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TOPIC PAPER

Technical aspects of lasers in urology

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Abstract During the course of history a variety of laser principles have been introduced in surgery. Some erroneous developments probably could have been kept out of the market place if not for the magic which accompanies the acronym LASER and with more understanding for the underlying principles governing the process when light meets tissue. The interaction of light with tissue is exemplified on the basis of natural body chromophores when compared with available lasers at different wavelengths and operational modes. Furthermore the meaning of fibre flexibility and durability is elucidated.

Keywords Laser · Laser tissue interaction · Light tissue interaction · Body chromophores · Absorption length · Extinction length · Absorption spectrum · Nd:YAG · Diode laser · Wall plug efficiency · KTP laser · PVP · Holmium laser · Thulium · Vaporisation · Vaporesection · Thermal relaxation time · Pulse peak power · Laser lithotripsy · BPH · 2 Micron continuous wave · Laser fibres · Low OH silica fibres · Side firing fibre · Bare fibre · Front firing fibre · Laser fibre · Laser probe · Revolix

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Introduction

The advent of endourology opened a demand for surgical tools to be compatible with rigid and flexible endoscopes. For soft tissue surgery the wish list included precise incision and excision, no uncontrolled tissue damage and excellent haemostasis for the preservation of the endoscopic vision. For endourological stone management flexible probes are of interest to accomplish the technical needs given by flexible ureterorenoscopes. The principle advantage of endourology is the use of the natural orifice to access the pathological organ. The upkeep of this advantage requires for thin instruments and also probes for the delivery of energy to the surgical sites.

Fibre delivered lasers promise to fulfil such demands. Therefore there is a focus of interest for the development of instruments for endourological surgery. During the course of history a variety of laser principles have been introduced. Some were taken into consideration due to their shear availability. Few were developed for a surgical purpose initially. Some erroneous developments probably could have been kept out of the market place if not for the magic which accompanies the acronym light amplification by stimulated emission of radiation (LASER) and with more understanding for the underlying principles governing the process when light meets tissue.

Light tissue interaction

The basic understanding required to choose a suitable laser for a surgical application is in the optimisation of the absorption process of light in tissue. With regard to this process, laser radiation simply is directed light of a narrow bandwidth. This is synonymous to a single colour and applies for all regions of the invisible and visible electromagnetic spectrum.

Absorption is the most important but not the only process of light tissue interaction. When the laser beam encounters tissue a percentage of the laser beam is reflected by the boundary layer. The reflected radiation not only is lost for the surgical purpose but may constitute a risk to surrounding tissue as it may cause heat where an increase in temperature is not desired. Reflection mainly depends on the optical properties of the tissue and the irrigant surrounding it. The process of reflection is not very dependent on wavelength and therefore may be neglected when evaluating a laser wavelength for a surgical purpose.

From an optical point of view tissue is not homogenous and it will scatter an intruding laser beam. Scattering takes some of the beam out of its intended direction and in most surgical applications this portion is lost for the intended purpose. The degree of scattering depends on the size of the particles which the laser beam encounters and on the wavelength of the laser. Shorter wavelengths are scattered to a much higher degree than longer wavelengths. Blue laser radiation will be scattered more than green, green more than red and red more than infrared. When choosing a laser for a surgical purpose the scattering behaviour should not be neglected.

Absorption is the most important light tissue interaction process. On entering into an absorbing medium the intensity of the laser beam decreases exponentially (Lambert-Beer's law). Absorbed laser radiation is converted into heat and causes an increase in temperature. Depending on the amount of heat involved this will result in coagulation or even vaporisation of tissue.

The exponential behaviour and the immediate onset of the absorption process on entering of the laser beam into the absorbing medium implies more generated heat next to the surface than further below.

In order to achieve absorption a chromophore is required. Available body chromophores are melanin, blood and water. The wavelength dependence of their Absorption Length is shown in Fig. 1. The Absorption Length defines the optical pathway along which (1 - 1/e) or 63% of the incident laser energy is absorbed. Another concept—the Extinction Length—defines a depth until which 90% of the incident laser beam is absorbed and is converted into heat. There are 2.3 Absorption Lengths per Extinction Length.

Haemoglobin and the water molecule are widely used as chromophores for surgical lasers. Melanin has no importance for urological applications. The Absorption Length at familiar laser wavelengths in these body chromophores are shown in Fig. 1 and may help to understand the tissue effect of these laser systems.

For a short moment after the absorption of a circular laser beam the generated heat is confined in a cylindrical shaped volume with the height of the Extinction Length and the approximate diameter of the fibre. The effect to the tissue is determined by the density of the absorbed energy. The achieved effect along the Extinction Length needs to be matched with the intended surgical effect. At the same power level a laser wavelength with a long Extinction Length may create a deep necrosis whereas a laser wavelength with a much shorter Extinction Length leads to an increase of temperature above boiling point and to immediate vaporisation of tissue.

Common lasers

The Nd:YAG laser represents the long Extinction Length. At this wavelength of 1.064 μm the penetration is approximately 1 cm. Therefore this laser is rather used for haemostasis and coagulation of tissue like in the treatment of BPH using the Visual Laser Ablation of the Prostate (VLAP) technique. Clinical outcomes are good taking into account that necrotic tissue is not entirely removed during laser surgery but has to slough off and postoperative catheterisation may extend for weeks [1].

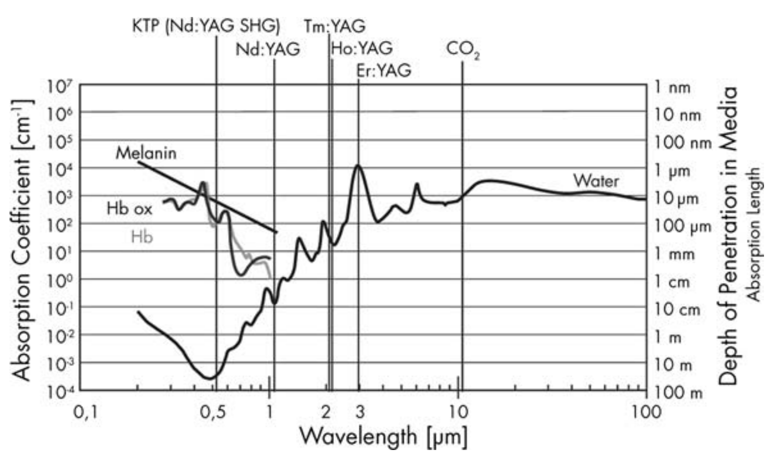

Fig. 1 Absorption spectrum of Melanin, Haemoglobin and Water in comparison to selected laser wavelengths. *Hb* haemoglobin, *ox* oxiginated

Diode lasers in the range of 808 to 980 nm experience a similar absorption in water and generate a similar tissue effect like the Nd:YAG laser [2]. Their main advantage above the aging Nd:YAG laser technology are smaller box size and much higher wall plug efficiency (meaning how much of the mains supply is converted into laser power). These differences result from the technical principles for the generation of the laser radiation. The traditional Nd:YAG laser uses an arc lamp to excite a Neodymium doped YAG crystal (Nd:YAG). YAG is the abbreviation for Yttrium Aluminium Garnet which is a synthetic crystalline material with garnet structure. Both arc lamp and laser crystal require intense water cooling. The overall efficiency of the Nd:YAG laser is of the order of 1%. In other words 10 kW of electrical power are required for the generation of approximately 100 W of laser power. This needs special electrical installation in most cases. Depending on the model the efficiency of diode lasers is more than one order of magnitude better. The thermal power loss of diode lasers is much less and therefore they are operated from a standard wall mounted power outlet.

The KTP laser is a derivative of the Nd:YAG laser. The additional KTP crystal in the laser resonator converts the fundamental Nd:YAG wavelength by Second Harmonic Generation into 532 nm radiation. This green wavelength is strongly absorbed by Haemoglobin. Therefore this laser represents the short Extinction Length in Fig. 1. The KTP wavelength penetrates vascular tissue only a few micrometers. In red, well circulated tissue the density of absorbed power is high and results in an immediate increase of tissue temperature above the boiling point. As a result tissue is vaporised leaving behind a coagulated seam where the increase in temperature provides haemostasis, and bleaches the haemoglobin, but is not sufficient for vaporisation. The following laser pass applied needs to travel through the coagulated seam where the laser beam experiences mainly scattering. The lack of absorption in coagulated tissue not only impairs its removal but the scattering of the green wavelength reduces the intensity and as well impairs the vaporisation effect of the next tissue layer. The KTP laser is currently widely used for Photoselective Vaporization of Prostate (PVP) [3].

The KTP laser operates at a wavelength where the absorption in water reaches a minimum. Without the presence of the Haemoglobin molecule the Extinction Length for KTP laser increases dramatically and the beam penetrates deeply into irrigant and/or tissue. In this case absorption takes place beyond direct visual control of the surgeon. Side firing fibres are used for PVP exclusively to ensure better visual control on the point of impact of the laser beam on the prostatic lobes.

The KTP laser was initially introduced for the dermatological applications at much lower laser power levels than currently used for PVP [4]. In addition the wavelength conversion reduces the poor Nd:YAG laser efficiency furthermore. Therefore KTP lasers sufficiently powerful for PVP require substantial electrical installation and in most cases external water cooling.

The understanding of the absorption principles in body chromophores initiated the development of the Holmium laser at a wavelength of 2.1 micron [5–8]. At this wavelength low Hydroxyl (OH) silica fibres for laser delivery are commercially available. Similarly to the Nd:YAG laser the crystalline matrix for the Holmium laser is YAG. Instead the Neodymium doping is replaced by Chromium, Thulium and Holmium which is mixed with the YAG melt from which the crystal is grown. The purpose of the Chromium is to absorb white light produced by a flash lamp and to transfer this excitation energy to the Thulium ions. Excited Thulium ions share their excitation energy with adjacent Thulium ions in the ground state and thereby double the number of excited ions at approximately half of the initial excitation energy. At this lower energy level the energy transfer into the Holmium ions and finally the laser transition takes place [8].

The sharing of excitation energy between Thulium ions (cross relaxation) is of paramount importance for the functionality of the Holmium laser. Without this feature the laser crystal itself would generate an excess of heat. An increase of crystal temperatures dramatically reduces the efficiency of this laser due to thermal population of the lower laser level and depopulation of the upper laser level. Finally the laser fails to operate due to overheating of the laser crystal. The cross relaxation between Thulium ions moderates the generation of heat during the excitation process, allows better temperature control of the crystalline laser rod and enables the Holmium laser to keep lasing in a repetition mode. Without the cross relaxation within the Thulium ions Holmium lasers would not operate at room temperature. Nevertheless high power Holmium lasers comprise big cooling systems for the enhancement of heat flow out of the laser crystals. Heat accumulation within the laser crystals restricts the Holmium laser under flash lamp excitation at room temperature to pulsed operation at moderate repetition rates. The overall efficiency is in the range of 1 % or less which requires special electrical installation for the operation of a high power Holmium laser.

The Holmium laser radiation experiences a short Extinction Length in tissue due to the strong absorption at 2.1 μm by the water molecule (Fig. 1). At this wavelength the depth of penetration is approximately half of a millimetre. The density of absorbed power in irrigant and/or in tissue is high and results in an immediate increase of temperature above the boiling point. In a typical endourological setting the onset of vaporisation is in the irrigant next to the fibre tip where a steam bubble is generated with each laser pulse.

The diameter of this bubble is of the order of a few millimetres depending on the laser pulse energy. The lifetime of this steam bubble is similar to the laser pulse duration which is of the order of 500 μs [9]. This is too short for standard videoscopic control and explains why the above mentioned steam bubbles remain invisible.

For Holmium Laser enucleation of Prostate (HoLEP) [10] the pulsating steam bubble is used to separate the prostatic lobes from the surgical capsule by tearing the tissue apart. In soft tissue surgery the tearing effect to tissue due to the pulsating steam bubble dominates tissue vaporisation in consequence of absorbed laser radiation in tissue. This explains the white fibrous appearance of the surgical sites during Holmium laser surgery on soft tissue under irrigation. The tissue effect is rapid and haemostasis of the Holmium laser in prostatic surgery is excellent.

Common pulse energy settings for Holmium lasers are in the range of 2 J. Depending on the flash lamp driver technology installed the laser pulse duration may be between 150 μs and 1 ms. The time required for heat to diffuse out of a short cylinder established by the fibre diameter and the Extinction Length (thermal relaxation time) is of the order of 100 μs. The heat generated during the absorption process accumulates during the duration of the laser pulse at the point of impact until heat conduction levels out the temperature profile. Applied to a urinary stone some of the laser radiation is absorbed inside the stone and generates an immediate build up of steam pressure which causes fragmentation.

In a way a laser pulse duration which is shorter or of the order of the thermal relaxation time confines the absorbed energy within the above mentioned cylinder. The shorter the laser pulse duration at a given pulse energy the higher the pulse peak power will be and the more effective is stone fragmentation. This feature is used successfully in laser lithotripsy with Holmium lasers using flexible silica fibres [11].

The awareness of the advantageous properties of the Holmium laser wavelength at 2.1 μm and recently available high power laser diodes allowed the development of a new laser at a wavelength of 2.0 μm and a continuous wave output dedicated for surgical applications [12–15]. Different from the flash lamp excitation of the Holmium laser, Thulium ions are directly excited by high power laser diodes. The narrow band excitation by laser diodes avoids excessive heat generation inside the crystal as in the case of broad band excitation of the Chromium ion by flash lamps. This technical improvement increases the power efficiency of the laser by a factor of about five and eliminates the need for special electrical installation.

Although the absorption characteristics in tissue are similar to the Holmium laser the properties of the continuous wave output in soft tissue surgery is superior. Due to the slightly shorter wavelength the Depth of Penetration is decreased to 1/4 mm. Instead of the tearing action on tissue which is caused by the pulsed operation of the Holmium laser the continuous wave output allows smooth incision and vaporisation of tissue with excellent haemostasis. Laparoscopic and open surgery is performed successfully opposed to the pulsed Holmium laser which generated a lot of tissue splattering with each laser pulse in a gaseous environment. For the treatment of BPH the 2 μm continuous wave laser removes prostatic tissue at a rate of up to 1.5 g/min [16].

A major technical advantage arises out of the temperature stability of the water molecule which is the target body chromophore for the 2 μm wavelength. Water retains its absorption properties when heated by the laser beam up to the boiling point which is the onset of tissue vaporisation. The tissue left behind after each laser pass is covered by a coagulated seam of tissue which provides haemostasis. It still contains sufficient water for efficient absorption of the following laser pass. Thus the laser tissue effect remains unchanged and effective throughout the entire surgical procedure.

Fibres

Laser radiation is a preferential energy in endourology if delivered through thin and flexible laser fibres which allow the combination with flexible endoscopes in surgical applications in the bladder and in the kidney. Laser fibres with a small outer diameter occupy less cross-section in any working channel for the benefit of irrigation flow and instrument diameter reduction.

In general the design of a laser fibre comprises a circular fibre core and two or three outer layers which are concentric to the fibre core. The core provides the delivery of the laser radiation whereas the circumjacent layers confine the laser radiation within the core and provide mechanical stability to the laser fibre. The core of a surgical laser fibre is made almost exclusively from silica. Visible and near infrared lasers (like KTP and most diode lasers) are delivered through fibres made from standard silica material. For the delivery of 2 μm continuous wave and Holmium laser radiation particular low OH silica fibres are required to avoid undesirable attenuation and power loss due to laser absorption in the Hydroxyl ion.

Different fibre designs at different pricing levels are on offer. Low OH silica fibres with an optical cladding of fluorine doped silica (silica/silica fibres) should be preferred instead of silica fibres with a cladding consisting of fluoroacrylate (TECS). The former provide superior confinement of the laser beam and allow a smaller bending radius when the fibre needs to follow the curvature of flexible instruments. In contrast laser radiation of the 2 μm range may leak from fibres with TECS cladding. Leakage of laser radiation from the fibre may cause severe damage to high value instruments like flexible ureterorenoscopes [17].

Excessive bending of laser fibres may be another reason for leakage of radiation from the fibre [18]. Even if the fibre mechanical is unharmed a point in bending may be reached where the confinement of laser radiation within the fibre core is not provided anymore [17]. Firing the laser in bursts instead of prolonged activation may reduce the risk of fibre failure if used in a deflected endoscope.

Notice should be given to the identification of the fibre diameter. Most vendors use the core diameter to identify a fibre. However the compatibility with an instrument depends on the outer diameter of the fibre which may be approximately twice as large as the core. Both dimensions should be available with the product literature. Thinner fibres in a general sense are more flexible and provide higher laser intensity at the distal tip.

Although surgical laser fibres are pretty robust each fibre should be examined to be free of kinks and free of shape and surface irregularities before it is connected to a laser and fired. Such damage may occur from improper handling and may result in explosive fibre failure.

Apart from differences in the design of core and cladding diameters known from bare ended front firing fibres, the industry offers surgical laser fibres with side firing tips. With respect to the laser tissue interaction side firing fibres provide no principle advantage compared to front firing fibres except for some beam expansion at the point of impact. Bare ended fibres implicate a contiguous hence sturdy design without transitions and bondings which may be reason of failure. Side firing fibres are used in surgical situations where forward firing may result in uncontrolled tissue damage and side firing provides better control on the point of impact of the laser beam.

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