Modified fiber tips: optical and thermal characteristics (Invited Paper)

Verdaasdonck, Rudolf, Borst, Cornelius

Rudolf M. Verdaasdonck, Cornelius Borst M.D., "Modified fiber tips: optical and thermal characteristics (Invited Paper)," Proc. SPIE 1649, Optical Fibers in Medicine VII, (1 August 1992); doi: 10.1117/12.60235

SPIE.

Event: OE/LASE '92, 1992, Los Angeles, CA, United States

MODIFIED FIBER TIPS: OPTICAL AND THERMAL CHARACTERISTICS

Rudolf M Verdaasdonk and Cornelius Borst
Medical Laser Center and Heart Lung Institute, University Hospital Utrecht,
3508 GA Utrecht, The Netherlands

I. INTRODUCTION

The strongly forward directed and torch like beam profile and spatial irradiance distribution of a laser beam delivered by a fiber may be changed by modifying the fiber tip optically. A selection of modified fiber tips is presented in Figure 1.

To increase the fluence rate, the fiber tip may be reshaped into a taper [1] (Fig.1e) or a ball [2] (Fig.1c). An additional optical shield on the modified fiber tip (Fig.1g) will also influence the optical behavior, especially in liquid or tissue [3]. In combination with this shield, the beam can be directed at a right angle out of the fiber (Fig.1h). Alternatively, a tapered rod (Fig.1d) or a lens (Fig.1b) may be positioned in front of the fiber tip [4,5]. These components may also protect the fiber tip from ablated debris. Spherical probes are used for contact coagulation and contact ablation of tissue [4,6] whereas tapered configurations are used as laser scalpel to cut tissue effectively with minimal bleeding due to photocoagulation [5,7,8].

To create an isotropic irradiating probe as used in photodynamic therapy, a sphere of highly scattering material is positioned on the fiber tip (Fig.1i) [9]. A peculiar concept is the use of an opaque cover such as a metal cap [10] that converts laser energy into heat producing a 'hot tip' (Fig.1j).

Typically the modification to the fiber end is made from the fiber material itself (fused silica) or a separate probe of silica or sapphire is added which introduces an interface between the fiber and the tip.

The flexibility of the delivery systems is increased by replacing a single fiber by a bundle of smaller fibers covering the same surface area (Fig.1f). This allows access to small tortuous lumina and has potential for guidance and selective ablation [11].

This paper briefly reviews the optical and thermal properties of several modified fiber tips used in conjunction with various lasers.

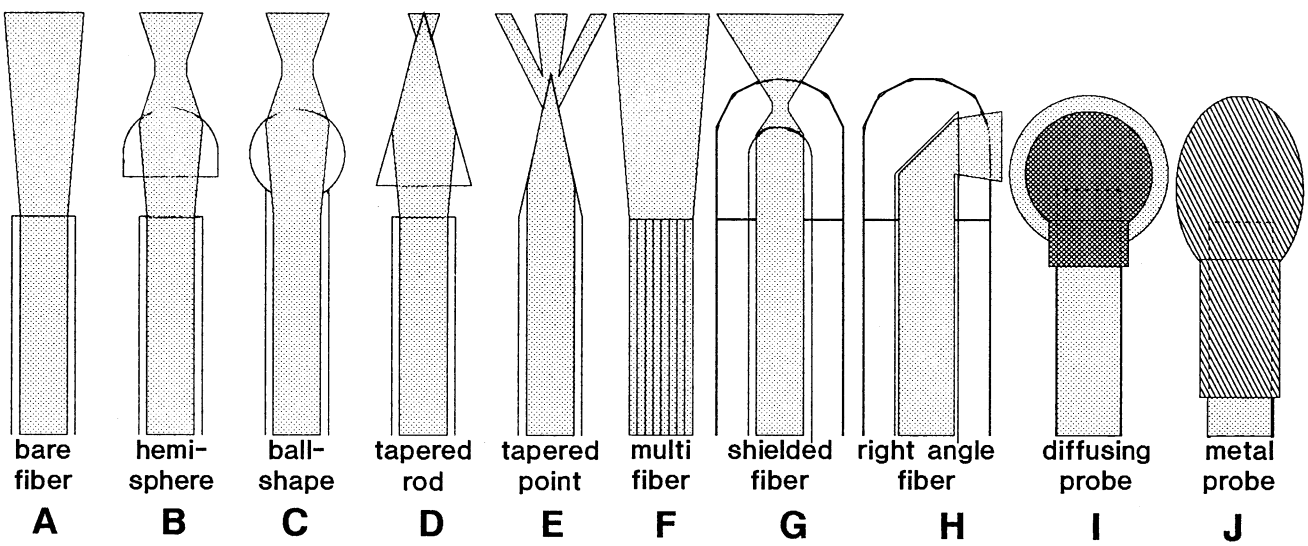

Figure 1. Schematic graph of various modifications of the fiber tip

0-8194-0795-X/92/$4.00

2. MODIFIED FIBER TIPS

2.1 Spherical probes

The modification of the bare fiber tip has mainly been encouraged by the use of cw lasers for recanalization of occluded arteries. The two most prominent fiber delivery systems emanating from this field are the metal laser probe [12] (Fig.1j) and the transparant hemispherical contact probe [13,14] (Fig.1b). The blunt, large diameter probes carry a reduced risk of perforation and create a larger lumen compared to bare fiber delivery of laser energy [6,15]. The recanalization mechanism of the probes was originally attributed to thermal ablation of the obstructive tissue [15,16]. Their physical properties, however, were ill defined at the time of their clinical introduction. As a result, a large number of patients have been treated successfully by these probes without fully understanding their mechanism of action [14,17].

2.2 Shielded fibers

In order to irradiate a larger surface area (Fig.1g) or to direct the laser beam at a right angle with the axis of the lumen (Fig.1h), optical shields were introduced. Making use of the maximum optical effect, the air interface between modified fiber tip and shield provides a large diverging beam or reflection surface to meet these goals. Examples of these probes are the Lastac system [3], the corolla probe [18] and the right angle fiber [19].

2.3 Metal probes

Metal laser probes were introduced by Trimedyne (Tucson, Ca) in 1982 [10]. The probes consist of a metal cap connected on top of a bare fiber end. The cap absorps all the light coming out of the fiber and converts it into heat (Fig.1j). Various types have been developed with diameters from 1 to 4 mm with shapes resembling spheres to footballs. Besides angioplasty, the probes are used in various surgical areas for tissue coagulation and ablation. In some designs, part of the laser light is able to exit through a small window in the tip of the probe.

2.4 Multi fibers

Delivery systems became more flexible by using bundles of very small fibers instead of a rigid single fiber [20] (Fig.1f). The fiber bundle can almost cover the same surface area except for the dead space between the fibers. The combination with an optical shield can be used to fill in this dead space [11]. The fibers can be positioned at preferential positions with respect to the tip symmetry, e.g. to attack asymmetric stenoses or to allow a central lumen for a guide wire. Besides the better guiding capabilities, it also allows selective ablation by probing and ablating tissue through individual fibers as performed by the MIT and Smart laser systems [21]. The change from the hot thermal cw lasers to the mildly hot pulsed lasers makes the use of delicate multifiber systems more practical.

2.5 Tapered fibers

Besides focusing a beam by means of a lens, high fluence rates of laser light are obtained by guiding the beam through the tapered end of a fiber [1] (Fig.1e) or a tapered rod [7,8] (Fig.1d). As the cross-sectional area of the rod decreases, the fluence rate increases until the beam starts to refract out of the taper with usually a large divergence angle. In contact with tissue, the high power density of the light at the point of the tapered end initiates tissue ablation[4]. Because of the large divergence angle of the exit beam, the irradiance decreases rapidly distal from the tip minimizing injury to adjacent tissue [8].

The very tip of the taper can be flat or spherical resulting in different beam profiles.

3. OPTICAL PROPERTIES OF MODIFIED FIBER TIPS

3.1 Physical Properties

The optical and thermal behavior of modified fiber tips will primarily be governed by the optical an thermal properties of the materials of the probes and of the environment in which they are used. In table 1 an overview is presented of the properties of importance. The refractive index determines the magnitude of optical interaction. A transition with a large change in refractive index can strongly affect the direction of a beam, but it also introduces undesirable reflections and power loss. Heat may be produced either by absorption of light by the probe itself or by tissue in contact with the probe. A high heat conduction contributes to control excessive temperature rises to protect the probe from damage. It is attractive to make probes of sapphire due to the high thermal conductivity combined with a high melting temperature. Also the hardness of sapphire is high but it is very brittle. The characteristics of the laser source like wavelength and pulse duration, and the geometry of the fiber tip or probe in front of the fiber, will contribute to the performance of the probe.

Table 1. Physical properties of sapphire, silica and water

	sapphire	*silica*	*water*	
Refractive index (1064 nm)	1.75	1.45	1.33	
Density	3.98	2.20	1.00	10^6 g m^{-3}
Heat capacity	0.75	0.75	4.18	J K^{-1} g^{-1}
Thermal conductivity	25.9	1.38	0.60	J K^{-1} m^{-1} s^{-1}
Max. temperature	1800	950	(100)	°C

3.2 Optical characteristics

The optical behavior of the modified fiber tip depends on the fiber diameter, the irradiance distribution of the beam from the fiber, the geometry of the fiber tip or probe in front of the fiber, the refractive index of the probe and the refractive index of the medium. It is deceptive to use standard geometrical optics theory to predict the beam profiles for modified fiber tips [22]. For instance, from their optical characteristics in air, (hemi)spherical probes were originally assumed to perform better and safer due to an increase in the fluence rate in front of the probe combined with a rapid decrease in fluence rate away from the probe [6]. However, in physiological media or in contact with tissue, the focusing by these probes is less pronounced or even absent due to the limited transition in the index of refraction combined with the nature of a beam from a fiber as well as the scattering of light by tissue [23,24]. The irradiance distribution of the beam from a fiber has properties that are not consistent with some of the prerequisites of paraxial optics. The beam diverges and the irradiance has both a spatial and an angular distribution [24].

3.3 Ray tracing

Ray tracing is a better method to evaluate the optical behavior of the modified fiber tips [24,25]. The optical characteristics of modified fiber tips were predicted by using a ray-tracing program that has been described before [24]. The program traced a beam from a fiber which had an irradiance distribution that was uniform in the near field and gaussian in the far field of the fiber. Graphic representations of the beam profile were produced in and distal from the probe. Calculated irradiance distributions at successive distances from the probe were combined to an iso-irradiance map.

3.3.1 Spherical probes

The optical behavior of spherical probes made of silica and sapphire was evaluated in air and water using the program. The geometry of the probes was defined by the radius of the sphere R_s in relation to the radius of the fiber R_f. For both ball shaped fibers and hemispherical probes the position of the maximum irradiance in front of the probe in water was closer to the probe than the calculated paraxial focal point. In paraxial theory the position of the focal point (which is also the point of highest irradiance) is determined by $F = R_s\, n_1/(n_2-n_1)$

where F is the distance of the distal focal point from the surface of the probe with refractive index n_2 in the medium with the refractive index n_1 and R_s is the radius of the probe sphere. In water, spherical probes of silica loose their focusing effect for a sphere/fiber radius ratio $R_s/R_f>2$ as illustrated in figure 2 (top). For spherical probes of sapphire, the focusing of the beam in water can still be appreciated (Fig.2, bottom). The discrepancy is illustrated in figure 3 and can be explained by the off-axis position of multiple focal points from the fiber beam which consists of a series of parallel beams at a small angle with the optical axis. Due to the off-axis location, the position of the waist and the position of the highest irradiance is closer to the probe surface than the paraxial focal point. As a result, the focusing effect observed in air almost vanishes in water, in contrast to what is expected from paraxial theory.

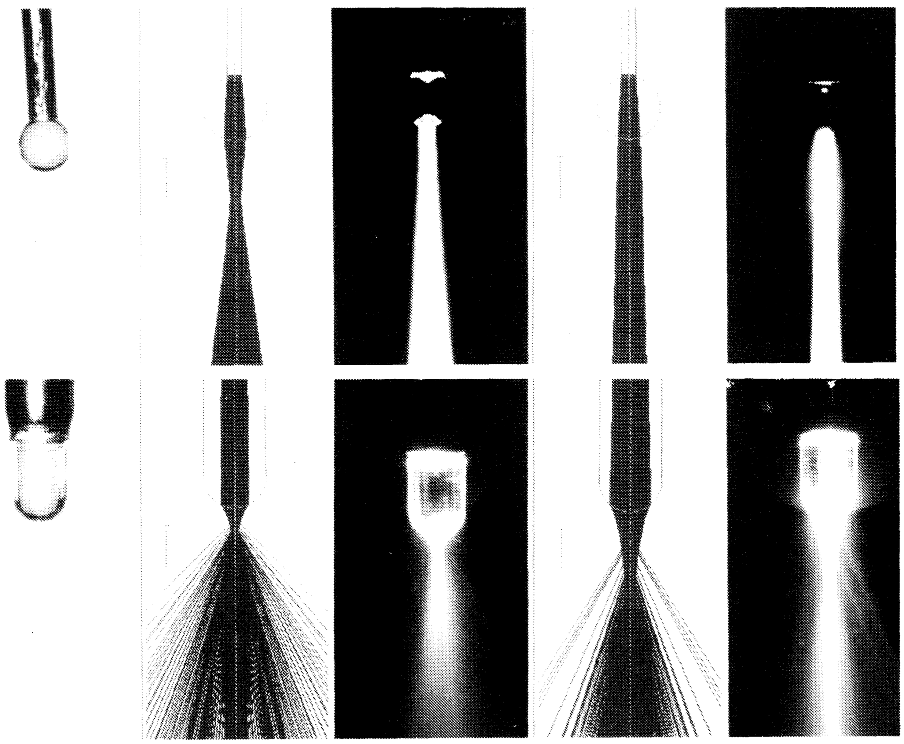

Figure 2.
Beam profiles. From left to right: probe shape, calculated beam in air, observed beam in air, calculated beam in water, observed beam in water.
Top: 1.5 mm silica ball tip.
Bottom: 1.5 mm sapphire hemisphere

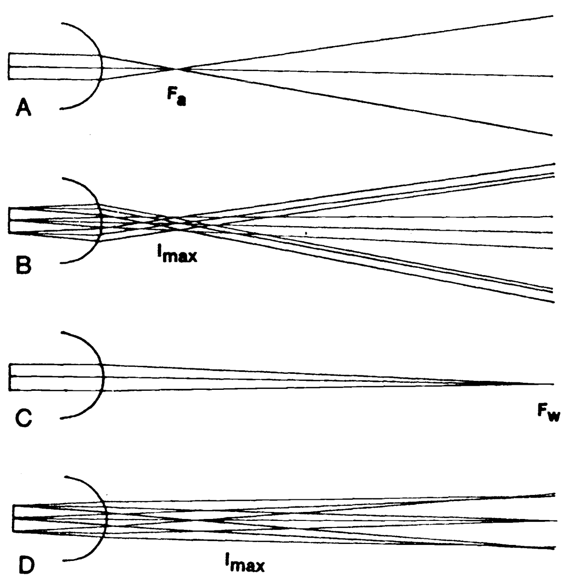

Figure 3.
The discrepancy between paraxial theory and ray-tracing in predicting the behavior of beams propagating from fibers. In air, both the paraxial beam (A) and the fiber beam (B) are focused at F_a. In water, on the other hand, the paraxial beam is still focused at F_w(C) but the fiber beam diverges (D). Consequently, the position of the highest local irradiance, I_{max}, does not coincide with the focal point F_w.

Care must be taken in case of ray-tracing calculations to use the correct index of refraction of a tip. For example, the transparant contact probe (Surgical Laser Technologies, Malvern, PA, USA) was originally manufactured from artificial sapphire [4,5,23] and referred to as "sapphire contact probes". Later on it was manufactured from silica but was still referred to as "sapphire" [24].

3.3.2 Shielded tips

The focusing power of a spherical tip in air is greatly reduced when the probe is placed in water or in contact with tissue. By using an optical shield in front of the tip which provides an air interface, the focusing properties can be preserved in a water environment. This configuration enables manipulation of the spot size, the power density distribution and the divergence angle. The shield concept is incorporated in the Lastac^{R} device [3] and in the modified ball tip (Advanced Cardiovascular Systems) resulting in a wide aperture tip (40^{o} divergence angle in water), which has a beam spot equal to the front of the shield [26].

If an optical quartz shield is used with air or vacuum between a bare fiber tip and tissue, the spot size will increase at the expense of power density. This modification has been implemented in some multifiber catheters to abolish dead space by overlapping beam spots which cover the entire front of the shield [11].

3.3.3 Multifibers

The optical behavior of a multifiber itself can be compared with a bare fiber tip. The beam of a small fiber in the fiber bundel can be considered analogous to the rays that are emitted from one position at the surface of a single fiber. When the light is uniformly distributed over the fibers at the incoupling end, then the irradiance distribution will be almost uniform in the near field and gaussian in the far field of the tip. At the fiber bundle surface itself, there will be gaps in the uniform distribution due to the dead space between the individual fibers. As mentioned above, this dead space is filled by overlapping beam spots at a short distance from the fiber bundle. If the beamspot at the incoupling end of the fiber bundle is gaussian or non-uniform, the irradiance distribution can still be uniform when there is no correlation between the position of the fibers at the input and output end. A correlation can be introduced on purpose to allow selective ablation by probing and ablating tissue through individual fibers as performed by the MIT [11] and Smart laser [21] systems. The fibers of the bundel can also be positioned in an asymmetral or ring geometry [20] resulting in a corresponding irradiance distribution.

3.3.4 Tapered tips

The geometry of the tapered fibers and laser scalpels, which are usually made of sapphire, can be defined by the taper angle α and the entrance diameter (Fig.1D,E). The taper angle may vary from 1 to 45 degrees. A ray reflects several times within the scalpel before its angle with respect to the scalpel surface exceeds the angle of total reflection. Then the ray refracts ('leaks') out of the scalpel. The scalpel may end in a point, in a flat or in a hemispherical tip. Figure 4 shows the beam profile of a sapphire laser scalpel (Surgical Laser Technologies) with a taper angle of 27 degrees.

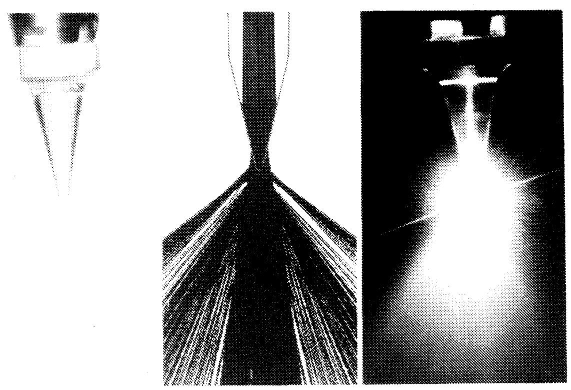

Figure 4:
Beam profile of 27° tapered sapphire tip.
From left to right: photo tip, calculated profile, observed profile.

The light from a laser scalpel consists of several conically shaped beams. The discrete angles of the beams (Fig. 4) are related to the number of reflections inside the taper [27].

For taper angles as small as 5 degrees a power density increase over 300 times relative to the power density at the entrance of the scalpel can be achieved. If the scalpel end is hemispherical rather than flat, the maximum irradiance increase is 10-30% smaller due to internal reflection losses [25,27]. The practical use of tip with a very small taper angle will be limited by mechanical strength. For effective tissue cutting, the scalpel tip may be shortened to optimize the irradiance increase for tissue cutting in combination with radial energy leakage to obtain controlled hemostatic coagulation, e.g. in liver resection [5].

4. PROBE INTERACTION WITH TISSUES: Optical, thermal and mechanical effects

The physical processes involved during interaction between optical probes and tissue can be differentiated in optical, thermal and mechanical effects [28]. The contribution of these effects are illustrated schematically in figures 5 and 7. The estimated importance of the physical processes is represented by the thickness of the lines used for arrows, boxes and characters. The physical processes involved, depending on the method of application of the probes, are discussed in more detail below. There is a significant difference in processes between probes used in conjunction with CW lasers (Fig. 5) and pulsed lasers (see paragraph 4.6 and Fig. 7).

4.1 Optical interaction

The refractive index of tissue varies between 1.33 and 1.5 depending on the water content [29]. Thus, in contact with tissue, probes behave as if in water. It is expected, however, that the beam properties of visible and near infrared light will already be lost after a few hundred micrometers due to scattering and absorption [30]. Thus, mainly the irradiance distribution at the surface of the probe is of importance.

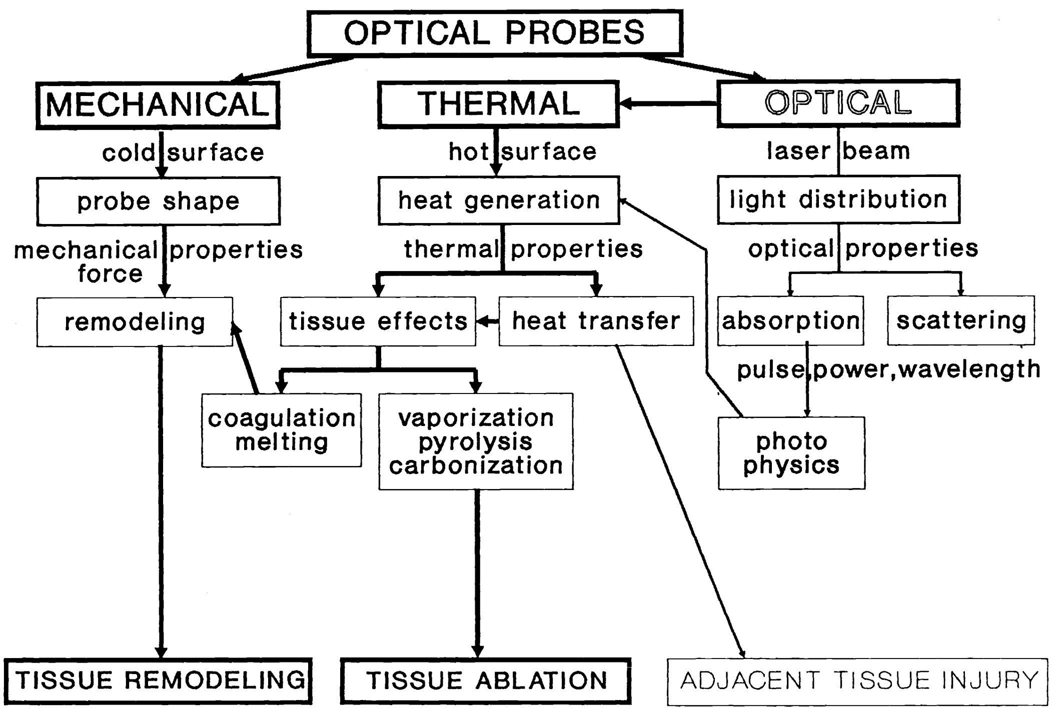

Figure 5: Contribution of the physical processes involved during interaction between optical probes and tissue for continuous wave lasers. The estimated importance of the physical processes is represented by the thickness of the lines used for arrows, boxes and characters.

4.2 Thermal interaction (CW lasers)

The thermal interaction of probes with tissue depends on the way energy is dissipated in a particular volume of tissue per unit of time. Thermal interaction is dominant using probes in combination with CW lasers.

4.2.1 Totally transparant probes

For totally transparant probes, the light has to be converted to heat by absorption in tissue, so heat generation is governed by the wavelength of the light and related optical properties of the tissue. Probe contact with the tissue ensures light delivery at the intended position without the risk of disturbance of the beam in the medium between probe and target tissue when there is no contact. Tissue contact also reduces energy loss due to scattering and reflections [31].

4.2.2 Low absorbing probes

When the probes are partly absorbing the energy, heat conducted from the probe to the tissue will contribute to the temperature rise and additional tissue effects [32]. Increased temperatures will affect the optical properties of the tissue and will enhance both scattering and absorption. Especially when the tissue carbonizes, the absorption increases tremendously and tissue is ablated effectively [33]. When the wavelength of the laser light is poorly absorbed by the tissue, energy is dissipated in a large volume. To ablate the tissue, the whole volume has to be heated to the ablation temperature. If only a small part of the tissue would carbonize, tissue would be ablated immediately. By using absorbers at the surface of a probe, hot spots can be created initiating tissue in direct contact with the probe surface to carbonize [34]. From that moment on, the carbonized tissue itself will continue the ablation process. An absorbing coating on the surface of the probe should be small in volume so only part of the absorbed energy is sufficient to reach ablation temperatures. The coating itself should be resistant to these temperatures.

4.2.3 Highly absorbing probes

When most of the light is absorbed in the probe itself, the tissue is heated solely by heat conducted from the probe. The heat flux from probe to tissue depends on the contact area with the tissue and the medium between probe and tissue which can be water, blood, steam or air. The resultant of light energy input and the thermal energy conducted to the environment, together with the thermal properties of the probe, will determine the probe temperature. If there is a feedback system present in the probe, the probe temperature can be controlled by varying the light input.

4.2.4 'clean' and 'dirty' probes

Ball shaped fibers, hemispherical contact probes and tapered tips are mainly used in combination with continuous wave lasers, especially the Nd-YAG laser. The probes are being used to coagulate and cut bloodrich tissues during surgery or for laser angioplasty. Tissue exposures up to 25 W for several seconds from a cw Nd-YAG laser (26 J/mm^2 fluence) are sometimes not sufficient to start ablation in tissues which have low absorption and high scattering coefficients (eg. vascular tissue) for the 1064 nm wavelength [33]. Since both sapphire and silica are entirely transparant to the Nd-YAG wavelength, a new probe is ineffective. Even a surface IR coating which enhances absorption [24], as used by SLT, is not effective. In the absence of an appropriate tissue chromophore, a new probe can be pre-carbonized by irradiation in contact with blood or bloodrich tissue. Carbonized blood or tissue particles adhering to the front surface of the contact probe (Fig. 6) form absorption kernels which cause a fast local temperature rise at the front surface of the probe to about 1000 °C or higher during one 15 W, 1 s pulse [35].

Figure 6.
Hemispherical contact probe (3 mm diameter) with a ring of carbon particles on the probe surface together with the crater in bovine aorta following irradiation by Nd:YAG (15 W for 1s) The clear center corresponds to the beam spot (diameter 1.0 mm).

4.2.5 Non-contact application

Spherical probes are used in non-contact and contact mode for surgery. Using the probes combined with cw lasers in non-contact mode, tissue can be coagulated by delivering the energy in a large diameter spot on the tissue. Tissue is ablated effectively by using the focus of the beam in front of the spherical probe. The concentrated energy will initiate carbonization and subsequent ablation of the tissue. Ablation is continued by the high absorption of the carbonized tissue also out of focus. The ability to have direct control over the power density in a large range just by varying the distance to the tissue makes spherical probes ideal for laser surgery.

4.2.6 Contact application

Besides spherical probes, the tapered tips are used specially for contact ablation of tissue. In contact with tissue, similar effects will occur as discussed in the previous paragraph. Carbonized particles that adhere to the probe surface will create a hot tip which is effective in tissue ablation due to very high local temperatures [34,35]. Since coagulation of the environment is usually an intended effect, the probes can be used for exposure times of tens of seconds. It is advisable, though, to limit the exposure time to several seconds to prevent damage to the probe and to have more control on the temperature distribution in the tissue. Surgeons confirm that the carbonized 'dirty' probes work more effectively.

4.2.7 Probe damage

Earlier experience showed that bare fiber tips in contact with tissue are easily damaged [4]. Ball shaped fibers, on the other hand, can be used for long times without significant damage (our experience is over 20,000 joules, 5 - 15 W, 5 second pulses). Due to the larger volume and mass of the sphere compared to the cylindrical fiber tip, the heat generated at the hot spot on the surface is distributed in the whole ball resulting in lower peak temperatures. Still, the bright white flares observed at the surface probes indicate that very locally, carbon particles are burned and vaporized at temperatures exceeding 2000 °C, resulting in degradation of the probe surface [35].

Sapphire probes connected to the fiber are more heat resistant due to the higher melting point and high heat conductivity. Due to the high conductivity the probes cool rapidly which may result in sticking to the tissue. Silica probes tend to cool more slowly and are possibly already retracted from the tissue when the probe is in the temperature range of sticking. Since modification of the fiber tip itself by melting it into a ball shape or taper is more simple than the connection with a sapphire crystal, most laser surgery systems marketed nowadays consist of remodelled fiber tips. The products are cheap to produce and are intended for single use. Another advantage is the lack of an interface between fiber and probe so cooling is not necessary.

4.3 Mechanical interaction

Experimental data as well as clinical experience have demonstrated that tissue ablation/remodelling increases with increasing axial force exerted by the probes on the tissue [34,36]. It is obvious that mechanical effects must be involved when using eg. a 2.2 mm diameter contact probe with a beam spot diameter of 0.85 mm at the front surface. Only in and near the beam spot, ablative temperature can be expected. Using a cw laser, there will be a large temperature decrease from front to side, so the contact probe has a substantial non-ablative rim [35]. As a result, it is likely that the mechanism of action of the contact probe is a combination of ablation frontally and thermal and mechanical remodelling laterally.

Mechanical effects are also significant using pulsed lasers as will be discussed in the next paragraph.

4.4 Continuous wave lasers versus pulsed lasers

4.4.1 No thermal interaction ?

The strategy to use pulsed lasers is to ablate tissue efficiently with minimal adjacent tissue injury [37]. By choosing a laser wavelength which is highly absorbed by tissue, the energy deposition will be restricted to a small volume. Either UV lasers like the Excimer (eg. 308, 351 nm) or an IR laser like the Holmium (2.1 μm) are highly absorbed in tissue and can be transported through silica fibers [38]. The use of a pulse length shorter than the thermal relaxation time of the tissue (about 0.01 s) may prevent energy loss and adjacent tissue injury due to thermal conduction [39].

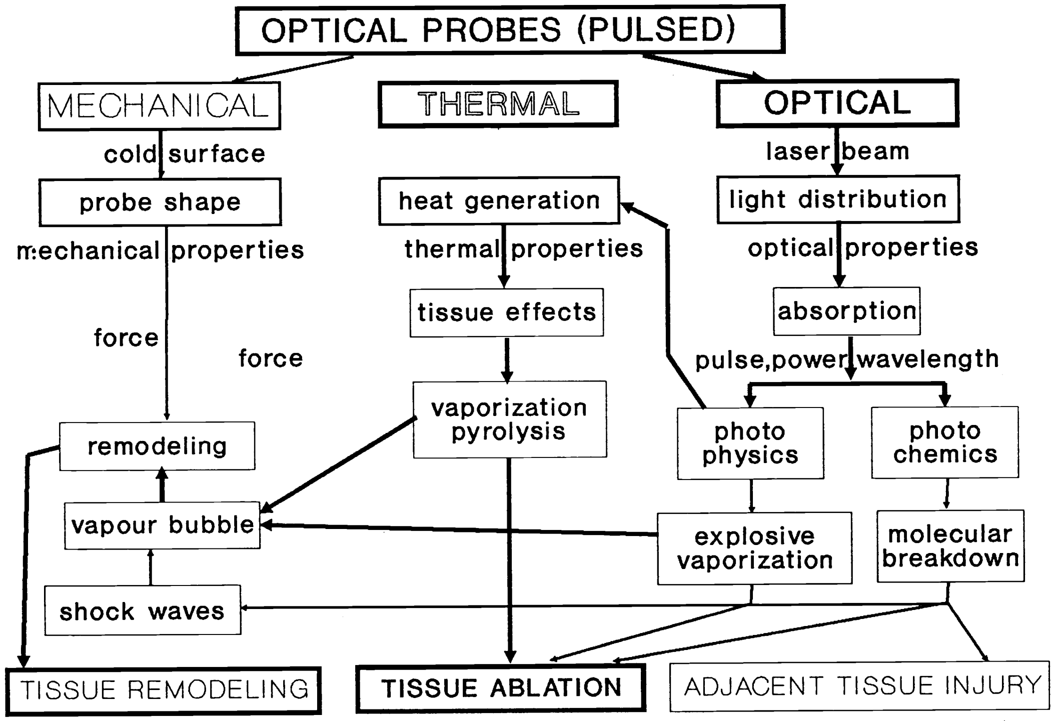

Figure 7: Contribution of the physical processes involved during interaction between optical probes and tissue for pulsed lasers. The estimated importance of the physical processes is represented by the thickness of the lines used for arrows, boxes and characters.

These pulses have been touted to be able to induce tissue ablation without heat generation but recent work has shown that a substantial part of the energy is dissipated as heat [40]. Already at pulse repetition rates of 5 Hz, there is a built-up of thermal energy near the probe resulting in a temperature rise which may be substantial.

In contrast to cw lasers, the probes coupled to the pulsed lasers do not need to be prepared with an absorbing layer to initiate tissue ablation, since the energy for ablation is used more efficiently.

From the experience with the optical probes coupled to cw Nd:YAG lasers, it was be essential to have a probe with the ablative beam covering the whole front surface [26]. If the ablated area in the tissue was smaller than the surface area of the probe, the thermal effects were insufficient to remodel the non-ablated tissue, necessary to advance the probe in the tissue. For this goal catheters with a bundle of fibers covering the whole probe surface have been designed [11,20].

4.4.2 Mechanical interaction

Short pulses (< 1 ms) result in other physical phenomena like explosive vaporization (comparable with 'popcorn' [33]). If a small volume of tissue water is heated above the boiling point, the volume will increase by a factor of 1600 at ambient pressure [41]. The moment water turns to vapour, the vapour pressure may be hunderds of atmospheres, and a bubble will expand forcefully until ambient pressure is reached. The mechanical effect of this vapour bubble may contribute to the ablation mechanism. It also induces mechanical injury to surrounding tissue which could be a drawback of the use of pulsed lasers [42].

5. CONCLUSIONS AND PERSPECTIVES FOR MODIFIED FIBER TIPS

5.1 Modified fiber tips in laser surgery

Ball shaped and tapered fibers allow good power control for cutting and coagulation of bloodrich tissue. These fiber tips are easy and inexpensive to produce and might even be repaired on site during treatment. The use of modified fiber delivery systems is promising for the rapidly expanding laparoscopic surgerical treatments because of there small size and flexibility [43].

5.2 Modified fiber tips with pulsed lasers

In combination with pulsed lasers, the thermal effects of modified fiber tips can be minimized or well controlled by the repetition rate of the pulses. High repetition rates will result in tissue effects comparable with contiuous wave lasers.

Mechanical effects of pulsed lasers and associated tissue damage can be substantial due to forcefully expanding vapour bubbles formed during tissue ablation.

5.3 Modified fiber tips in laser angioplasty

Multifiber delivery systems might be preferred for angioplasty due to their flexibility and steerability. In combination with pulsed lasers they allow precise ablation of tissue. It is possible to diagnose and selectively ablate tissue through individual fibers.

6. REFERENCES

1. Chang CT, Auth DC. Radiation characteristics of a tapered cylindrical optical fiber. J Opt Soc Am 68:1191-1196, 1987.
2. Russo V, Righini GC, Sottini S, Trigari S. Lens-ended fibers for medical applications: a new fabrication technique. Appl Optics 23:3277-3283, 1984.
3. Foschi A, Myers G, Crick WF, Friedberg HD, Snyder D, Nordtrom LA. Laser angioplasty of totally occluded coronary arteries and veingrafts: preliminary report on a current trial. Am J Cardiol 63:9F-13F, 1989.
4. Daikuzono N, Joffe SN: Artificial sapphire probe for contact photocoagulation and tissue evaporation using the Nd-YAG laser. Med Instrum 19: 173-178, 1985.
5. Joffe SN. Resection of the liver with the Nd-YAG laser. Surg Gyn Obstet 163:437-442, 1986.
6. Geschwind HJ, Blair JD, Mongkolsmai D, et al: Development and experimental application of contact probe catheter for laser angioplasty. J Am Coll Cardiol 9:101-107, 1987.
7. Kim LY, Roxey TE, Day AL. The Argon "contact" laser scalpel: technical considerations. Neurosurgery 21:858-860, 1987.
8. Hukki J, Krogerus I, Castren M, Schroder T. Effects of different contact laser scalpels on skin and subcutaneous fat. Lasers Surg Med 8:276-282, 1988.
9. McKenzie AL. How to construct a bulb-tipped fibre: part I - Theory. Lasers Med Sci 3:267-272, 1988.
10. Hussein H: A novel fiberoptic laser probe for treatment of occlusive vessel disease. SPIE: Opt Laser Tech Med 605:59-66, 1986.
11. Cothren RM, Hayes GB, Kramer JK, Sacks B, Kittrell C, Feld MS. A multifiber catheter with an optical shield for laser angiosurgery. Lasers Life Sci 1:1-12, 1986.
12. Sanborn TA, Faxon DP, Kellett MA, Ryan TJ. Percutaneous coronary laser thermal angioplasty. J Am Coll Cardiol 8:1437-1440, 1986.
13. Pilger E, Lammer J, Bertuch H, Stark G, Devinis M. Pfeiffer KP, Krejs GJ. Nd:YAG laser with sapphire tip combined with balloon angioplasty in peripheral arterial occlusions. Long term results. Circulation 83:141-147, 1991.
14. Lammer J. Pilger E, Karnel F et al. Laser angioplasty: Results of a prospective, multicenter study at 3-year follow-up. Radiology 178:335-337, 1991.
15. Sanborn TA, Faxon DP, Haudenschild CC, et al: Experimental angioplasty: circumferential distribution of laser thermal energy with a laser probe. J Am Coll Cardiol 5:934-938, 1985.
16. Fourrier JL, Brunetaud JM, Prat A, et al: Percutaneous laser angioplasty with sapphire tip (letter to the editor). Lancet I:105, 1987.
17. Sanborn TA, Cumberland DC, Greenfield AJ, et al: Peripheral laser-assisted balloon angioplasty. Initial multicenter experience in 219 peripheral arteries. Arch Surg 124:1099-1103, 1989
18. Sottini S, Russo V, Margheri G, Crea F. Probe for laser angioplasty radiating a corolla shaped beam. Applied Optics 28:995-999, 1989.
19. Hashimoto D. Development and early clinical application of a lateral aiming laser probe. Lasers Med Sci 2:25-28, 1987.
20. Duda SH, Karsch KR, Haase KK, Huppert PE, Claussen CD. Laser ring catheters in excimer laser angioplasty. Radiology 175:269-270, 1990.
21. Leon MB, Almagor Y, Bartorelli AL, Prevosti LG, Teirstein PS, Chang R, Miller DL, Smith, PD, Bonner RF. Fluorescence-guided laser-assisted balloon angioplasty in patients with femoropopliteal occlusions. Circulation 81:143-155, 1990.
22. Royston D, Waynant R, Banks A, Ramee SR, White CJ. Optical properties of fiber optics surgical tips. Applied Optics 28:799-803, 1989.
23. Verdaasdonk RM, Cross FW, Borst C: Physical properties of sapphire fibre tips for laser angioplasty. Lasers Med Sci 2:183-188, 1987.
24. Verdaasdonk RM, Borst C: Ray tracing of optically modified fiber tips. 1: Spherical probes. Appl Optics 30:2159-2171, 1991.
25. Ward H. Molding of laser energy by optic fiber tips. Lasers Surg Med 7:405-413, 1987.

26. White CJ, Ramee SR, Aita M, Samson G, Godfrey M, Hillier L, Murgo JP. Results of laser-assisted balloon angioplasty for peripheral arterial obstruction using a lensed fiber-tip delivery catheter. Am J Cardiol 66:1526-1528, 1990.
27. Verdaasdonk RM, Borst C. Ray tracing of optically modified fiber tips. 2: Laser scalpels. Appl Optics 30:2172-2178, 1991.
28. Verdaasdonk RM, Leeuwen AGJM van, Gijsbers GHM, Sprangers RLH, Gemert MJC van, Borst C. Modified fiber tips for laser angioplasty: mechanisms of action. J Interv Card 3:243-253, 1990.
29. Bolin FP, Preuss LE, Taylor RC, Ference RJ. Refractive index of some mammalian tissues using a fiber optic cladding method. Appl Optics 28:2297-2303, 1989.
30. Keijzer M, Jaques SL, Prahl SA, Welch AJ. Light distribution in artery tissue: Monte Carlo simulations for finite-diameter laser beams. Lasers Surg Med 9:148-154, 1989.
31. Welch AJ, Bradley AB, Torres JH, et al: Laser probe ablation of normal and atherosclerotic human aorta in-vitro: a first thermographic and histologic analysis. Circulation 75:1353-1363, 1987.
32. Welch AJ. The thermal response of laser irradiated tissue. IEEE J Quantum Electron QE-20:1471-1481, 1984.
33. Verdaasdonk RM, Borst C, van Gemert MJC: Explosive onset of continuous wave laser tissue ablation. Physics Med Biol 35:1129-1144, 1990.
34. Verdaasdonk RM, Jansen ED, Holstege C, Borst C. Mechanisms of cw Nd:YAG laser recanalization with modified fiber tips: Influence of temperature and axial force on tissue penetration in vitro. Lasers Surg Med 11:204-212, 1991.
35. Verdaasdonk RM, Holstege FC, Jansen ED, Borst C. Temperature along the surface of modified fiber tips for Nd:YAG laser angioplasty. Lasers Surg Med 11: 213-222, 1991.
36. Berengoltz-Zlochin SN, Westerhof PW, Mali WPThM, et al: Percutaneous Nd-YAG laser angioplasty of occluded femoropopliteal artery: 'hot' versus 'cold' recanalisation by hemispherical transparant contact probe catheter. Radiology 182:409-414, 1992.
37. McKenzie AL. An extension of the three-zone model to predict depth of tissue damage beneath Er:YAG and Ho:YAG laser excisions. Phys Med Biol 34:107-114, 1989.
38. McKenzie AL. Physics of thermal processes in laser-tissue interaction. Phys Med Biol 35:1175-1209, 1990.
39. Gemert van MJC, Welch AJ. Time constants in thermal laser medicine. Lasers Surg Med 1989; 9:405-421.
40. Gijsbers GHM, Sprangers RLH, Keijzer M, et al. Some laser-tissue interactions in 308 nm Excimer laser coronary angioplasty. J Interv Card 3:231-241, 1990.
41. Leeuwen AGJM van, Veen MJ van der, Verdaasdonk RM, Borst C. Non-contact tissue ablation by holmium:YSGG laser pulses in blood. Lasers Surg Med 11:26-34, 1991
42. Leeuwen AGJM van, Erven L van, Meertens JHJM, Motamedi M, Post MJ, Borst C. Origin of arterial wall dissections induced by pulsed excimer and mid-infrared laser ablation in the pig. (in press) J Am Coll Card 1992.
43. General Surgery (abstracts). Lasers Surg Med (suppl 3):24-26, 1991.