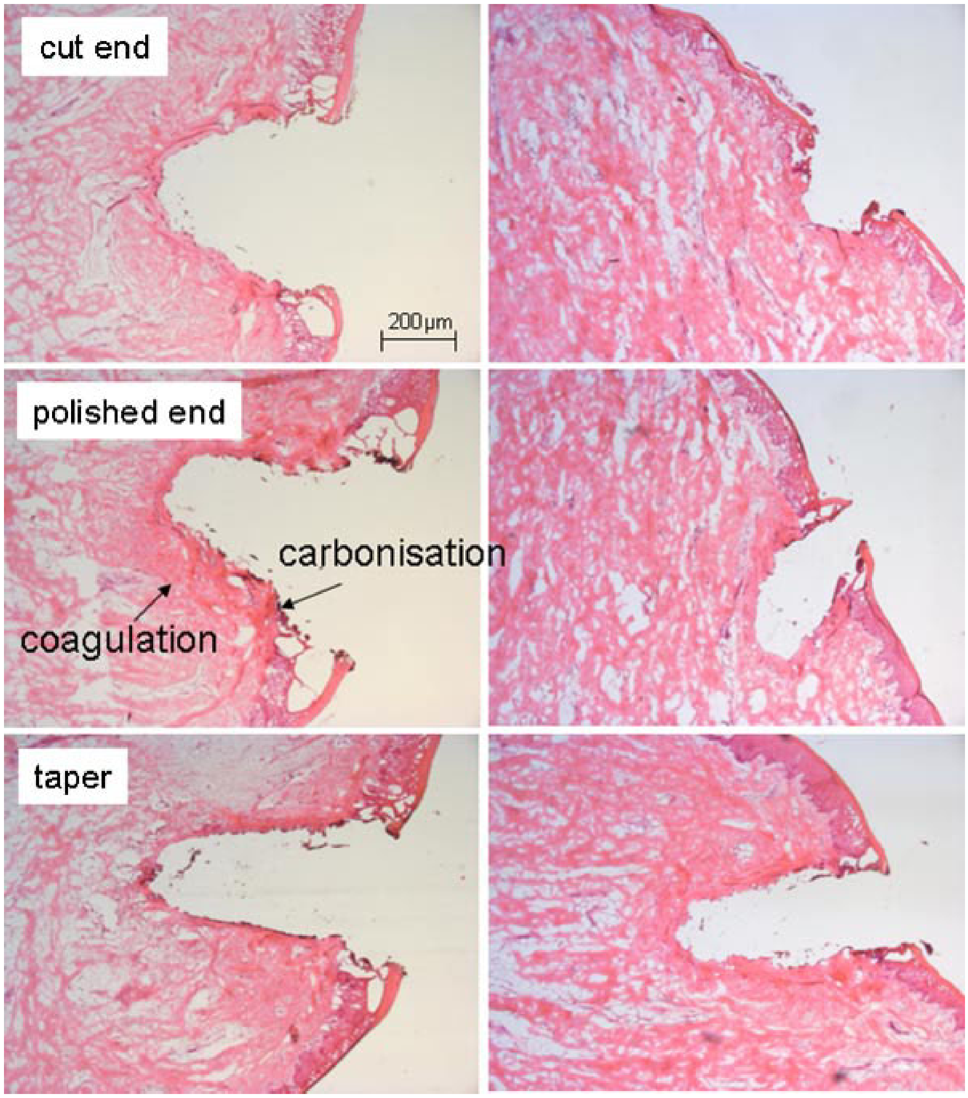

Fig. 7. Photomicrographs of cryo-microtome histological sections (HE-staining) of the test cuts, made with the taper and with 300 µm bare fiber; the laser hand piece was manually moved for a duration of 15 seconds per cut; **left**: 5 W, cw; **right**: 7 W, 1 kHz.

obtained by manually guided movement of the hand piece and fixed tissue.

Now the hooking of the bare fibers is less pronounced, which is caused by the better manual handpiece control. The handling and cut quality using the taper are still excellent. In Figure 7 photomicrographs of the histological sections (cuts made with 5 W, cw and 7 W, 1 kHz modulated) are depicted. The mean values and standard deviations of the resulted cut depths, measured from the histological sections, are diagrammed in Figure 8. The deepest cuts were achieved with the sapphire taper (5 W/cw: 1,673 ± 368 µm; 7 W/mod.: 1,124 ± 223 µm), followed by the polished bare fiber (5 W/cw: 1,083 ± 413 µm; 7 W/mod.: 832 ± 209 µm), and the cut bare fiber (5 W/cw: 711 ± 268 µm; 7 W/mod.: 580 ± 294 µm). The differences are significant both for comparison of taper and cleaved fiber ($P < 0.05$) and for comparison of taper and polished fiber ($P < 0.1$). As also seen on the photomicrographs of the histological sections in Figure 7 the widths of the thermal damage zones of the cuts (about 100 to 400 µm) are similar for the taper and fibers.

DISCUSSION

Part 1: Simulation and Optimization of Several Fiber Tip Geometries

The optical simulations afford the design and optimization of various tip geometries to attain the aimed focusing

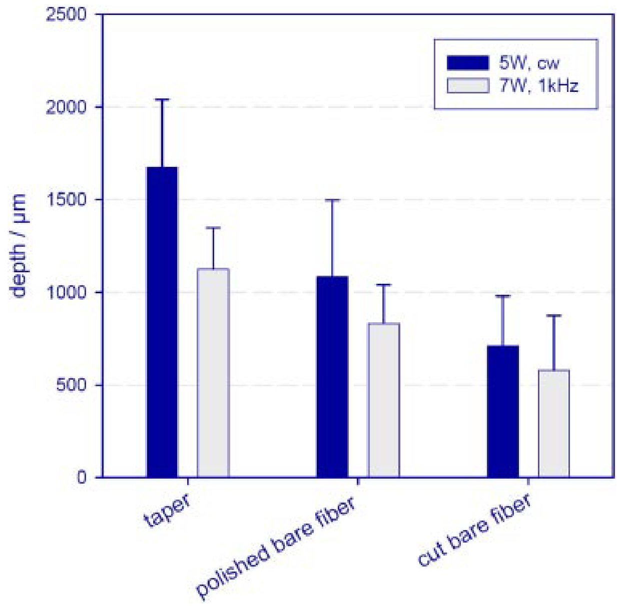

Fig. 8. Mean values and standard deviations of the resulted cut depths, measured from histological sections.

effect. The glass sphere is not only a easily producible optical element but also shows a strong focusing factor in air ($F_o = 16.2$). Beside the spheres also tips with hemispherical end show a high focusing effect, which is caused by the refraction of the output surface. A great advantage of the hemispherical end is the simple potential realization by melting a plane tip end [10]. However in tissue contact the focusing effect is reduced, caused by the higher refraction index of the tissue compared to air. The resulting focusing factor decreases from $F_o = 11.6$ to $F_o = 3.41$, when water is surrounding the tip end. Verdaasdonk and Borst [10] used optical ray tracing to design spherical tip ends for laser angioplasty and they compared the computations with measured intensity distributions. They also showed an increase of irradiance compared to plane fiber end, which is diminished in water caused by the higher refraction index of water. Another important aspect is light scattering by the tissue. It has to be expected that for all tip geometries having a focus in front of the tip, the scattering will reduce the focusing effect. Also with increasing contamination of these tips the absorption and scattering of the out coupled laser light may diminish focusing.

In case of the conically shaped tips with flat ends the focusing effect is caused not by refraction but by total reflection of rays by the cone surface. Thereby the focusing is less influenced by the contamination of the output surface and the degree of tissue contact. The simulations with water as surrounding medium show no reduction of total reflection by the cone surface and therefore no reduced transmission. Solely the contamination of the cone surface may lead to an attenuated total reflection and with this to an energy loss. As an additional advantage the large input end of the taper compared to the output end of the delivery fiber simplifies the laser light coupling into the taper.

In a similar way Melnik et al. [11] used optical ray tracing to calculate the light distributions of aspherically modified tips (parabolic, hyperbolic, etc.). They showed, qualitatively in agreement to our results, a higher focusing effect of aspherical relative to plane and hemispherical tips. With aspherical tips it is possible to increase the irradiance of tissue in water by about six to nine times compared to plane fiber end. Aside from the complex geometry we suppose that also here the contamination of the fiber end by ablated tissue will destroy the focusing effect.

Part 2: Development, Realization, and Test of the Most Promising Tips

The huge broadness of the cuts made with a sphere as contact tip can be explained by the large size of the sphere (1 mm), which results in a large hot contact area with a corresponding wide zone of ablation and thermal injury. As discussed prior, it has to expected that the scattering of the tissue will derogate the focusing effect. So, in conclusion, the advantage of the sphere as an easily producible optical element with a high focusing effect (in air) is foiled by the observed poor cutting quality.

The *in vitro* tests with the conically shaped tips show that the specially designed tip geometry leads not only to better handling and cut quality but also to an increase of efficacy, confirmed by the increase of the cut depth with the sapphire taper, compared to the cut bare fiber. In our understanding the improved handling and cut quality is not only based by the focusing effect but also by the enhanced mechanical properties of the taper. The conical shape leads to a more rigid tip which prevents hooking, which is present while cutting with the bare fiber. The reduced cut depths in case of the modulated operation mode (1 kHz) can be explained by the less mean laser power of 7 W/2 = 3.5 W compared to 5 W, cw.

A comparison of the measured cut geometries and thermal damage zones with other published results of *in vitro* tests is impossible because of the different test parameters (tissue, fiber diameter, laser power, velocity, and manner of movement) [12–15].

CONCLUSION

In conclusion the investigations show that the cutting efficacy and quality of the diode laser on soft tissue depends on the used fiber tip geometry. Best results were achieved by using conically shaped fiber tips, for which improved handling, better cutting quality, and a significant increase of cut depth can be observed. The investigations were performed using a wavelength of 970 nm and 200 μm delivery fiber (NA = 0.2). We would expect similar results for other lasers in the NIR spectral region, which have the similar absorption in tissue. The results can presumably not transferred to lasers providing high absorption in tissue, for example, Er:YAG-laser or CO_2-

laser. Especially for the so called thermomechanical ablation mechanism of the Er:YAG-laser other influences of the treatment parameters (e.g., beam diameter, fiber to tissue distance) have to be expected.

In general it has to be noticed that all results are derived from *in vitro* experiments, which might be different from the results of *in vivo* or clinical tests (i.e., influence of absent blood). From this point of view the clinical test of the novel fiber tips should be the next step.

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