# Introduction and aims

The Neodymium:yttrium-aluminum garnet (Nd:YAG) laser has an emission wavelength of 1064nm in the near infrared region of the electromagnetic spectrum. Infrared lasers have heat-producing characteristics which interact with tissues. Nd:YAG laser energy has been regarded as the most effective wavelength for soft tissues photocoagulation, because of its abilities to deeply penetrate and widely scatter in soft tissues. In dental hard tissues, it is transmitted in enamel, and is poorly absorbed and widely scattered in dentine. Studies have demonstrated that Nd:YAG energy is also negligibly absorbed by the dental pulp. This led to the possibility that Nd:YAG laser energy within human tooth is preferentially deposited at the dentino-enamel and pulpo-dentinal junctions. Since the thermal diffusibility of enamel and dentine is relatively low, this may cause subsurface heating, resulting in melting or microcrack formation at the dentino-enamel junction, and may also cause pulpal warming, resulting in irreversible pulpal injury.

In recent years, the introduction of a free running pulsed Nd:YAG laser (American Dental) in dentistry, was claimed to have the ability to induce surgical anaesthesia for cavity cutting, without resorting to supplementary

local anaesthesia. However, the procedure described requires exposure of the tooth crown to Nd:YAG pulse energy for a relatively long duration (240sec). Because the Nd:YAG energy is heat-producing, the possibility of thermally induced adverse side effects (immediate or late), on the dental hard tissues and the pulp, subsequent to such a prolonged surface irradiation and tissue warming, are the prime concerns of researchers.

The purpose of this study was firstly, to investigate this claim, in a double-blind, randomised, comparative clinical study with a potent topical anaesthetic EMLA 5% cream. Secondly, to carry out morphologic and histological investigations to detect any alterations or adverse side effects (immediate) on the dental hard tissues and the pulp. Thirdly, to study the laser's thermal profile and thermal safety limits during prolonged surface irradiation, as in clinical induction of pulpal anaesthesia and finally, to compare and correlate its thermal profile with highspeed bur cutting and other wavelengths such as Er:YAG and argon.

# ABSTRACT

It has been stated that "with Exposure to Nd:YAG laser pulses, the tooth became insensitive, resulting in surgical anaesthesia". This phenomenon was investigated clinically in a double-blind, randomised, comparative study with a potent topical anaesthetic, EMLA 5% cream, on human premolars. The degree of anaesthesia was estimated by an Electric Pulp Tester and by highspeed bur cutting of a standardised cavity. Results of this study demonstrated the clinical evidence of pulp anaesthesia with the application of pulsed Nd:YAG laser, and the depth of anaesthesia induced was comparable to that produced by EMLA 5% cream. It is postulated that the pulsed Nd:YAG laser can suppress the response of intradental nerves to an electric stimulus.

Subsequently, the extracted teeth from the above study underwent a series of morphologic and histological investigations which confirmed that exposure to Nd:YAG laser pulses for clinical induction of pulpal anaesthesia caused no adverse effects on the enamel and dentine, caused no immediate, atypical, superficial pulpal changes, but may cause a reduction in dentinal permeability.

Since the pulp absorption of this wavelength has been shown to be negligible, it seems the mechanisms of so called "Laser analgesia in lieu of anaesthesia" involve some local effects on dentine coupled with changes to neural functions either operating or initiating within the neuro-odontoblastic complex, at the dentino-pulpal interfaces.

In an *in vitro* laboratory study the laser's thermal profile and safety limit were investigated, by exposing caries-free extracted premolars (placed in a 37°C water bath) to prolonged Nd:YAG pulses, at a range of power parameters, using the procedure as in clinical induction of pulpal anaesthesia, while recording the pulp temperature changes by means of pulpally inserted thermocouples and a multi-channel chart recorder.

Results showed that the temperature changes followed a plateau profile and demonstrated a thermo-equilibium. The time to reach the maximum temperature, the mean maximum temperature, and the recovery time were a function of power and frequency. When using the energy parameters based on the manufacturers recommendation for clinical induction of pulpal anaesthesia, over a $9mm^2$ area, a rise in the intrapulpal temperature beyond the defined safety limit (5.5 °C) was likely.

A further laboratory study which compared the changes of intrapulpal temperature of Er:YAG laser, highspeed bur cutting, and argon laser confirmed that the Nd:YAG energy has limited thermal safety margin, especially when it is used at higher energy density and prolonged exposure.

It is concluded that clinical induction of pulpal anaesthesia by pulsed Nd:YAG laser is safe and effective, but when a higher energy dose is used, preventive measures such as forms of cooling or modifications of the irraditated surface area and exposure time, should be done to increase the thermal safety margin of this procedure.

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# CHAPTER I REVIEW OF THE LITERATURE

## 1. LASERS

### 1.1. History of lasers

The use of light as a therapeutic agent, or the use of an exogenous photosensitiser to absorb light and render a therapeutic effect, dates back to as early as 1400 B.C. In ancient Greece, the sun was used in heliotherapy to restore health. Chinese used the sun to treat various pathologies such as rickets, skin cancer and even psychosis and referred to it as phototherapy. The Indians and the Egyptians used a plant extract called Psoralens on skin, which was then exposed to sunlight, to treat vitiligo and psoriasis (Miserendino, 1995).

Less than half a century ago, a light source brighter than the sun was developed and given the name " LASER ," heralding a new era in photochemotherapy.

LASER is an acronym for "Light Amplification by Stimulated Emission of Radiation," which describes the basic principle by which all lasers operate. The roots of laser technology began in 1917, when Albert Einstein (Einstein, 1917) formulated the theory of " Stimulated Emission," which was to be the foundation of laser technology. He

theorised the existence of the phenomenon by which a photon can stimulate an atom or molecule and produce multiple identical photons. Microwaves were the first to be used to demonstrate this effect. During the early 1950s, American Physicist Charles H.Townes first reported the principle of Amplification of Microwave frequencies by the Stimulated Emission of Radiation process and this was called "MASER".

In 1958, Townes and Schawlow published a paper in which they discussed extending the maser principle to the optical portion of the electromagnetic field. The theory was applied by Maiman in 1960 to produce the first laser. A ruby crystal was used as the medium and was excited by a pulsed flashlamp to produce stimulated emission at a visible wavelength of 694nm. Since then, a vast number of lasers have been developed producing both pulsed and continuous wave coherent light on all parts of the visible and invisible spectrum. The range of use of these lasers is now great and diverse, from simple domestic use to sophisticated medical, industrial and even military applications.

## 1.2. Development of laser in medicine

Not long after Maiman's first firing of the ruby laser in 1960, the early medical researchers who saw potential value in lasers as surgical tools began their trials. A year later, the first gas laser was developed by

Javan *et al.*, (1961) using a mixture of helium and neon. In the same year, Elias Snitzer (1961) reported achieving laser action on barium crown glass doped with neodymium ions. The argon laser, with a range of emission wavelengths, 476-514nm in the visible portion of the spectrum, was developed by Bennett *et al.*, in 1962. This trend stimulated Geusic *et al.* (1964) and Patel *et al.* (1964) to pursue the development of the Nd:YAG and carbon dioxide lasers respectively, which they completed in 1964. In the same year, Goldman *et al.* investigated the transmission of laser energy through the optic fibre; unfortunately they were unable to transmit high energy laser beams without severe destructive changes.

In 1966, Yahr and Strully who were working with the laboratories of the American Optical Corporation developed a $CO_2$ surgical laser system and demonstrated that they could make a fine incision in skin and perform a partial liver resection with minimal blood loss on human subjects. Encouraged by these findings, Stellar *et al.*, (1974) first used the $CO_2$ laser to vaporise neural tissue directly at the time of surgery and Kaplan *et al.* (1973) first reported application of $CO_2$ laser in gynaecology. Between 1969 and 1975 numerous concurrent investigations were undertaken in the various surgical specialties using this laser. By the mid-1970s, the $CO_2$ laser had dominated the field of laser surgery and became the primary laser in neurosurgery, podiatry, gynaecology, otolaryngology

and general surgery. Despite promising findings in these early studies, more widespread use of lasers awaited the development of improved delivery systems and equipment which was largely due to the result of efforts by Takizawa (1978).

Since then the carbon dioxide laser has become the treatment of choice in the management of neoplastic lesions. This is due to its well-established capacities for greater precision, cleaner incisions, rapid cut and vaporisation and sealing of small vessels at the same time.

The Argon laser has remained the predominant laser type used for ocular photocoagulation (L'Esperance, 1968), and the treatment of choice to vaporise superficial or cutaneous vascular lesions such as port wine haemangioma and telangiectasia (Cosman, 1980; Apfelberg *et al.*, 1981). This is because its energy is strongly absorbed by haemoglobin, melanin and cytochromes and is readily transmitted through clear aqueous tissues. The Argon laser offers significant advantages in the treatment of smaller, superficial, moderately vascular lesions, compared to other wavelengths.

In contrast to the $CO_2$ laser, the Argon laser requires higher power densities and is dependent on tissue pigmentation and scattering to cause rapid vaporisation of tissue. Scattering, which results in an intratissue spread of the laser light before final absorption, disperses the tissue heating effect. The higher laser power available and the water dependent

absorption of the beam make the $CO_2$ laser a superior instrument for debulking of large, fibrous and relatively avascular lesions.

The first clinical application of the Nd:YAG laser in surgery was by Kiefhaber *et al.* in 1977, who used it to provide haemostasis during gastrointestinal surgery in humans. Hofstetter *et al.* (1978), tried coagulation of tumours of the bladder wall, transmitting the laser beam through an optical fibre inserted via the urethra. In 1979, Toty *et al.* reported the endoscopic use of the Nd:YAG laser in the treatment of tracheobronchial lesions.

In early 1980s, the Food and Drug Administration of the United States had approved the ophthalmic application of Nd:YAG laser in posterior capsulotomy. Utilising ultrashort pulsed Nd:YAG laser energy (Fankhauser, 1983) focused to an extremely small spot on an opaque posterior lens capsule, a photodisruption of the membrane was achieved. (Aron-Rosa and Aron, 1981; Keates, 1983). Since then, the short-pulsed Nd:YAG laser has become a standard tool of the ophthalmologist, used to perform posterior capsulotomies and other precise intraocular procedures that are difficult to achieve by means of the Argon laser or other modalities.

The Nd:YAG laser due to its fiberoptic capability, which allows the laser beam to pass down the channel of the endoscope and bronchoscope, as

well as its deep tissue coagulative and haemostatic abilities, has become the preferred modality for palliation of obstructing lesions and tumours of the airway (Fleischer, 1984) and photocoagulation of deep, extensive and vascular lesions (Shapshay and Oliver, 1984) of the head and neck. Although the Argon laser was the first laser used for endoscopic surgery in urology (Staehler *et al.*, 1976), it was gradually replaced by the Nd:YAG laser (Smith and Dixon, 1984; Stein and Kendall, 1985).

In 1984, synthetic sapphire laser tips had been developed for use with Nd:YAG laser. The sapphire tips vary the characteristics of the Nd:YAG laser by adding vaporisation to its coagulative property (Joffe, 1985).

Today, in medicine, the Nd:YAG laser has become a standard instrument in general laser surgery, complementing the $CO_2$ laser which originally dominated the field. Its applications extend to bronchology, ophthalmology, urology, gastroenterology, neurosurgery, dermatology and vascular surgery. Furthermore, new developments such as the possibility of concurrent and synergistic use of 1.318micron Nd:YAG laser-induced regional hypothermia to supplement photochemotherapy (Suaasan *et al.*, 1985; Waldow *et al.*, 1985) and laser tissue welding (Niijma *et al.*, 1987) are also receiving attention.

## 1.3. Development of laser in dentistry

Only a few years after the discovery of the first ruby laser in 1960, research was undertaken to introduce laser technology into dentistry with a view to replacing the conventional drill for cutting teeth.

The early dental researchers placed their attention on developing the basic laser energy parameters in relation to their effects on dental hard and soft tissues. Stern and Sognnaes (1964), reported that exposure to a millisecond (ms) pulse of ruby laser energy density at 500-2000 J/cm$^2$, resulted in cratering, melting, fusion and vaporisation of enamel, and postulated that ruby laser was not a suitable wavelength for tooth cutting. They then began to investigate the possible use of a ruby laser to reduce subsurface demineralisation of enamel (Sognnaes and Stern, 1965). The first case report of ruby laser exposure to a human tooth was by Leon Goldman, a physician, who applied two, one millisecond pulses of (17J/ pulse) ruby laser to the tooth crown of his brother, ironically a dentist. He reported that laser caused only superficial damage to the crown and the patient recorded no pain (Goldman *et al*., 1965). Taylor *et al*. (1965) first reported the histologic effects of the ruby laser on the dental pulp and the surrounding structures of incisor teeth and lateral borders of the tongues of hamsters. They demonstrated that when the tooth was irradiated with a 3ms pulse of the ruby laser (35-55J), it caused extensive pulpal necrosis

and damage to adjacent teeth and surrounding structures. These reports were later confirmed by Stern *et al*. (1969) and Adrian *et al*. (1971). They concluded that the energy density level of the ruby laser required to vaporise enamel was well above the safety threshold of the pulp.

By the end of 1960s, most dental researchers agreed that ruby laser had no suitable wavelength for the removal of tooth substance and continued to search for other lasers which might be more effective for application on dental hard tissues. Basic studies with Nd:YAG and $CO_2$ lasers followed into the 1970s.

In 1972, Stern and Sognnaes, examined the effects of superpulsed $CO_2$ laser on enamel surface under demineralised condition, in a combined *in vitro* / *in vivo* study. They demonstrated the lased enamel surfaces were more resistance to subsurface demineralisation than the unlased enamel, which was in agreement with their previous findings using ruby laser (Stern and Sognnaes, 1964). They also postulated that the mechanism behind the laser's ability to decrease demineralisation of enamel was due to the reduction of permeability through the fusion and sealing of its natural irregularities on the surface (Stern *et al*., 1972). Many investigators have attempted to transfer the result of these basic studies to clinical research to prevent dental caries (Kantola, 1972; Kantola *et al*., 1973).

Reviewing the early papers, Stern concluded that laser technology may not replace the conventional drill unless heat-related structural changes and damage to the surrounding dental tissues could be reduced dramatically by new laser systems (Stern, 1974). Since then investigators turned their attention to basic studies of the effect of laser on teeth, to decide whether laser applications really do offer new possibilities for advancement in dentistry.

In 1974, Yamamoto and Ooya, first studied the Nd:YAG laser's ability to alter demineralisation of enamel and showed that the Q-switched Nd:YAG laser at low energy density of 20 J/cm$^2$ induced physical changes on the enamel surface, such as a decrease in crystal size, and that this altered surface was less susceptible to demineralisation.

Adrian (1977) repeated and applied the methodology of his last threshold response experiment on ruby laser, but utilising a Nd:YAG laser. He found that even at energy densities great enough to cause cavitation of the enamel (4494- 6772 J/cm$^2$) no coagulative pulpal necrosis occurred and he concluded that the pulp was more resistant to injury by the Nd:YAG than the ruby laser.

The first reported application of a laser in the field of oral surgery was by Lenz, in 1977, who used an argon laser to create a nasoantral window (Lenz *et al.*, 1977). In the early 1980s, due to predominant use of the $CO_2$ laser in medicine, especially the successful and beneficial use in the speciality of otolaryngology, the dental profession searched for the possible application of $CO_2$ laser in clinical dentistry. Shortly afterwards, a number of papers were presented on the use of $CO_2$ laser in periodontology (Pick *et al.*, 1985) and oral surgery (Pick and Pecaro, 1982; Pecaro and Garehime, 1983). While other studies confirmed the usefulness of the $CO_2$ laser for the treatment of benign (Frame, 1985) and premalignant oral lesions (Frame *et al.*, 1984; Duncavage and Ossoff, 1986), attempts were made to fuse a synthetic hydroxyapatite compound to the enamel surface of extracted teeth by means of a $CO_2$ laser. They succeeded when the energy densities used were below 23.5 J/cm$^2$ (Stewart *et al*. 1985).

In the mid 1980s, possibly due to the widespread use of Nd:YAG laser in ophthalmology, Terry Myers (1985a; 1985b), a dentist and William Myers, an ophthalmologist, demonstrated on a human subject utilising a 30 picosecond ($10^{-12}$ sec) pulsed ophthalmic Nd:YAG laser operating at 3.4mJ with a spot size of 50 micron, vaporisation of visible incipient

carious lesions, leaving a clean enamel surface while the patient recorded no pain. Soon afterwards, Terry Myers pioneered the research into developing a free-running, pulsed, neodymium doped. yttrium aluminium garnet (YAG) laser which was later manufactured and supplied by Sunrise Technologies, Inc., California. This model dLase 300, has a varying power range of 30-150millijoules. The laser energy is delivered to the tooth through a standard, quartz optical fibre having varying diameters from 200-600 micrometers. The average power of the device can be selectively varied from 0.3-3.0Watts with a frequency range of 10-30 Hz. The pulsed Nd:YAG laser operates with pulses of 150 microseconds duration. The Nd:YAG beam is invisible and requires a visible helium-neon beam of 1mW as an aiming beam, which is emitted coaxial with the Nd:YAG laser.

In 1987, an important year for use of laser in dentistry, the Food and Drug Administration of the United States gave its first marketing approval for the use of both $CO_2$ and Nd:YAG lasers for oral soft tissue surgery and for clinical study on patients, thus initiating an entirely new technology to dentistry.

Today, over 1000 papers have been published worldwide on the use of lasers in dentistry, and the number of people involved in laser research

both clinically and scientifically continues to grow rapidly (Frentzen and Koort, 1990). The lasers that have been accepted by the Food and Drug Administration for use in the United States are the $CO_2$, Neodymium:YAG, Argon, Erbrium:YAG, Holmium:YAG, and Diodes. The Excimer, soft lasers and frequency-doubled Alexandrite lasers are currently under review. At this point of writing, these lasers are only approved for oral soft tissue applications, but studies already done demonstrate the clinical efficacy and safety for the removal of enamel and caries using Er:YAG and $CO_2$ (9.3-9.6micron) lasers. The future of laser dentistry will be bright, but a long journey lies ahead.

# 2. FUNDAMENTALS OF LASERS

## 2.1 LASER PHYSICS

The process of laser emission has its foundation in the quantum theory of physics, in a model first postulated by Danish physicist Niels Bohr, in 1913 (Fig. 2.1). According to Bohr, an atom can exist in a certain number of discrete quantum or excited states, each having a well-defined energy associated with it. An atom can be excited by absorption of a quantum of energy (Fig. 2.2.a) and as it does so, the excited atom will become unstable and will soon return to its ground state by releasing the stored energy in the form of an emitted photon. This process is called spontaneous emission of radiation (Fig. 2.2.b). However, in spontaneous emission, the emitted photon travels randomly in any direction and bears no relation to an incident radiation field.

In 1917, Albert Einstein, in his treatise "The Quantum Theory of Radiation", proposed that an excited state is more likely to produce a photon when in the presence of a photon of identical energy or wavelength. The emitted photon would then be in phase and coherent with the incident photon. This process is called the stimulated emission of radiation (Fig. 2.2c).

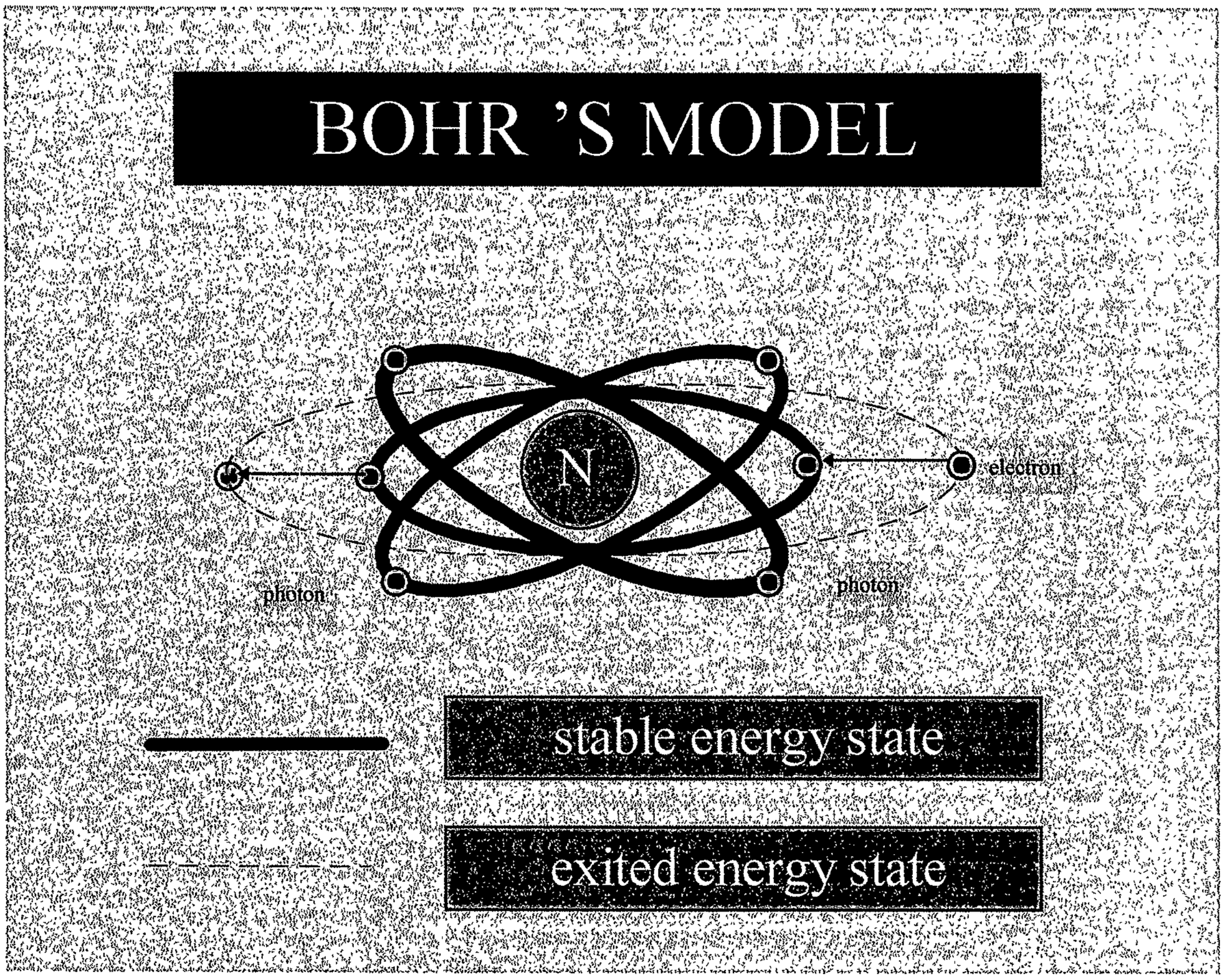



**Fig. 2.1. BOHR' s model- Quantization of the energy levels of atoms: Electrons are capable of changing their energy states by either gained or emitted quantum of energy, called photon.**

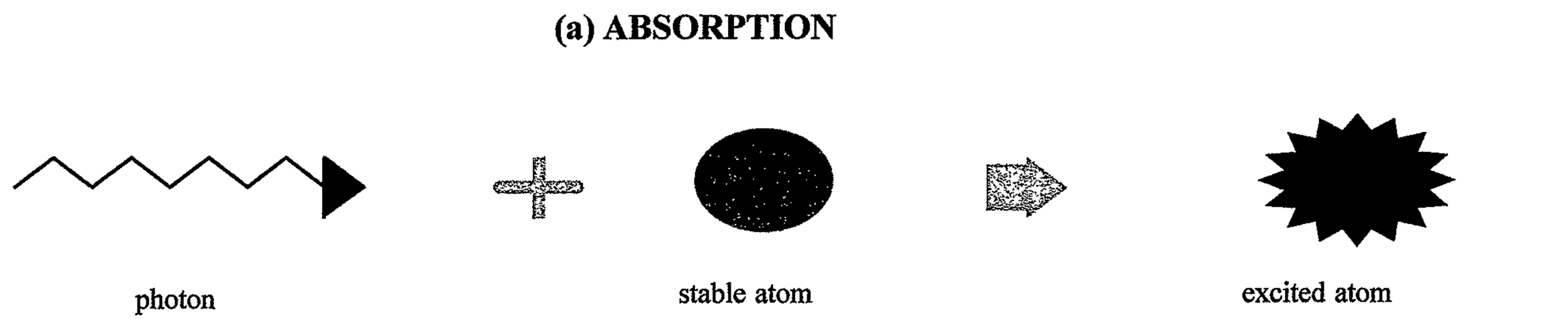



**Fig. 2.2. Diagram illustrating the process of absorption, spontaneous emission and stimulated emission. From Carruth and Simpson (1988).**

The result of stimulated emission is two identical photons travelling in the same direction and in phase. If these photons are not absorbed but allowed to continue then on encountering another excited atom in a similar fashion inside an optical resonator, an avalanche builds up, with virtually all the stimulated photons of identical wavelength (monochromatic) travelling parallel to each other (coherent) and in the same direction (collimated). Because the original incident photon has been amplified in numbers by stimulated emission, the process gives rise to the phase "Light Amplification by Stimulated Emission of Radiation", from which the word "laser" is derived as an acronym.

## 2.2. LASER SYSTEMS

Every laser system (Fig. 2.3) is composed of five basics components: an active medium, an excitation source, a cooling system, an optical resonator (laser head), and a delivery system (with focusing optics).

Lasers are named after their active media. The species of atom or molecule determines the wavelength of the laser beam. The substance which can be made to emit laser light must be sufficiently long-lived for a population inversion to occur. Active media are classified into three major groups based on the phase of the laser medium: solids (such as

# LASER SYSTEMS

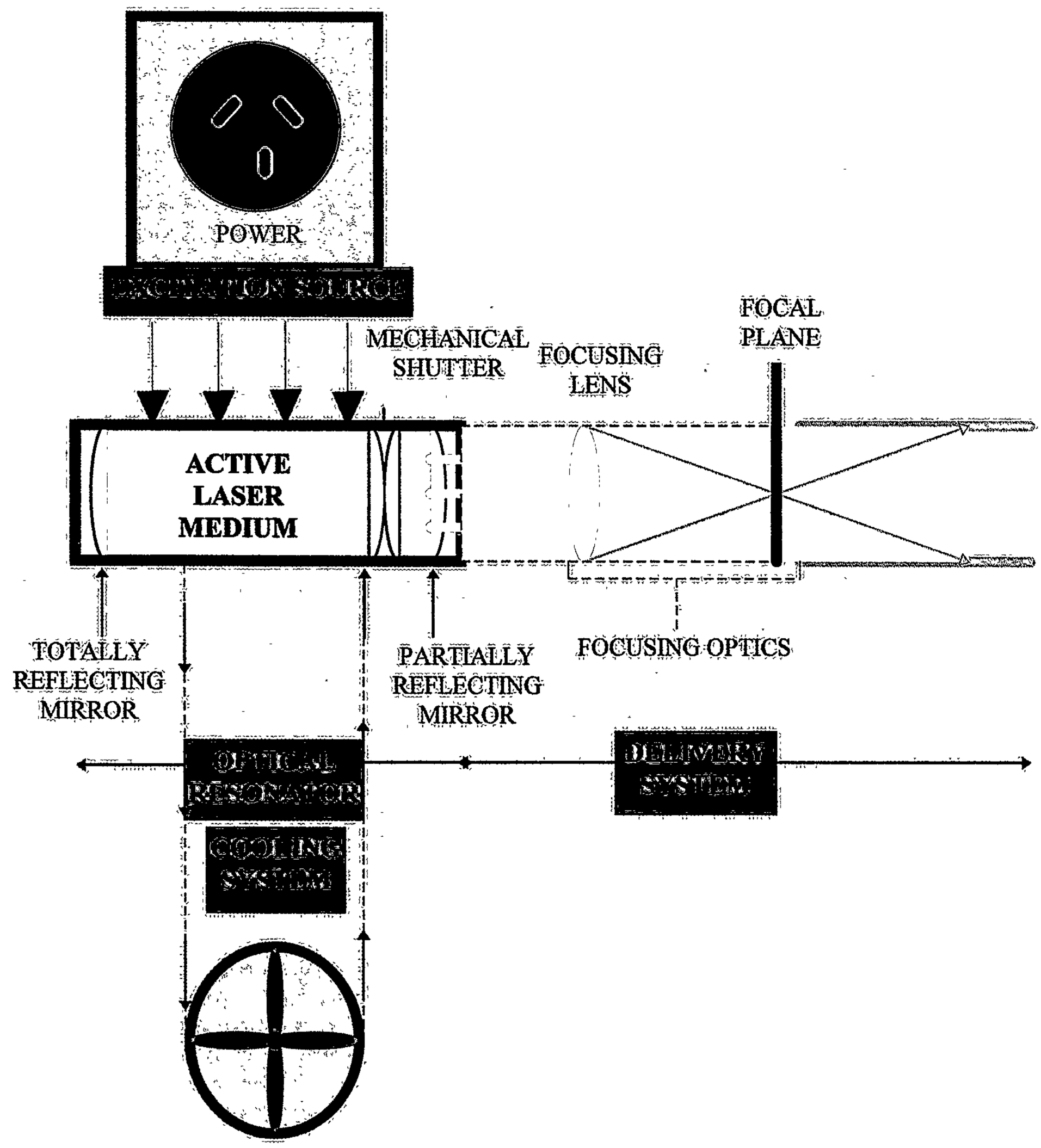



**Fig. 2.3. Laser system comprises of five basic components: an active medium, an excitation source, an optical resonator, a cooling system and a delivery system.**

doped glasses or semiconductor crystals); liquid (dyes); gas mixtures (such as carbon dioxide, the noble and halogen gases).

An excitation source (electrical, chemical, flashlamps, radio frequency emission, or other laser systems) that stimulates and amplifies the active medium, creates a greater population of atoms in a higher energy state. This process of creating a majority of excited atoms is known as population inversion (Fig. 2.4). The excitation of the active medium and subsequent continuous photon emission can be moderated by pulsing the pump source at various time intervals to produce pulsed laser output.

The medium used to create the laser energy is enclosed within an optical tube called an optical resonator (or laser head) with highly reflective mirrors at each end. These mirrors resonate the emitted photons back and forth in the active medium, and account for the characteristic parallelism of the laser beam. Each time the photons pass through the active medium, they are amplified further by the process of stimulation emission until a steady state level is reached, at which point all additional energy created within the cavity is used to maintain laser output under this positive feedback mechanism. This closed system that creates a very coherent, monochromatic and collimated form of light found nowhere else in nature. One of the mirrors is designed to be partially reflecting so as to allow transmission of a fraction of the

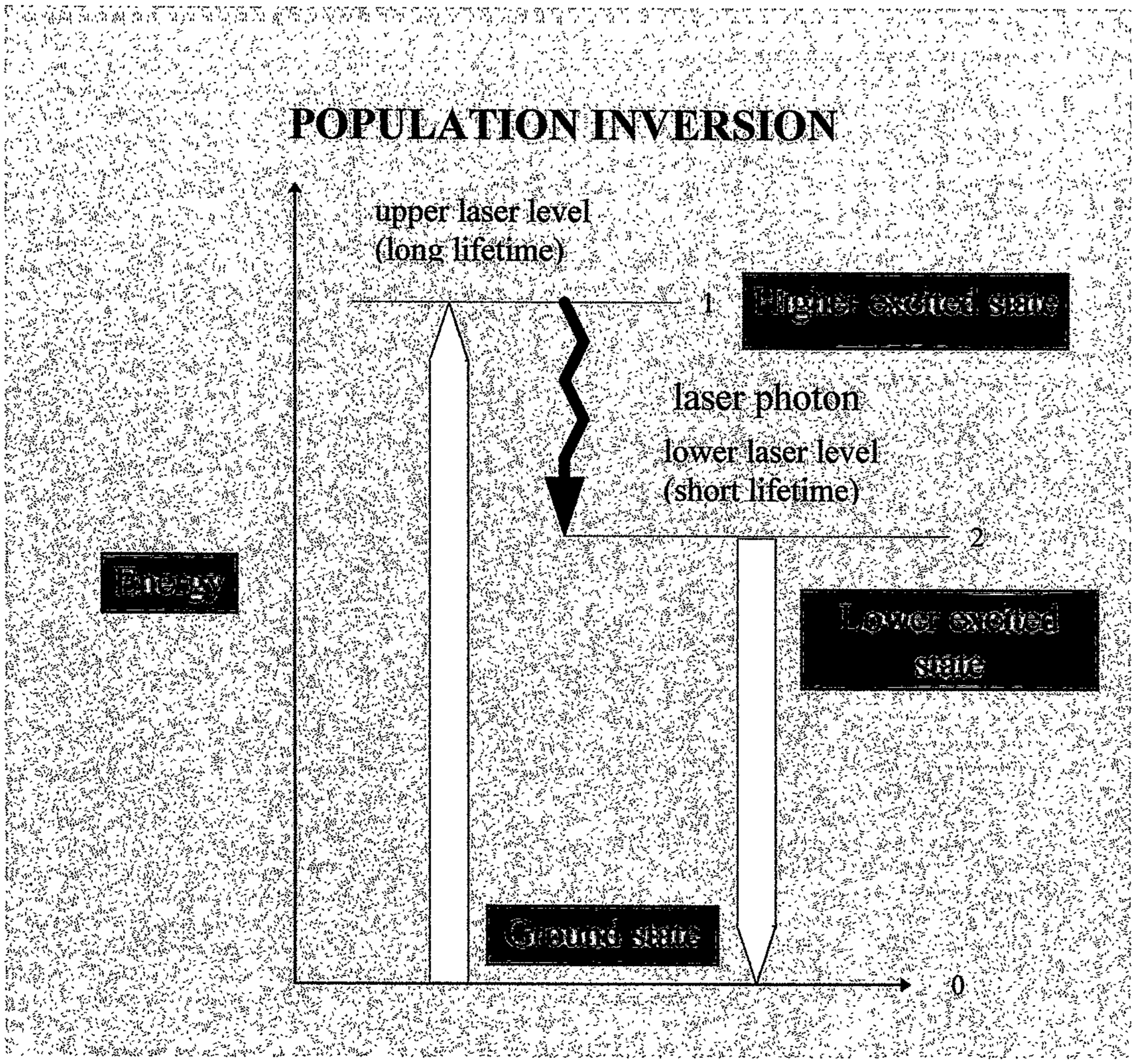



**Fig. 2.4. Lasing occurs when there is a greater population of atoms in a higher energy state than a lower energy state. From Carruth and Simpson (1988).**

concentrated laser beam out of the resonator where it is focused by a len or passed into an appropriate delivery system.

Some lasers have a mechanical shutter built into the laser head and positioned in the beam path. The opening and closing of the shutter is used to moderate the rate of laser emission and is controlled by a timing circuit which can be regulated by the operator.

Since the process of laser production is not completely efficient and some energy is dissipated as heat, a cooling system using either air(fan), water or liquid (radiator) is necessary to cool off the system during lasing process.

The delivery system is the device or attachment that transmits the laser energy to the tissue from the laser head (Fig. 2.5). There are rigid lenes with mirrors fixed to an articulated arm, non-contact and contact fibre delivery systems. In the rigid len system, the size and shape of the lenes determine the focal length and the spot size at the focal point. The beam from a len system can focus with minimum energy loss, whereas a bare optical fibre allows for beam divergence. For the fibre optic delivery system, the beam divergence is based on the numerical aperture of the fibre and whether the beam is used in air or in a fluid medium. The fibre diameter, beam divergence angle and the distance to the tissue determine the tissue spot size for fibre delivery and

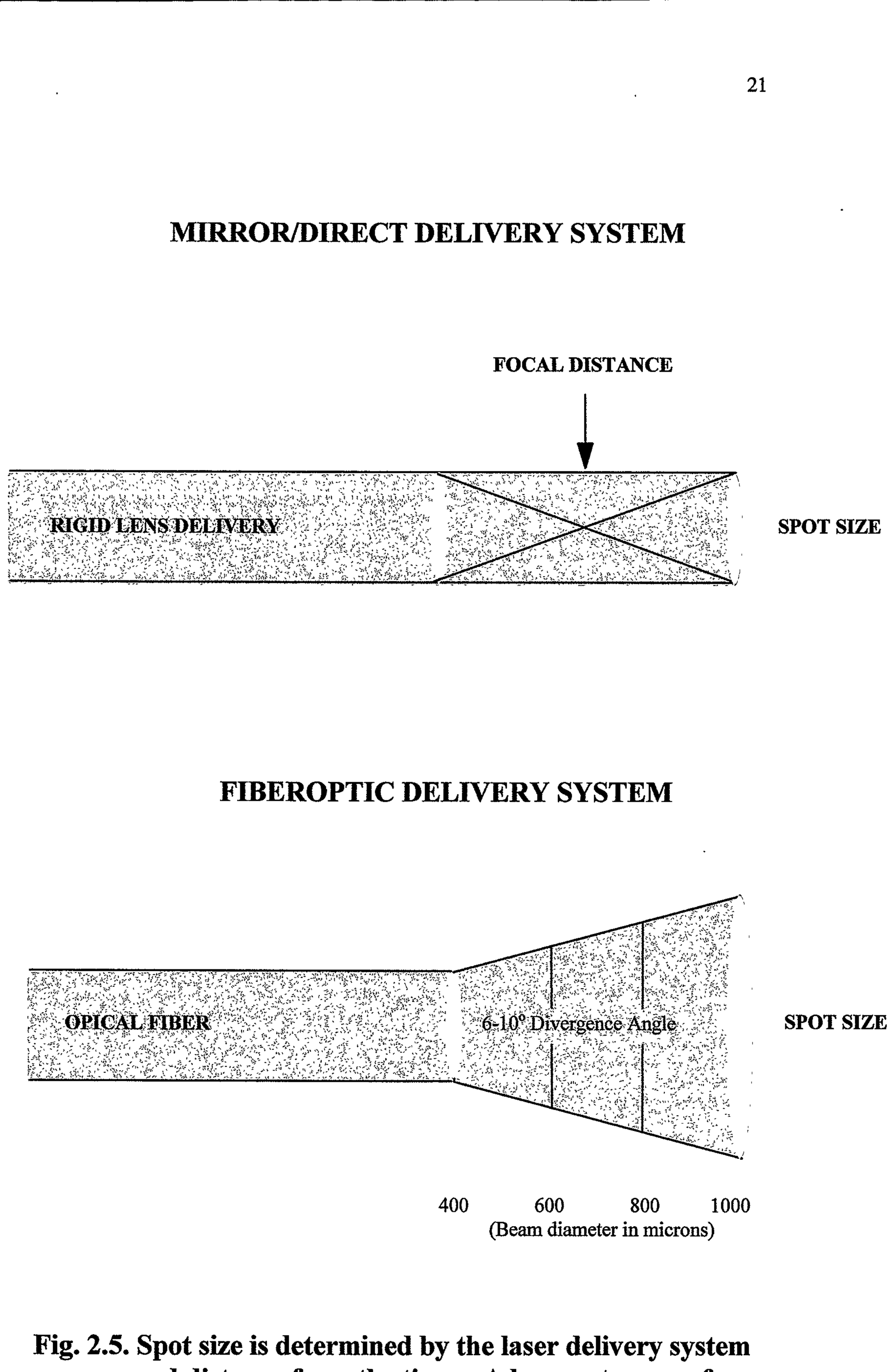



**Fig. 2.5. Spot size is determined by the laser delivery system and distance from the tissue. A lens system can focus the beam while a bare optical fibre allows for beam divergence, therefore lower the energy density.**

therefore determine the energy density at the target tissue. Maximum energy density or the smallest (brightest) spot size is achieved when the focal point is in line with the tissue plane, ensuring therefore the beam is focused. Other attachments to the various delivery systems are also available that further refine the method of laser delivery to the tissue, to give tactile sense on application and to minimise collateral tissue thermal damage.

## 2.3. LASER CHARACTERISTICS

### 2.3.1 ELECTROMAGNETIC SPECTRUM

Laser light is part of the spectrum of electromagnetic energy and is released as photons and possesses the properties (wavelength, amplitude, velocity and frequency) of a wave. The wavelength determines the functional properties of the laser energy. Surgical lasers fall between the longest and shortest wavelengths; in the infrared, and the visible as well as the ultraviolet portions of the electromagnetic spectrum (Fig. 2.6).

### 2.3.2. BEAM PROPERTIES

There are three optical properties that differentiate laser light from ordinary light (Fig. 2.7).

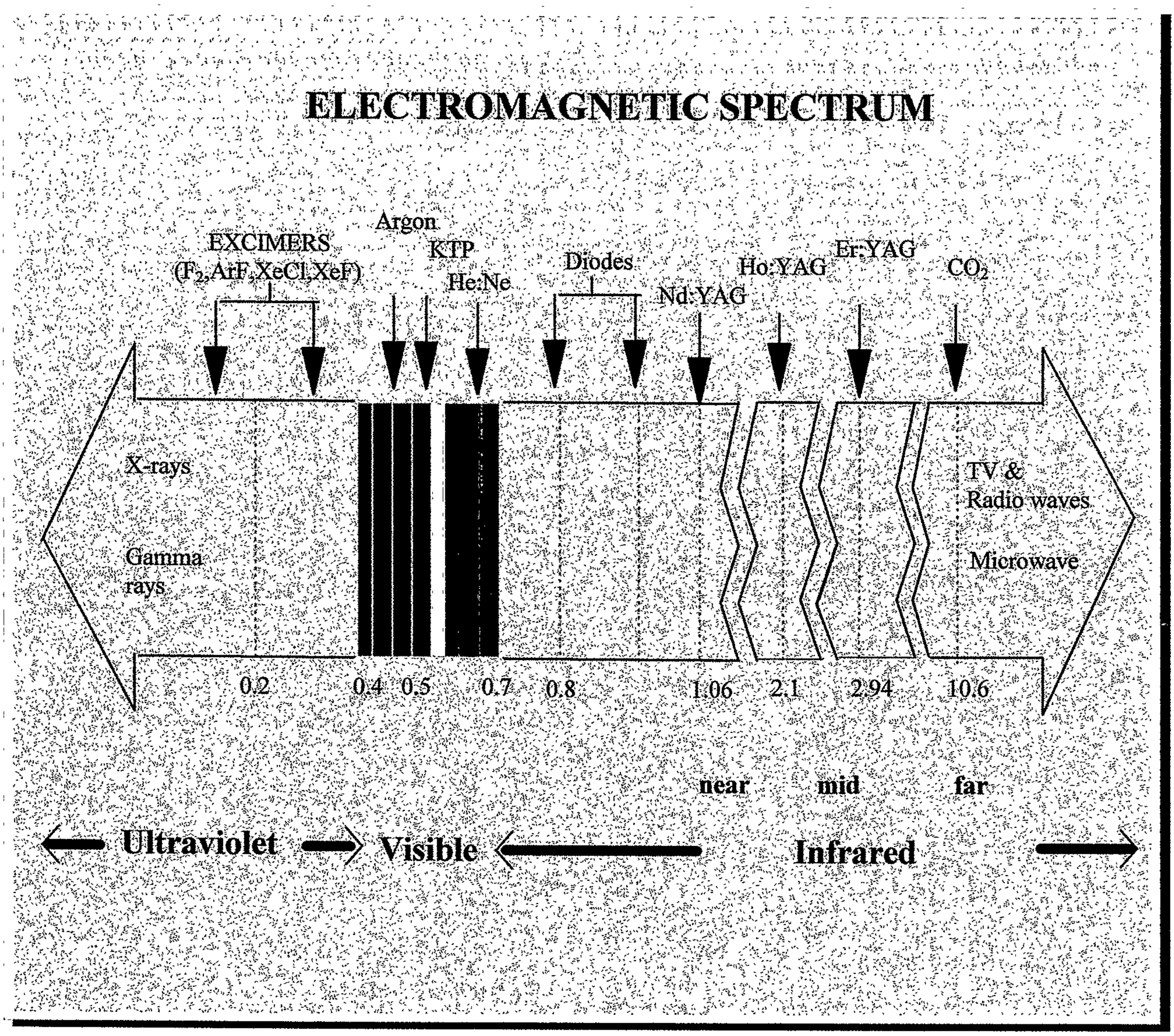
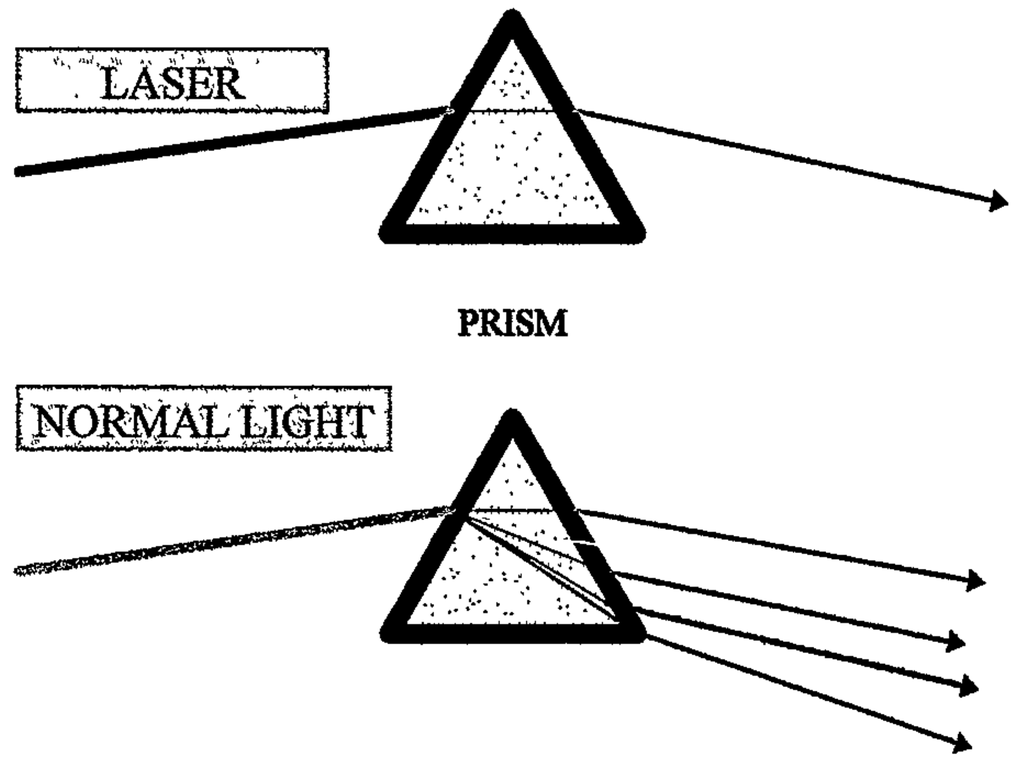
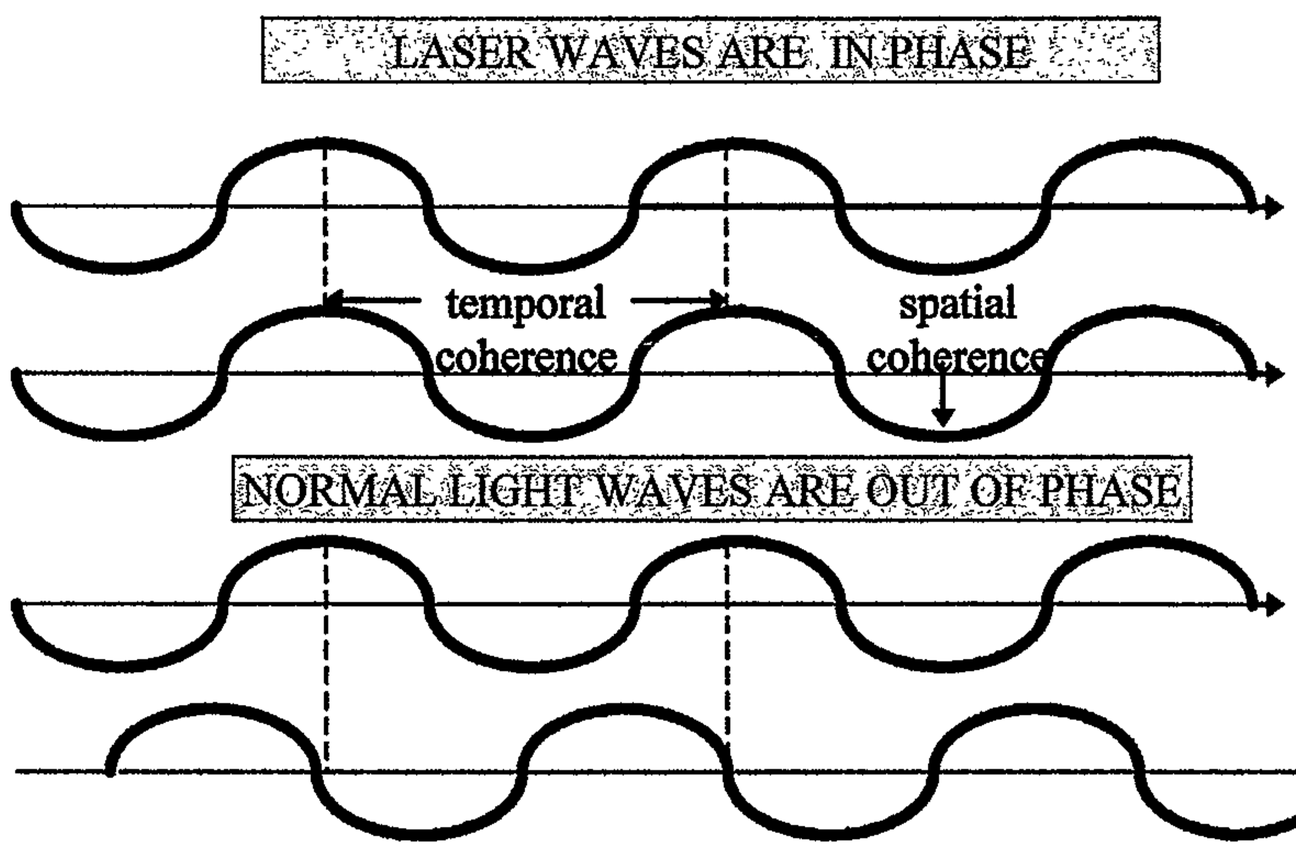



**Fig. 2.6. The electromagnetic spectrum of the most frequently used lasers in the dental and oral surgery. Modified from Trost *et al.* 1992.**

**MONO-CHROMATIC**




**COHERENT**




**COLLIMATED**

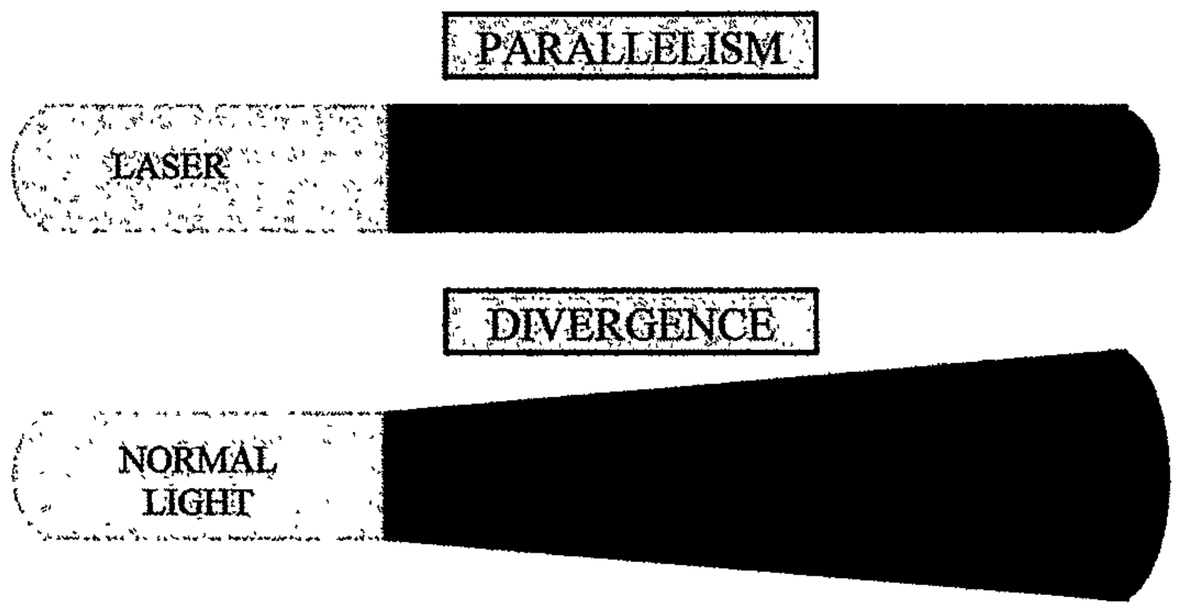



**Fig. 2.7. Laser beam properties: monochromatic, collimated and coherent.**

i) Monochromatic

Laser light consists of either one colour or single wavelength, or a narrow spectral of wavelength or colour. This property influences the focal spot size of the beam but also can be useful in instances where different target tissue absorption characteristics can be exploited.

ii) Coherent

Which means the laser waves travel in phase together in both time and space in the same direction.

iii) Collimated

Which means the laser waves travel parallel to each other and in a single direction with very little divergence even over long distances. This property enables the laser beam to be focused with great precision and high intensity either directly onto tissue or into a narrow optical fibre.

## 2.3.3. BEAM GEOMETRY

Beam geometry (Transverse electromagnetic mode-TEM) represents the profile of distribution of power across the diameter of the beam (Fig. 2.8). The quality of this beam geometry is determined by the stability of the laser system and the optics of the laser delivery system.

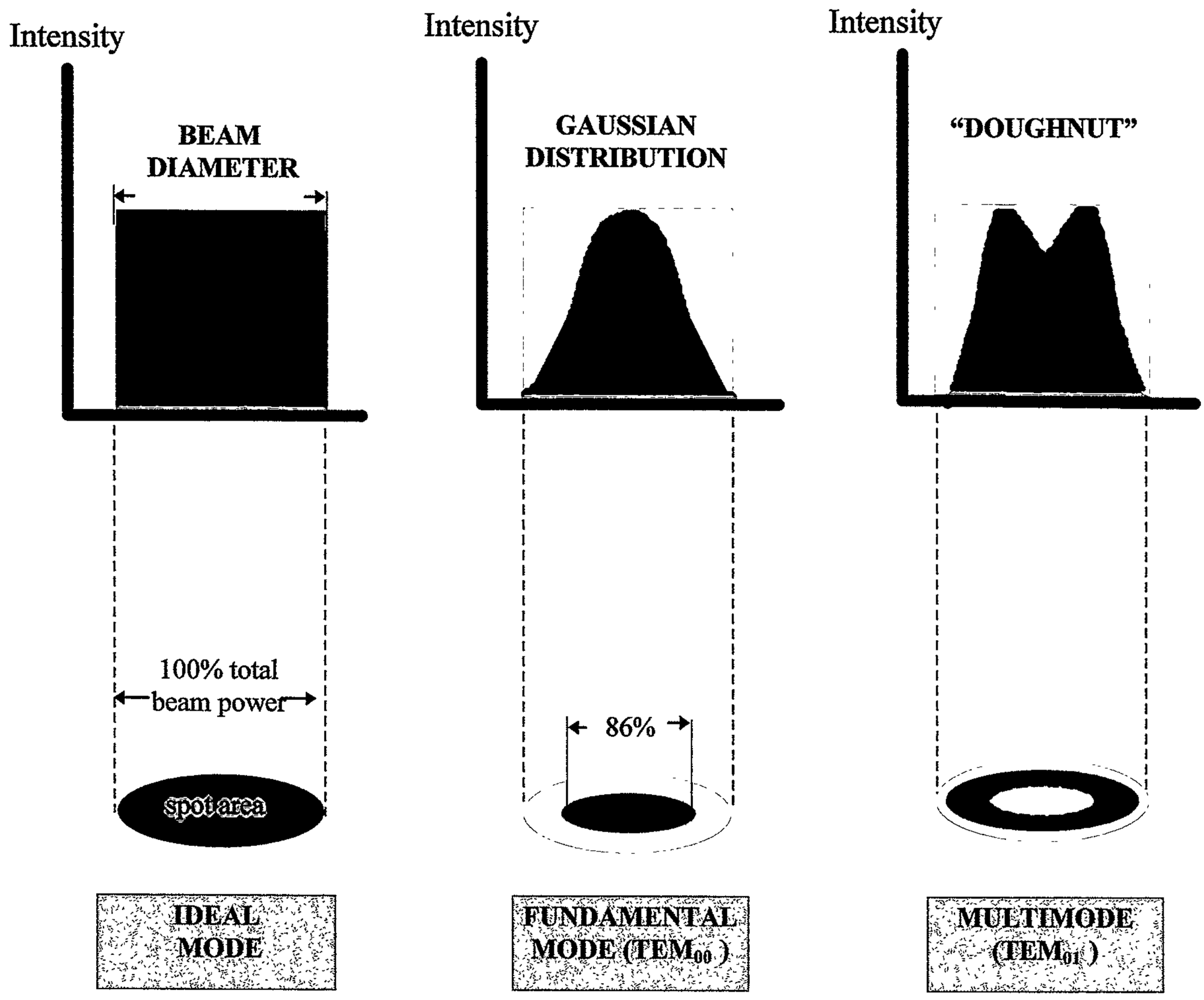



**Fig. 2.8. Beam Geometry represents the distribution of power across the diameter of the beam.**

The projection of the beam on the target tissue is called the spot. The beam geometry determines the precision of the spot by the power distribution over the spot areas, and therefore affects the power density at the target tissue (Polanyi, 1983). Ideally, the focal spot has 100% of the total beam power. The Gaussian or fundamental mode ($TEM_{00}$), which has a bell-shaped distribution, is the only mode that maintains the same intensity profile whether in focus or out of focus, and gives the smallest focal spot, approximately 86% of the total beam power (Fuller, 1984). Other beam geometries are referred to as multimode beams, and focus to larger spot diameters than the fundamental mode. A common example of multimode beams is a "doughnut" multimode ($TEM_{01}$), which has a cold spot in the centre and generates much lower power density than the Gaussian mode.

### 2.3.4. BEAM OPERATING MODES

Lasers are generally operated in one of the following modes:

i) Continuous wave mode (Fig. 2.9a) generates laser beam of constant power continuously until the termination of lasing.

ii) Gated mode (Fig. 2.9b) chops the continuous wave by means of a mechanical shutter into burst of laser energy that lasts approximately 1-200 msec with power less than that of a true pulsed laser.

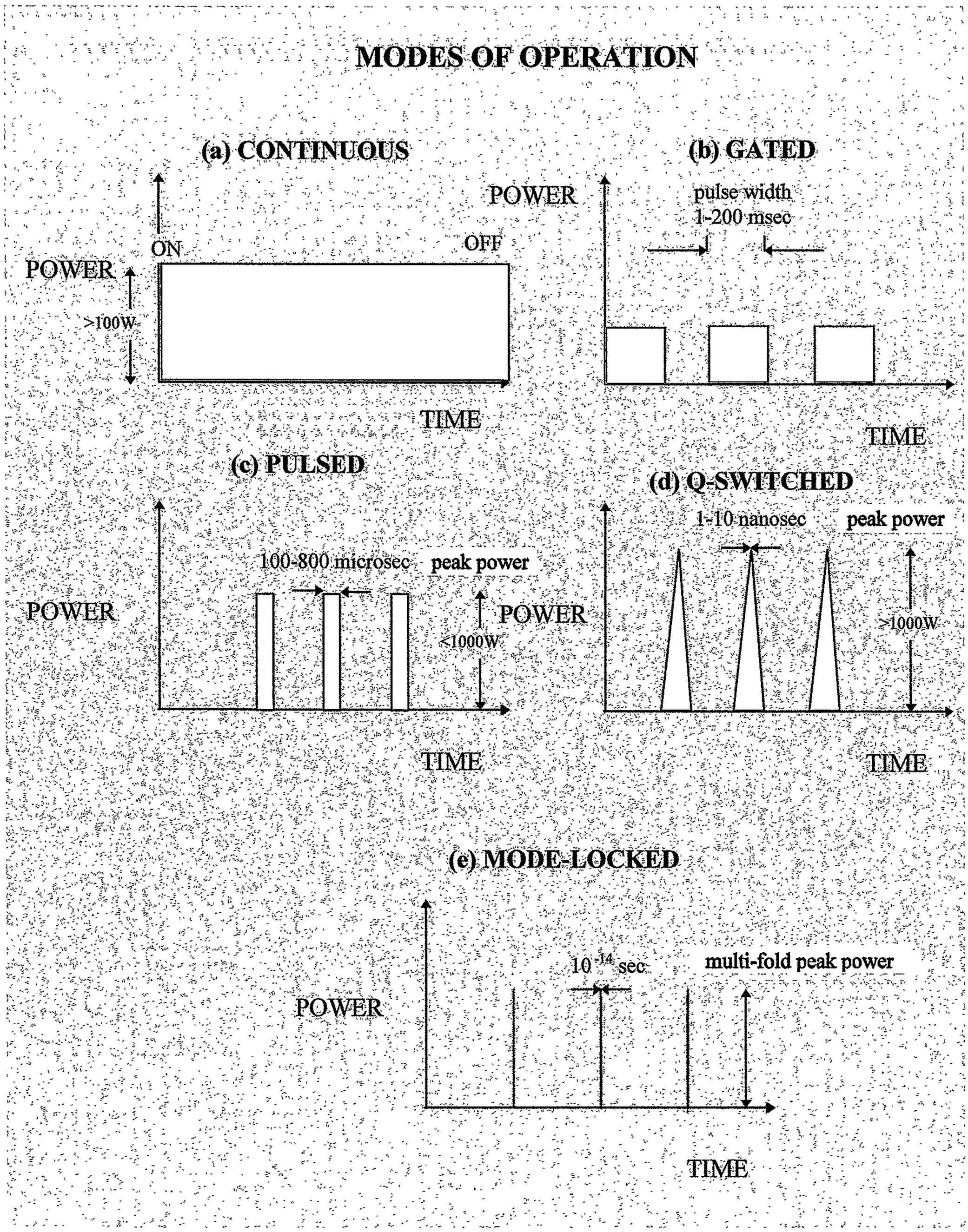



**Fig. 2.9. Beam operating modes: continuous wave, gated, pulsed, Q-switched and mode-locked.**

iii) Pulsed mode (Fig. 2.9c) is where the excitation of the active medium and subsequent continuous photon emission is moderated by the rate of pulsing of the excitation source at various intervals. This will produce pulsed laser output up to several hundred watts of peak power. A typical flash-lamp used for pumping the laser medium has pulse duration range between 100-800 microsecs, which determines the duration of the output pulse. Pulse mode allows tissue cooling between each burst of energy, to narrow the zone of thermal damage. It also increases the peak power density while keeping energy density constant, without a great rise in collateral tissue temperature.

iv) Q-Switched mode (Fig. 2.9d) produces high energy of several hundred millijoules, short-duration (1-10 nanosec) pulses of laser energy. Q-Switched mode uses some form of shuttering mechanism such as a rotating mirror as part of the optical resonator. The opening and closing of the shutter can restrict the lasing to a very short time interval. Higher energy can be stored within the resonator during the closure of the shutter. When the shutter is opened, all the energy accumulated in the active medium is discharged in a giant pulse.

v) Mode-locked mode (Fig. 2.9e) is used to generate ultrashort pulses as short as 10 femtoseconds ($10^{-14}$ sec) and this leads to a multi-fold increase in the peak power of the pulses. The process of mode-

locking involves the establishment of a definite phase and amplitude relationship between modes by selective locking of longitudinal modes which will usually oscillate simultaneously, so that the generating of ultrashort pulse will be permissible.

### 2.3.5. LASER POWER PARAMETERS

There are a number of important terms which are commonly used to describe the laser energy irradiation on tissue (Fig. 2.10).

Laser power is expressed in watts (W) and is used to describe continuous wave lasers. Peak power (as high as 1000W) is usually used in pulsed laser to describe the point at which laser power is highest within the pulse width. For very short duration pulses, the average power over length of the pulse closely approximates actual peak power.

In continuous wave laser, average power is equal to laser power over the lasing time. In the case of pulsed laser system, because multiple laser pulses of very short duration are fired per second, the average laser power for a given pulse energy is defined as the rate of energy delivered to the tissue. In other words, the average power for a pulse laser is a function of the pulse energy and the repetition rate. Pulse energy (J/pulse) or energy per pulse is commonly used in pulsed laser system to describe the amount of work done or tissue removal per single pulse.

# LASER POWER PARAMETERS

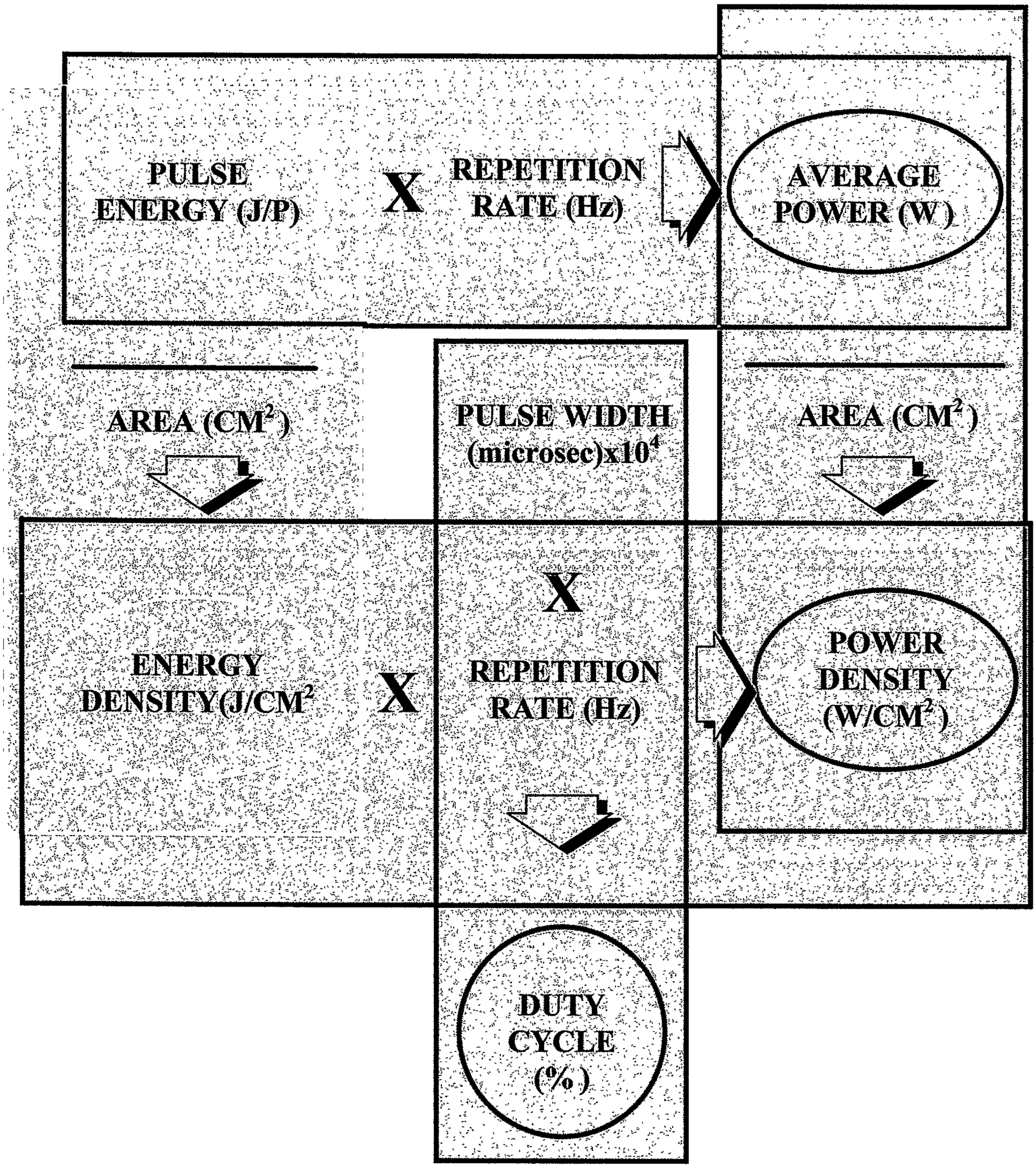



**Fig. 2.10. Diagram illustrating the relationships between various laser parameters that are used in the pulsed laser system.**

The term repetition rate (Hz) or pulses per second describes how quickly the laser energy is delivered and determines the total energy dose.

When tissue is irradiated, the most important parameter is energy density ($J/cm^2$) or fluence which describes the amount of work done per unit area on the tissue (Fuller, 1984). Multiplication of the energy density by the repetition rate will give power density ($W/cm^2$) or irradiance. Energy density can be regulated by altering either the power, the exposure time or the spot size. If the exposure is constant, the ratio between energy density and the depth of tissue injury is linear (Karlan and Ossoff, 1984).

Another parameter that relates to pulsed laser system is duty cycle (%), which describes the fraction of time that the laser is actually impacting tissue and is a function of repetition rate and pulse width. The duty cycle is 100% for the continuous wave laser system. The pulse width or pulse duration indicates the amount of time in microseconds that the laser is being charged and can be applied to tissue. Ideally, if the pulse width is shorter than the thermal relaxation of the tissue, this will reduce the possibility of detrimental thermal tissue effects.

# 3. LASER-TISSUE INTERACTION

The previous section described how the electric or flashlamp energy was transformed into laser energy by the process of light amplification and with stimulated emission. When biological tissues is irradiated with a laser beam, the laser light can either be reflected, transmitted, scattered or absorbed (Fig. 3.1). Only the photon energy that is absorbed will undergo some form of energy transformation, by the interaction of matter, and this is responsible for the biological tissue reactions. This phenomenon of laser tissue interaction reverses the energy formula ( $E=mc^2$ ), which was previously described by Albert Einstein in 1917.

## 3.1. TISSUE RESPONSES TO LASER INTERACTION

The extent to which irradiant energy is absorbed by tissue can be classified according to the wavelength, the power density and exposure time of the laser (Fig 3.2), into four basic types (Harris and Werkhaven, 1989) and these responses may further be subdivided according to how they are manifest in clinical applications (Miserendino *et al.*, 1995). These are : 1) photochemical interactions; 2) photothermic (mechanical) interactions; 3) photomechanical (photodisruption) interactions; and 4) photoelectric interactions.

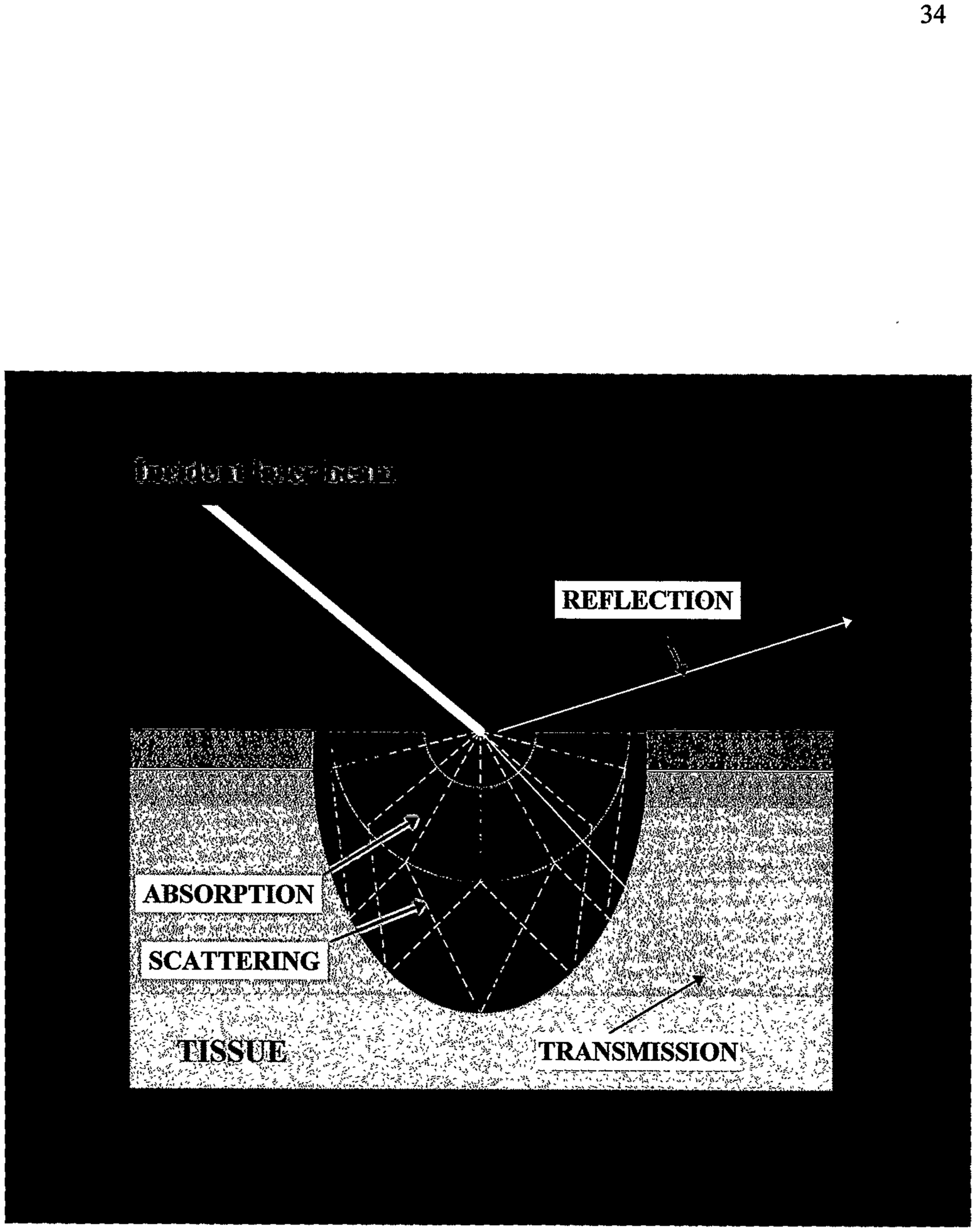



**Fig. 3.1. Four basic interactions of laser light with tissue: reflection, absorption, scattering and transmission; only photon energy that is absorbed will be responsible for the laser-tissue interactions.**

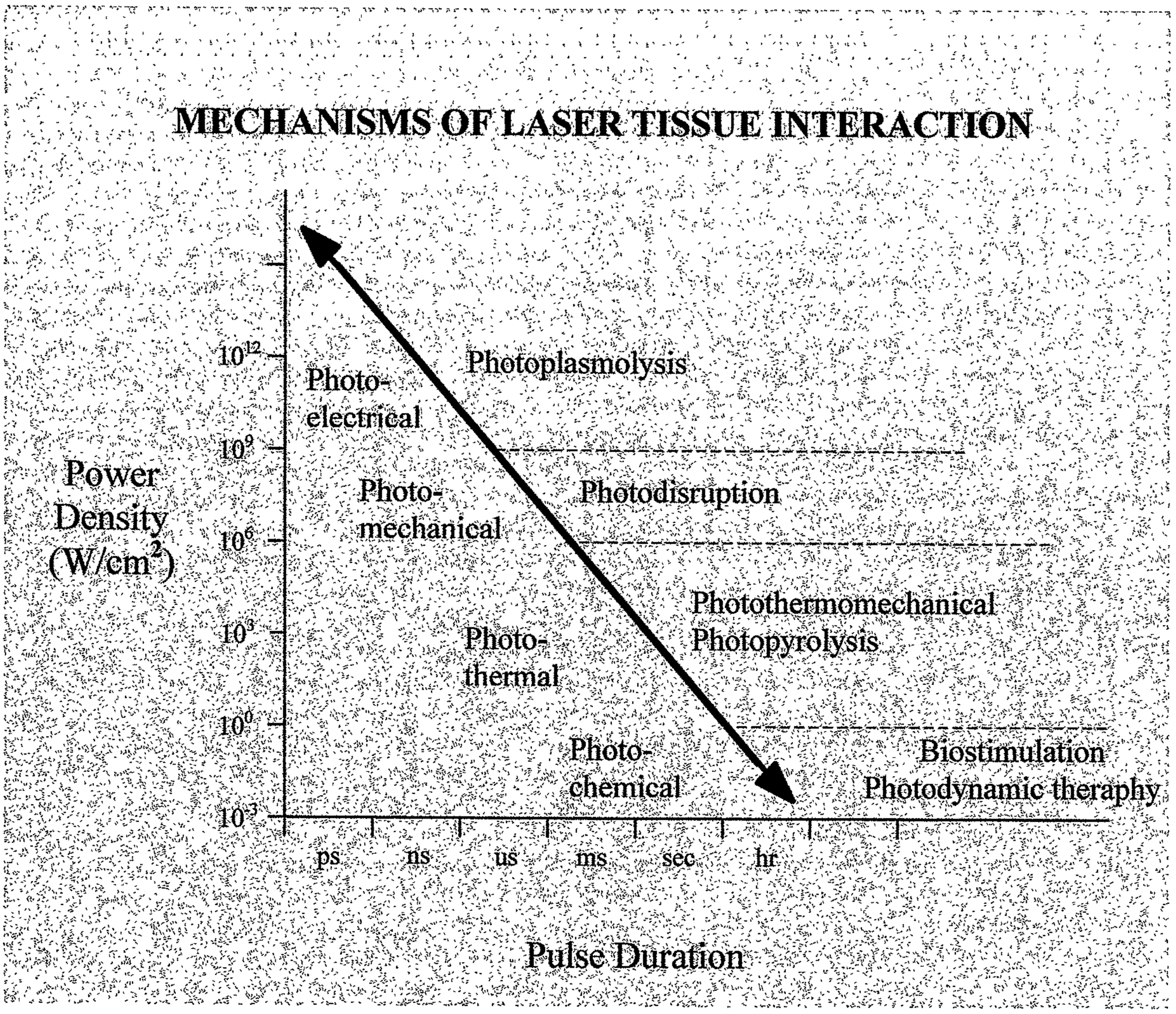



**Fig. 3.2. Relationship of power density and interaction time with the four mechanisms of laser tissue interaction and their corresponding clinical effects in tissue. From Harris and Werkhaven (1989).**

1) Photochemical interactions

The fundamental principle of the photochemical interaction is that particular wavelengths of laser light are absorbed by naturally occurring photoabsorbing substances, i.e. a wavelength specific light absorbing substances that are able to bring about biochemical reactions at the cellular level. Photochemical processes can become photothermic effects if energy densities are allowed to rise. Subdivisions of photochemical interactions are as follows:

a) Biostimulation; this describes the process in which the photon energy is converted into cellular or molecular energy which can produce either microcirculatory effects or can stimulate or inhibit certain cellular activities. Generally, it involves laser irradiations of relatively low intensity, with laser powers measured in milliwatts, producing little or no temperature changes (Basford, 1989). However, a long exposure duration resulting in a rise in temperature may be sufficient to cause thermal injury at cellular level (Matsumoto *et al.*, 1991).

b) Photodynamic therapy: This involves a light absorbing chemical such as haematoporphorin (photosensitiser) which is retained by highly metabolically active cells (tumour cells). Tissue treated with this chemical when exposed to laser energy of specific wavelength will

produce cytotoxic free radicals (eg. singlet oxygen) which result in destruction of target tissue.

c) Photochemical decomposition ( laser-induced tissue fluorescence or phosphorescent re-emission): When light energy is absorbed by the tissue components, it acts as a biologic capacitor, by momentarily storing the energy and subsequently releasing it again as light, with the energy level lower than the incident radiation. This interaction may be a viable method for incipient caries detection, and is currently under investigation.

2) Photothermic interactions

Photon energy absorbed by tissue components becomes transformed into heat energy. This kind of interaction is the basis for most types of surgical laser applications. When the laser energy density increases, the soft tissue (oral) temperature rises, and the tissue will undergo a series of changes from tissue warming, coagulation(photo), through vaporisation (photo) to carbonisation and finally tissue combustion (Table 3.1).

Photothermic-mechanical interactions involve local extensive heating of the water molecules within the mineralised tissues. This leads to water vapour expansion which causes internal microexplosions and fragmentation of mineralised tissues. For example, the Er:YAG laser can thermally induce ablation of dental hard tissue through this process.

**Table 3.1. The effects of rise in temperature on oral soft tissue**

| TISSUE TEMPERATURE °C | ENERGY DENSITY J/cm$^2$ | VISUAL CHANGE | TISSUE EFFECTS |
|---|---|---|---|
| 37 | 0 | NO | NORMAL |
| 43-60 | 10-150 | NO VISUAL CHANGE | WARMING |
| 60-65 | 150-200 | BLANCHING | COAGULATION |
| 65-90 | 200-900 | WHITENING | PROTEIN DENATURIZATION |
| 90-100 | 900-1000 | SHRINKAGE | VAPORIZATION |
| 100-150 | 1000-5000 | CHARRING | CARBONIZATION |
| > 175 | > 5000 | SMOKEY PLUME | RAPID CUTTING |

The thermal injury of the adjacent tissue can possibly be reduced by water cooling during laser application (Walsh and Deutsch, 1989) to restrict the laser effects on the surface.

3) Photomechanical (photodisruption) interactions

These are nonthermal processes, characterised by the transformation of photon energy into mechanical kinetic energy which can create rapid internal expansions and cause rupturing of the intermolecular or atomic bonds resulting in microexplosion of tissue. These interactions require an extremely high power density delivered in ultra pulse-duration ($<10^{-6}$ sec) from the radiant energy with short wavelengths (<319nm).

4) Photoelectric (photoplasmolysis) interactions

These are also nonthermal processes. They operate under a similar mechanism as in photomechanical interactions, except they demand even higher power density and extremely short pulse-duration ($<10^{-9}$ sec). For example, the excimer and Q-switched YAG lasers. They apply only to lasers that have high absorption in tissues. These interactions are characterised by the formation of plasma (a cloud of ionised particles, the overall charge of which is neutral and exists in a high-energy state) and acoustic shock waves on the target, which are responsible for the mechanical breaking and shattering of the tissue.

## 3.2. THE DETERMINING FACTORS OF LASER-TISSUE INTERACTIONS

The final outcome of the laser energy interaction with tissue will be determined by the spatial distribution of the deposited of radiant energy within tissues; it is a function of the depth of tissue penetration and the ratio of tissue absorption versus scattering. Only the absorbed component produces the desirable laser-tissue interactions and excess energy will be almost immediately converted into heat which will diffuse into the adjacent tissue (Apfelberg, 1987). The scattered component of the beam has energy density insufficient to induce desirable changes but instead causes detrimental localised tissue warming and collateral tissue damage by spatial redistribution of the beam energy beyond the extent of the thermally affected area.

The depth of tissue penetration and the extend of tissue absorption or scattering are in turn dictated by three dependent variables: 1) the inherent laser parameters; 2) the optical and physiological characteristics of the target tissue; and 3) the user-controlled elements (Fig. 3.3).

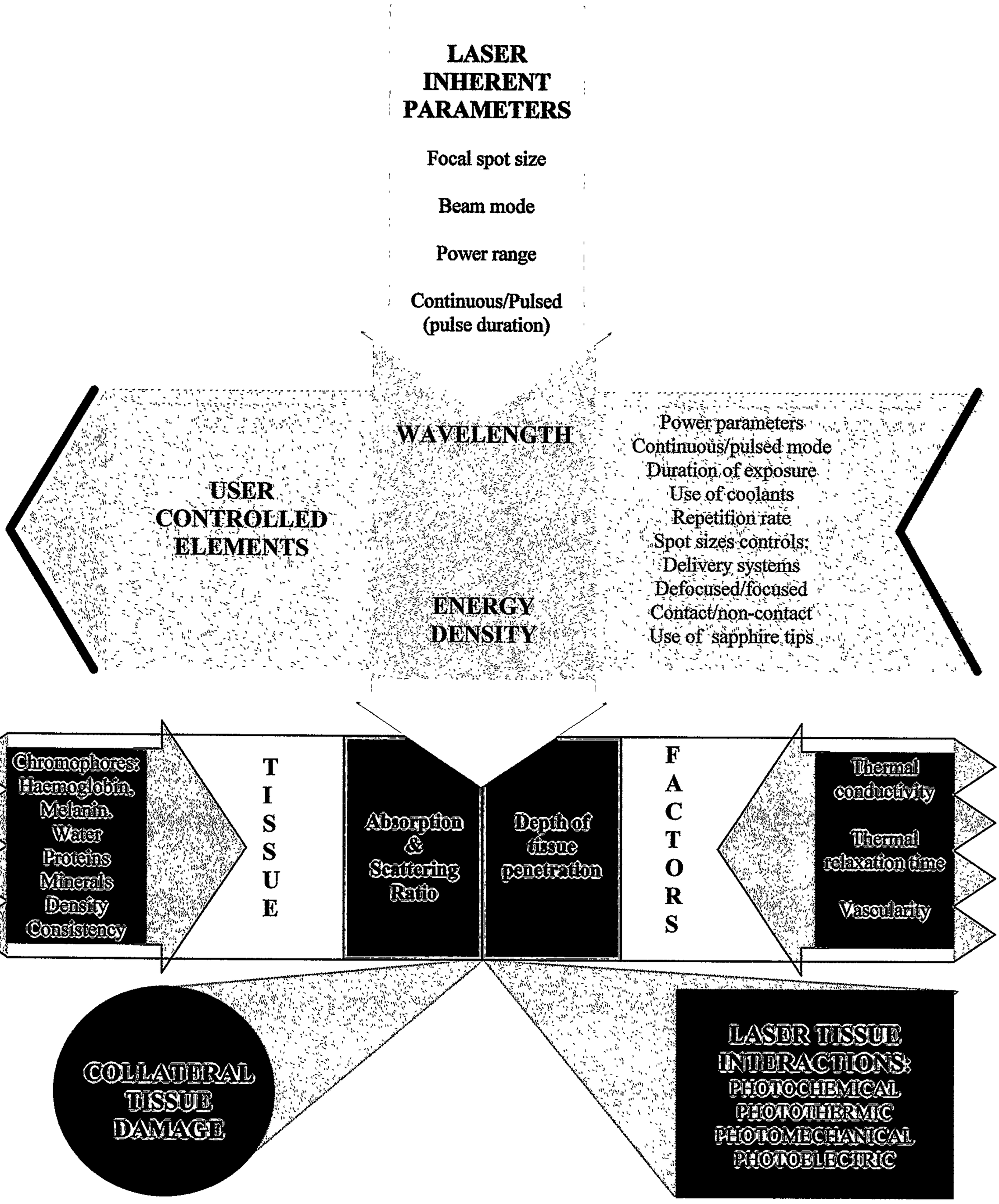



**Fig. 3.3 . Determining factors of laser tissue interactions: the laser inherent parameters, the user controlled elements and the optical and physiological characteristics of the target tissue.**

1) Laser inherent parameters

The laser inherent parameters (refer to section 2.3) are instrinic fixed elements that vary from system to system, which include the parental wavelength which determines the focal spot size and the absorption specificity of the radiant energy at the target tissue, the quality of the beam geometry, the built-in power capacity, and finally, the modes of operation (eg. continuous *vs* pulsed) which directs the time and rate of energy delivery to the target site.

2) Biological factors

The biological factors are more complex in nature. This complexity is due to the presence of variations in the chemical and molecular compositions found within tissues and between tissue types. Whether particular molecular and chemical components in tissue react with specific laser wavelength will be governed partly by the optical characteristic of various tissue components such as tissue chromophores (eg. haemoglobin, melanin), water content (75-90% ), proteins, minerals etc., and partly by the physiological properties of the tissues that include the tissue vascularity and thermal conductivity and thermal relaxation time of various tissue types. Additional factors such as the dentinal tubule density and the orientation of the tubules and the arrangement of enamel

prism can further influence the laser effect on dental hard tissues (Altshuler *et al*., 1991; Zijp and ten Bosch, 1991).

The thermal relaxation time measures the rate at which heat can diffuse to surrounding tissue. For continuous wave laser emissions, the energy is deposited within the tissue without allowance for thermal relaxation time and therefore has a linear relationship directly proportional to the total time of the exposure (Miserendino *et al*., 1989). A pulsed wave laser emission allows tissue cooling between each burst of energy and generates high peak power while maintaining the energy density constant to reach the desirable tissue temperature. In theory, chosing a pulsed laser system that has pulse duration less than the thermal relaxation time, and the pulse interval (no emission) longer than the relaxation time, can minimise and prevent any collateral tissue thermal injury. However, in practice, the thermal relaxation time can vary much between tissue components and tissue types.

3) User-controlled elements

The user-controlled elements are laser-parameters which can be changed and manipulated by the clinician to give precise control of the laser energy dosage at the target tissues. When this is coupled with the ability to choose the appropriate wavelength for the specific tissue, it will enable the operator to achieve the desired final clinical outcomes.

These parameters include the applied power density, energy density and the duty cycle (Fig. 2.10), the choice of mode of delivery (rigid mirror articular arm *vs* fiberoptic) system (Fig. 2.5), and focused *vs* defocused beam mode (Fig. 3.4). Other factors such as the use of some form of coolants (eg. air or water) during laser emission or the addition of a pigment to a non-pigmented area to increase the absorption of the pigment-specific wavelengths, will further moderate the final clinical manifestation of the laser-tissue interaction .

### 3.2.1. LASER ABSORPTION AND TISSUE PENETRATION SPECTRUM

The absorption spectrum describes the absorption coefficient (1/cm), for tissue chromophores such as haemoglobin and melanin, tissue water, and tissue protein at various wavelength range (from 0.1 to 10 micron) of the electromagnetic spectrum (Fig. 3.5). The absorption coefficient is a measure of how strongly light is absorbed in a particular material and it determines the absorption length (mm)of various lasers in human tissue. The absorption length is the distance the light travels in the tissue before it is 63% absorbed. The longer the wavelength the higher the

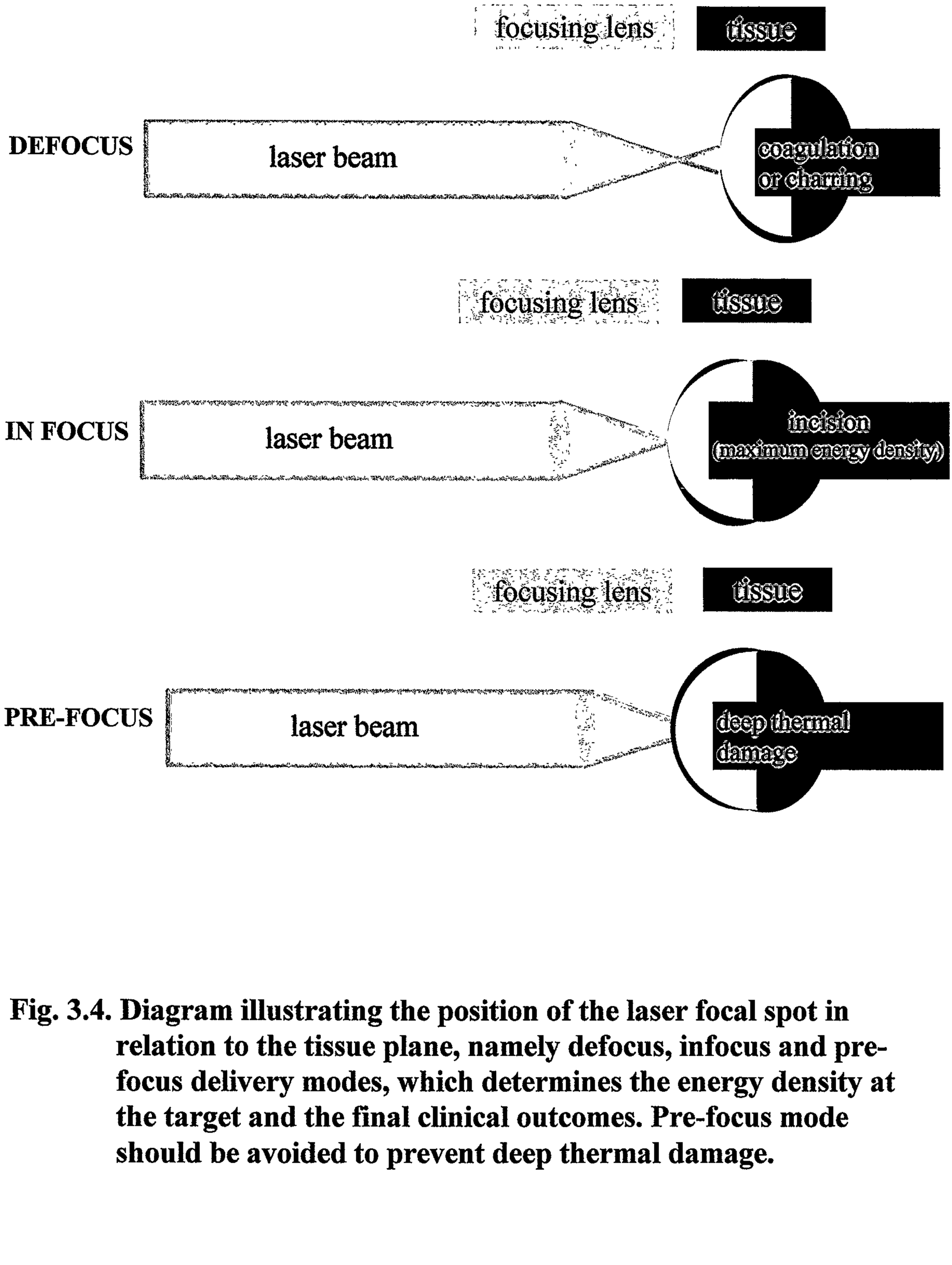



**Fig. 3.4. Diagram illustrating the position of the laser focal spot in relation to the tissue plane, namely defocus, infocus and pre-focus delivery modes, which determines the energy density at the target and the final clinical outcomes. Pre-focus mode should be avoided to prevent deep thermal damage.**

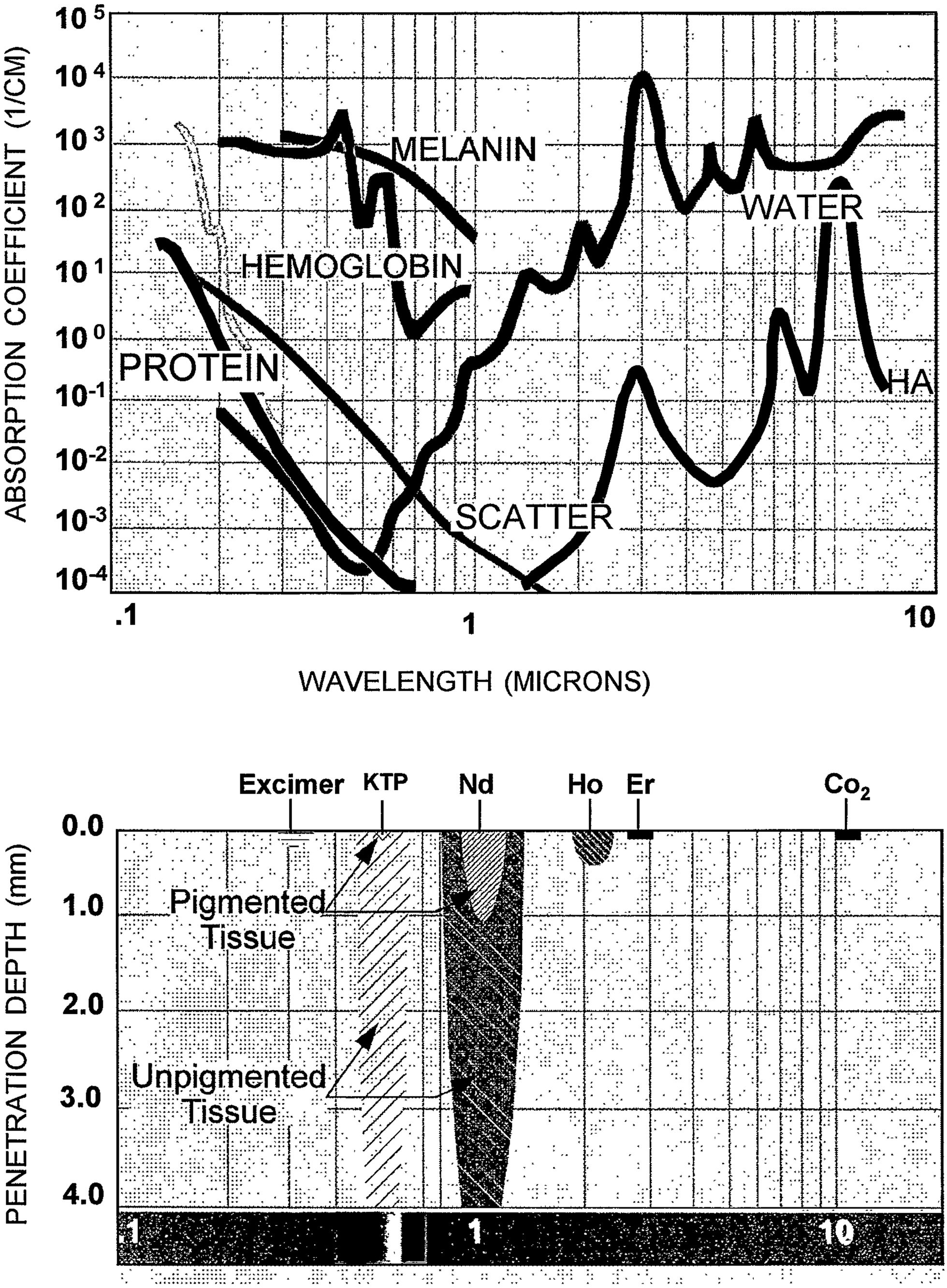



**Fig. 3.5 Laser absorption and tissue penetration spectrum**

**(modified from Coherent Medical Catalogue 1994)**

absorption of the beam and the lower the dependence on tissue colour (Karlan and Ossoff, 1984).

Water is transparent to visible light (wavelengths, 0.4-0.7micron) but is highly absorbed in the mid-infrared portion of the spectrum (2-10micron), with the highest peak at 3.0 micron (Er:YAG). Haemoglobin and melanin are strongly absorbed by visible light especially between 0.488-0.514 micron wavelengths. Consequently, the tissue penetration depth of the commonly used lasers in ascending order are Er:YAG, $CO_2$, Excimer, Ho:YAG, KTP & argon and Nd:YAG lasers. With argon and Nd:YAG wavelengths the penetration depth are further influenced by the tissue pigments.

Hydroxyapatite (HA) is absorbed strongly in the ultraviolet, mid-infrared and far-infrared wavelengths and there are peaks at wavelengths around 0.15, 3, 7, and 10 micron in the absorption curve of hydroxyapatite (Fig 3.5). It is however, a weak absorber of visible and near-infrared wavelengths (between 0.7 to 1.5 micron).

## 3.3. LASER USE IN DENTAL AND ORAL SURGERY

The most frequently used lasers in dentistry, either for basic research or clinical application, are the Carbon dioxide, the Erbium:YAG,

Holium:YAG, Neodymuim:YAG, the Diodes, the doubled neodymium:YAG (KTP), argon ion, and the Excimer lasers (Fig. 2.6).

### 3.3.1. Carbon dioxide laser ($CO_2$)

Carbon dioxide laser has emission wavelengths of 9.3, 9.6, 10.3 and 10.6 micron in the far infrared region of the electromagnetic spectrum. Therefore, it is invisible and an aiming visible beam is required for guiding.

At 10.6 micron, the $CO_2$ laser is highly absorbed by water. Since biological tissues, contain 75-90% water, the laser acts by vaporising tissues at its focal point, leaving adjacent tissue practically unaffected and enabling a fine haemostatic incision. The water absorption accounts for the shallow soft tissue absorption length (0.02 mm) of this laser.

The $CO_2$ laser often employs a mirror articulated arm delivery system (with a hollow wave-guided option) coupled with both continuous and pulsed modes (eg.supra-,ultra-pulse) of operation. The mode of tissue interaction is photothermic (photovaporisation, photoablation and photocoagulation).

The primary advantage of the $CO_2$ laser in dental and oral surgery is its ability to cut like a knife, vaporise soft tissue rapidly and precisely, and seal small vessels at the same time with minimal collateral thermal

damage. A defocused $CO_2$ laser can also accomplish precise surface ablation or vaporisation of superficial lesions with precision, as well as controlling the bleeding.

The disadvantages of $CO_2$ laser are its inability to obtain adequate tissue haemostasis where large vessels (>0.5 mm diameter) are involved, to transmit through clear aqueous tissues and the absence of a fiberoptic delivery system. For these reasons interest has been shown in the use of the Nd:YAG laser in oral surgery and other surgical specialities. Clinical reports (Tuffin and Carruth, 1980; Fisher and Frame, 1984; Rockwell, 1986) have shown that $CO_2$ laser-treated tissues heal with less postoperative pain, swelling and scarring than tissues treated with conventional surgical methods.

All lasers cause some thermal damage at the site of application and absorption, and an area of thermal necrosis will affect the progress of healing. However, this results in retarding the process of wound healing by at least 2-3 days beyond that of scalpel incisions (Hall, 1971). Tissue pre-cooling (Wakaki, 1979) or combined use of $CO_2$ and Nd:YAG lasers (Pinheiro, 1996) can reduce the thermal tissue injury without causing a delay in healing. Laser wounds that were deliberately contaminated with bacteria were found to have a higher degree of infection than those made by conventional scalpel (Mester *et al.*, 1971). However, early reports

suggested that infection in $CO_2$ laser-created wounds would be reduced due to the high temperature created at the site of application (Fidler *et al.* 1975).

Since the spectral output of the $CO_2$ laser overlaps the strong absorption bands of the phosphate and carbonate groups of hydroxyapatite and some dental ceramics, this laser has been under investigation for enamel fusion and ceramic welding.

Oral applications of $CO_2$ laser (Frentzen and Koort 1990) include proven procedures such as removal of precancerous, benign and malignant oral lesions, uvulopalatoplasty (cure snoring); and experimental procedures such as assistance with endo-periodontal surgery, wound sterilisation, ceramic welding, caries prevention, enamel etching, dentine hypersensitivity, pulp capping and apatite fusion (pit and fissure sealing).

### 3.3.2 Neodymium yttrium-aluminium garnet laser (Nd:YAG)

The Neodymium yttrium-aluminium garnet laser consists of a single crystal of yttrium-aluminium-garnet (YAG) doped with 1% to 3% of Neodymium (Nd) ions. It has emission wavelengths of 1.06, 1.32 and 1.44 micron in the near infrared region of the electromagnetic spectrum. Therefore, it is invisible and an aiming visible beam is required for guiding.

The 1.064 micron Nd:YAG laser energy is highly absorbed by black pigmented tissue and tissue proteins, and transmits through water, enamel and transparent materials. Its coefficient of absorption in water is 0.18/cm as compared with that of the $CO_2$ laser which is 770/cm (Fuller, 1987). This accounts for the long absorption length of 4-6mm (pigment-dependent) in soft tissue. As it is not absorbed effectively by human dental enamel (Spitzer and ten Bosch, 1975), the addition of a pigment on enamel will enhance the absorption (Hess, 1990).

The Nd:YAG laser is a photocoagulator, but is not designed for tissue vaporising (Fisher, 1993). It is an excellent wavelength for tissue coagulation, haemostasis and penetration, but gives little ablation effect. The Nd:YAG laser, because of its low coefficient of absorption in water, deeper tissue penetration (4-6mm) and scattering effect in soft tissue, allows for a comparatively large volume of tissue to be heated uniformly thus resulting in homogeneous coagulation and improved haemostasis. Therefore, it is well suited for the control of haemorrhage, to obtain tissue coagulation and to destroy small volumes of tissue by thermal devitalisation if desirable.

Due to its deep tissue penetration, the disadvantages related to the use of the Nd:YAG laser include inadvertent coagulation causing detrimental thermal damages to surrounding normal tissues, delayed tissue healing up

to 10 days (Luomanen *et al.*, 1994), subsurface vaporisation of target tissue (popcorn effect) and potential harmful effects on the dental pulp and the periodontal tissue.

In a recent development, a pulsed Nd:YAG laser system employing a xenon arc flash lamp with a pulse duration in the order of 100-800 microsec to generate laser pulses with peak power in the kilowatt range, has the ability to deliver the energy through an optical fibre. This pulsed Nd:YAG laser system allows cooling between bursts of energy to narrow the zone of thermal damage. White *et al.* (1991c), in a clinical study to compare laser and scalpel surgery in terms of pocket depth reduction, postoperative pain, inflammation and the treatment time, has reported that laser surgery gave less postoperative bleeding and decreased requirement for anaesthesia, or when compared to scalpel surgery. Synthetic sapphire laser tips have been developed for use with Nd:YAG laser. These laser tips vary the characteristics of the Nd:YAG laser to add vaporisation to its coagulative property (Joffe, 1985). Tissue pre-cooling with water and air during application of the laser has been proven to be effective in reducing the collateral thermal tissue injury (Levy, 1991).

Due to the fiberoptic capability, the Nd:YAG laser offers much better flexibility of application than that of the rigid articulating mirror delivery system. This also allows the laser beam to be delivered to hard-to-reach

areas intraorally, and to pass down the channel of an endoscope to be delivered to the temporomandibular joint as in arthroscopic surgery.

The major applications for the Nd:YAG laser include palliation of obstructing lesions (Fleischer, 1984) and photocoagulation of deep, extensive and vascular lesions such as haemangiomas, malignant tumours and deep seated lesions of the head and neck (Shapshay and Oliver,1984).

The use of the Nd:YAG laser for welding in prosthetic dentistry (Gorden and Denton, 1970; ven Benthen and Vahl, 1989) has shown clinical advantages over conventional methods. When compared with soldered dental alloys the grain structure, the tensile strength, the hardness and the corrosion behaviour of laser-welded alloys were shown to be better (Kasenbacher and Dielert, 1988; Sjogren *et al*., 1988).

Other oral applications (Frentzen and Koort 1990) include proven procedures such as caries vaporisation, enamel conditioning, dentine modification, assisted peri, endo-sterilization, assisted dental calculus removal, dentine desensitisation, induction of pulpal analgesia, gingival troughing and welding of dental alloys and titanium (power density dependent).

Other Nd:YAG emission wavelengths:

The 1.32 micron Nd:YAG laser is more uniformly absorbed by blood and water than the 1.064 micron counterpart, but is only available in continuous operating mode. Its major indications are the concurrent and synergistic use to induce regional hypothermia to supplement photochemotherapy (Suaasan *et al.*, 1985; Waldow *et al.*, 1985) and laser tissue welding (Niijma *et al.*, 1987).

The 1.44 micron Nd:YAG emission wavelength is a pulsed laser system and is highly absorbed in water, thus the tissue-interaction is similar to 2.1micron HO:YAG laser. It has been used in temporomandibular joint arthroscopic surgery.

Today in oral medicine and oral surgery, the Nd:YAG laser has become an important instrument in general laser surgery, complementing the $CO_2$ laser, which originally dominated the field. Its applications also extend to temporomandibular joint arthroscopic surgery, prosthodontics, endodontology, periodontology, conservative and preventive dentistry.

### 3.3.3 Holmium yttrium-aluminium garnet laser (Ho:YAG)

The Ho:YAG laser has an emission wavelength of 2.1 micron in the mid infrared region of the electromagnetic spectrum. Therefore, it is invisible and an aiming visible beam is required for guiding.

The 2.1 micron Ho:YAG laser is readily absorbed by water (1/16 that of $CO_2$ laser) and because of a relative short tissue absorption length of 0.5 mm (>$CO_2$), results in a zone of collateral tissue thermal necrosis of 80 micron (>$CO_2$ ).

It is a pulsed laser (pulse duration of 250 microsec) equipped with fiberoptic delivery system. The mode of tissue interactions are photovaporisation, photoablation and photocoagulation.

It cuts and ablates precisely and simultaneously coagulates tissues with the cutting profile similar to that of $CO_2$ laser using a waveguide. Because of its fiberoptic ability and transmission through saline medium, it is the preferred laser for temporomandibular joint arthroscopic surgery.

Oral applications (Frentzen and Koort 1990) include proven procedures such as soft tissue ablation, haemostasis, wound debridement, arthroscopic joint surgery and hard tissue ablation is only experimental.

### 3.3.4. Argon laser (Ar)

The argon laser has emission wavelengths range between 0.488-0.515 micron (0.488 micron blue singleline, 0.515 micron green singleline, 0.488 & 0.515 micron bluegreen multilines) in the visible region of the electromagnetic spectrum. Therefore, no aiming visible beam is required for guiding.

The blue line argon laser emission is responsible for photo-polymerisation of dental resins, whereas the green line portion is for soft tissue cutting, vaporisation and haemostasis.

argon laser energy is highly absorbed by haemoglobin with a moderate tissue absorption length of 1-2mm. It is also fiberoptic capable, and both continuous and pulsed (shuttering mechanisms within the resonator) operating modes are available.

Laser energy in the argon wavelengths is capable of rupturing red blood cells, damaging platelets, and initiating clotting cascades (Anderson and Parrish, 1981). Endothelial damage, protein denaturation and perivascular oedema likewise contribute to vessel closure (Kiefhaber *et al.*, 1977). Each laser, however, has limitations as to the size of vessels that can be controlled. The defocused $CO_2$ laser can coagulate vessels up to approximately 0.5mm; the argon laser up to 1mm; and the Nd:YAG laser up to 3-5mm (Fuller, 1983). In contrast to the $CO_2$ laser, rapid

vaporisation of tissue using the argon laser requires higher power densities, and is tissue pigment dependent.

The photo-absorption spectrum of light activated restorative materials has its peak around 480nm (Cook, 1982). The blue line argon laser emission allows photoactivation of dental composite resins (Severin and Maquon, 1989), but at a faster rate and gives better physical properties of the cured materials (Kelsey *et al.*, 1989; 1992).

Argon laser has remained the predominant laser type used for dental resin photopolymerisation and is the treatment of choice for vaporising superficial or cutaneous vascular lesions such as port wine haemangioma and telangiectasia (Cosman, 1980; Apfelberg *et al.*, 1981) because its energy is strongly absorbed by haemoglobin, melanin and cytochromes. The argon laser offers significant advantages in the treatment of smaller, superficial, moderately vascular lesions, especially where the fiberoptic capability and tissue transmission are preferred.

Other oral applications (Frentzen and Koort 1990) include proven procedures such as reduction in apatite demineralisation in conjunction with topical fluoride application (APF), caries detection (trans-illumination), tissue welding (vascular); and experimental procedures such as dentinal desensitisation, pulp capping, and sealing the root canal with resin materials (Potts and Petrou, 1990).

### 3.3.5. Potassium titanyl phosphate or frequency doubled Nd:YAG laser (KTP)

The Potassium titanyl phosphate laser has an emission wavelength of 0.532 micron, which is exactly half the 1.064 micron wavelength (or frequency-doubled) of the Nd:YAG laser, in the visible (green) region of the electromagnetic spectrum. Therefore, no aiming visible beam is required for guiding. The emission wavelength of the 1.064 micron Nd:YAG laser is halved by passing through a potassium titanyl phosphate (KTP) crystal.

This laser energy is highly absorbed by haemoglobin & melanin and the tissue absorption length is similar to the argon laser (1-2mm). It transmits through water and has fiberoptic capability. It is an excellent tool for cutting, vaporising and coagulating tissues and gives a dry operating field, like combining the benefits of Nd:YAG and $CO_2$ lasers. The primary disadvantage for the use of this laser is that it requires special plumbing and cooling facilities and is therefore costly to run.

Oral applications (Frentzen and Koort 1990) such as temporomandibular joint arthroscopic surgery and tissue welding are only experimental.

3.3.6. Erbium yttrium-aluminium garnet laser (Er:YAG)

The Er:YAG laser has an emission wavelength of 2.94 micron in the mid infrared region of the electromagnetic spectrum. Therefore, it is invisible and an aiming visible beam is required for guiding.

The 2.94 micron Er:YAG laser emits near the infrared peak of the water absorption curve and the $OH^-$ absorption peak of hydroxyapatite. Tissue absorption length and interactions are similar to the $CO_2$ laser, but with less coagulation effects. The zone of collateral tissue injury is minimal (20-40 micron) and the healing time is comparable to that of scalpel wound.

It is a pulsed system with pulse duration range between 200-400 microsec and it employs articulating arm delivery system (a fiberoptic is experimental).

It has been hypothesised that Erbium laser energy induces water vapour expansion inside the tissues causing internal microexplosions and fragmentation of dental hard tissues (Hibst and Keller, 1993) including enamel, dentine and bone, with minimal temperature rise and no signs of cracking, in the presence of water irrigation (Hibst and Keller, 1989a, 1993).

Hard tissue ablation is predominated by water absorption (Hibst and Keller, 1989), hence the ablation rate is faster in bone than dentine

(20% water) and enamel (10% water), and similarly higher in carious material than sound tissue (Rechmann *et al.*, 1993). The ablation rate is 2-3 times slower than the dental drill (Chao *et al.*, 1996) and is pulse energy, frequency and pulse duration dependent (Hibst and Keller, 1989).

It is also proven to be effective in removal all dental filling materials except gold & ceramic and is contraindicated for amalgam removal due to mercury plume hazard (Hibst and Keller, 1991).

Other oral applications (Frentzen and Koort 1990) include ablation of benign mucosal lesions and preprosthetic surgical procedures if no coagulation effect is desired. Laser procedures include enamel conditioning, pulp capping, assistance with perio-endo-surgery and assistance with osteotomy are only experimental.

3.3.7. Excimer laser (ArF, KrF, XeCl, XeF)

The Excimer is an excited dimer, a gas mixture of a noble and a halogen gas. The laser medium is excited via pulsed electrical current, to generate pulsed laser emission, with wavelengths range between 0.193-0.351 micron in the ultraviolet region of the electromagnetic spectrum. Ultraviolet laser energy is very strongly absorbed by tissue proteins and

lipids so penetration depth is small; of the order of a few micrometres. It is also fiberoptic capable (0.308micron XeCl).

Since the pulse duration of this laser energy is extremely short (20-60 ns), and is highly absorbed by the tissues when under extremely high power density, the photon energy is transformed into mechanical kinetic energy creating rapid expansions, rupturing tissue peptide and nuclear bonds, and resulting in microexplosion of tissues, in a nonthermal manner, accompanied with plasma and acoustic short wave formation. The predominant disadvantages of this laser include shock wave and plasma formation which may cause collateral tissue damage, and might have mutagenic potential (0.248micron KrF). It is also expensive to set up and to run. Otherwise, it is extremely effective in hard tissue ablation without thermal collateral tissue damage if the energy density used is above the photomechanical threshold.

It is also proficient for debridement of degenerate articular cartilage and meniscus as used in temporomandibular joint arthroscopic surgery. Other oral applications (Frentzen and Koort 1990) including soft and hard tissue ablations, enamel conditioning and assist endodontology, are experimental.

### 3.3.8. Lower level laser irradiation or soft laser (LLLI)

Lower level laser irradiation utilises large portions of the visible and infrared spectrums of the electromagnetic spectrum. The most commonly used wavelengths are 0.633 micron He:Ne, 0.904 micron GaAs and 0.830 micron GaAlAs.

Soft laser therapy employs lower level irradiation, with an average power range of 10-90mw, 1-4.3 J/cm$^2$ dosage per site, continuous or pulsed (1-4000Hz), with delivery techniques and irradiation site variations, and consequently does not produce any significant thermal effect (temperature elevation: 0.5-0.75 °C) in the tissues.

Its effect is predominantly biostimulation (or inhibition) which includes modulating cellular function (Karu, 1988) and fibroblast proliferation (Pourreau-Schneider *et al.*, 1989), lessening pain (Anderson *et al.*, 1973; Walker, 1983; Rochkind *et al.*, 1987) and accelerating healing (Mester *et al.*, 1971) of soft tissue injury in a nonthermal, wavelength dependent manner. However, some controlled studies of wound healing utilising low power helium neon lasers have not indicated acceleration of wound healing (Hunter *et al.*, 1984 ).

The advantages of minimal collateral thermal damage, minimal eye safety hazard, low set-up and running cost and portability, makes the soft laser superior to other laser systems for low level laser therapy. It is also

claimed to reduce inflammation, promote healing and lessen pain. The other oral applications (Frentzen and Koort 1990) of soft lasers include tissue welding (with photosensitiser), treatment of musculoskeletal injury, dentine desensitisation, and assistance with pulpal diagnosis.

Table 3.2. summarises various laser types, with their corresponding basic laser components, as well as the optical absorption properties which include the tissue absorbing elements, the absorption length and the predominant type of tissue interaction. Table 3.3., lists the clinical applications (including experimental) of various lasers in different specialities in dentistry.

A fast, precise, pain-free, dry field operation and a compact delivery system providing access to hard to reach areas, enable lasers to offer advantages over conventional methods in certain dental and oral surgery procedures. Furthermore, the ability of some lasers to photo-polymerise dental resin materials at a faster rate, to weld dental alloy and ceramic, to condition tooth surfaces for better bonding and resistance to demineralisation and hard tissue ablation has provided fertile ground for clinical applications and research in various fields of dentistry. The future of laser in dentistry is in the hands of both scientists and clinicians and surely will be as bright as the sun.

# Table 3.2. Laser use in dental and oral surgery. Modified from Collins (1989).

| Laser Type | Predominant Laser Wavelength (nm) | Active Medium | Excitation Source | Operating Mode | Delivery System | Tissue Absorbing Elements | Absorption Length (mm) | Tissue Interactions |
|---|---|---|---|---|---|---|---|---|
| $CO_2$ | 10600 | $CO_2$, $N_2$, He gas mixture | DC/RF discharge | CW/ Pulsed/ Repetitive | Mirror/ Hollow guide | Water | 0.02 | Photothermic |
| Er:YAG | 2940 | Erbium in yttriumaluminum garnet | Flash-lamp | Pulsed | Mirror/ Fiber (development) | Water/ Hydroxy-apetite | 0.01 | Photothermic-mechanical |
| HO:YAG | 2140 | Holmium in yttriumaluminum garnet | Flash-lamp | CW/ Pulsed/ Repetitive | Fiber | Water (1/16 of $CO_2$) | 0.5 | Photothermic Photomechanical |
| Nd:YAG | 1064 | Neodymium in yttriumaluminum garnet | Flash-lamp | CW/ Pulsed | Fiber | Dark pigments | 4-6 | Photothermic Photomechanical |
| Nd:YAG | 1440 | Neodymium in yttriumaluminum garnet | Flash-lamp | Pulsed | Fiber | Water | 0.5 | Photothermic |
| HeNe<br>GaAlAs<br>GaAs | 633<br>830<br>904 | Helium Gallium-Arsenide Gallium-Arsenide | DC discharge | CW | Fiber/ Direct | Dark pigments | 2-3 | Photochemical Photothermic |
| Frequency doubled Nd:YAG (KTP) | 532 | Neodymium in yttriumaluminum garnet | Flash-lamp | CW | Fiber | Melanin/ Haemoglobin | 2-3 | Photothermic |
| Argon: | | | | | | | | |
| Green | 514 | Argon ions | DC discharge | CW/ Pulsed | Fiber | Haemoglobin | 1-2 | Photothermic |
| Blue | 488 | | | | | | 0.5-0.8 | Photochemical |
| Excimers: | | | | | | | | |
| XeF | 351 | Xenon fluoride | | | | | 0.2 | Photomechanical |
| XeCl | 308 | Xenon chloride | DC discharge | Pulsed/ Repetitive | Fiber/ Direct | Proteins/ Water | 0.1 | Photoelectric |
| KrF | 248 | Krypton fluoride | | | | | < 0.1 | |
| ArF | 193 | Argon fluoride | | | | | < 0.1 | |

# Table 3.3. Clinical applications of lasers in dental and oral surgery. Modified from Frentzen and Koort (1990).

| APPLICATIONS | LASER TYPES |
|---|---|
| **ORAL AND MAXILLFACIAL SURGERY:** | |
| Removel of benign, precancerous or malignant lesions. | $CO_2$ |
| Treatment of haemangiomes or deep site lesions | Nd:YAG |
| Tissue coagulation and haemostasis | Nd:YAG (3-5mm diameter vessel), Ar (1mm diameter vessel), $CO_2$ (0.5mm diameter vessel) |
| Removel benign mucosal lesions | Er:YAG, $CO_2$, Ar, Excimer |
| Tissue welding | 1.32micronNd:YAG, Ar |
| TMJ arthroscopic surgery | 1.44micronNd:YAG, Ho:YAG, $CO_2$, KTP, Excimer |
| Uvalopalatoplasty (cure snoring) | $CO_2$ |
| **CONSERVATIVE DENTISTRY:** | |
| Preventive dentistry - reduction in apatite demineralization with topical fluoride (APF) | Ar, Nd:YAG, $CO_2$ |
| -apatite fusion (fissure sealing) | Nd:YAG, $CO_2$ |
| Caries treatment | $CO_2$, Nd:YAG (with photoinitiator), Er:YAG, Excimer, GaAlAs (with indocyanine green dye) |
| Composite resin photopolymerization | 0.488micronAr, HeCd, dye |
| Tooth surface conditioning- enamel etching (enhance resin bonding), fusing apatite, dentinal modification | Excimer (ArF), Nd:YAG(with photoinitiator), $CO_2$, Er:YAG |
| Dentinal desensitization | Ar, $CO_2$, Ho:YAG, Nd:YAG, GaAlAs, GaAs |
| Enamel ablation | Excimer, Er:YAG, TEA-$CO_2$, 9.3micron$CO_2$ |
| Dental materials welding- dental alloy | Nd:YAG |
| - dental ceramic | $CO_2$ |
| **ENDODONTICS:** | |
| Pulp capping, pulpotomy or pulpectomy | 0.514micronAr, $CO_2$, Nd:YAG, Er:YAG |
| Debridement or sterilization of root canal | Nd:YAG, $CO_2$, Excimer |
| Apicectomy or apical fusion | CO2, 0.514micronAr ,Nd:YAG, Er:YAG, Excimer |
| Obturation of root canal | 0.488 micronAr (with resins) |
| Sterilization of endodotic instruments | $CO_2$, Nd:YAG, Ar |
| **PERIODONTICS:** | |
| Assist calculus removel | Nd:YAG |
| Excision of gingival tissues | $CO_2$, Ho:YAG, Er:YAG, Nd:YAG, Ar, Excimer |
| **STIMULATION OF WOUND HEALING** | HeNe, GaAlAs, GaAs |
| **ANAESTHESIA:** | |
| Inflammatory | HeNe,GaAlAs |
| Pulpal | Nd:YAG(pulsed) |
| **DIGNOSIS:** | |
| Laser doppler flowmetry | HeNe, Diodes |
| Caries detection | Ar (transillumination), $CO_2$, Laser-induced fluorescence or phosphorescent re-emission |

## 3.4. LASER SAFETY

The use of lasers in dental and oral surgery is rapidly expanding and impinges upon many disciplines from the clinical laboratory to the office practice to the operating room. It is essential that users of these powerful tools are knowledgeable of their potential hazards and the means to protect patients and personnel against injuries of undesired effects.

Lasers are classified (Table 3.4.) according to the amount of laser energy that is accessible to the skin or eyes in normal use. Laser accidents most commonly result from the misdirection of the laser beam. Direct or reflected radiation can burn skin, hair or the cornea, or cause permanent retinal damage.

The smoke or plume produced by the laser is an area of concern to operating personnel. Two areas should be monitored regarding the content of the laser plume. Laser plume consists of water, carbonised particles, and gives off an offensive odour. One area of concern is the size of the particulate matter; when it is less than one micron, it can easily be deposited in the alveoli of the lungs causing a number of problems relating to the respiratory system. The other area of concern is the viability of the cells in the plume that may be a vector for transmitting infectious materials such as the human papilloma virus (Becker *et al.*, 1987). It has been shown to be mutagenic in *Salmonella typhimurium*

(Mihashi *et al.*, 1981) which is similar to cigarette smoke. The plume cell viability is determined by the power density of the laser beam on the tissue (Walker *et al.*, 1986; Byrne *et al.*, 1987) and the absorption length of the type of irradiated wavelength. For example, the smoke plume from malignant tissue vaporised by a continuous $CO_2$ lase does not contain viable cells (Bellina *et al.*, 1982). In contrast, plume from a pulsed Nd:YAG laser may contain viable cells (Voorhies *et al.*, 1984). Thus, removal of the plume of smoke is essential.

Other hazards includes fires or explosions during laser procedures through the ignition of various substances (for example the use of highly inflammable anaesthetic agents) in the surgical field, and those associated with the operation of equipment with high voltage supplies and water cooling systems. Included below is the summary of the safety considerations listed by the Australian/New Zealand Standard AS/NZS 4173:1994 on laser safety for class IV lasers.

1) Direct supervision is required by an individual knowledgeable in laser technology and safety.

2) Appropriate warning signs are posted: the word "DANGER" and the phase " Laser Radiation; avoid eye or skin exposure to direct or

**Table 3.4. Laser hazard classification according to American National Standard Institue (ANSI)**

| Class | Description |
|---|---|
| **I** | Low-powered lasers that are safe to view |
| **IIa** | Low-powered visible lasers that are hazardous only when viewed directly for longer than 1,000 secs |
| **II** | Low-powered visible lasers that are hazardous when viewed for longer than 0.25 secs |
| **IIIa** | Medium-powered lasers or systems that are normally not hazardous if viewed for less than 0.25secs without magnifying optics |
| **IIIb** | Medium-powered lasers (0.5 W maximum) that can be hazardous if viewed directly |
| **IV** | High-powered lasers ( > 0.5 W ) that produce ocular, skin, and fire hazards |

scattered radiation" should be prominently displayed. If an infrared wavelength is to be used the word "Invisible" should also be included in the warning sign.

3) Use only diffusely reflective materials, such as dull-surfaced instruments, in or near the beam path when feasible.

4) Any potentially hazardous beam should be terminated in a beam stop of an appropriate material that is highly absorbent. For example, In the case of $CO_2$ laser, wet gauze is used, placed near the target tissue to absorb any scattered energy. All specular reflective material in or nearby the beam path should be removed.

5) Cover or restrict all windows, doorways and open portals in areas with levels above the ocular maximum permissible exposure.

6) In an oxygen-enriched atmospheres, the use of a ClassIV laser should be avoid. A fire extinguisher should be kept in the room or nearby.

7) All personnel include staffs, patients and spectators should wear some forms of impenetrable barrier eyeshields with side panels that are appropriate for the particular type of laser in use.

8) Use an appropriate smoke evacuation system with filters for the amount of plume produced.

9) Always wear masks and gloves and protective clothing during the laser intervention.

Laser accidents are caused by carelessness and inattentiveness of the safety guidelines. As the progress of laser technology continues, the potential for laser hazards also increases. The practice of laser safety is the key in prevention of any possible dangers.

# 4. Neodymium:YAG (1064nm) laser effects on dental hard tissues and the pulp.

## 4.1. Neodymium:YAG laser effects on dental enamel.

The Carbon dioxide ($CO_2$) and Neodymium (Nd):YAG laser energies can induce structural, phase and physicochemical changes in calcified tissues (Yamamoto and Ooya, 1974; Kuroda and Fowler, 1984; Fowler and Kuroda, 1986; Hèss, 1990). These interactions are characterised by a thermal mechanism, which is largely dependent on wavelength specificity (Featherstone and Nelson, 1987), the energy density, the interaction time and the organic and mineral composition of the calcified tissue (Frentzen and Koort, 1990).

At relatively high laser energy fluence density of 200-10000 J/cm$^2$, continuous wave $CO_2$ laser energy can cause cratering and melting of enamel (Lhuisser, 1980; Bonin *et al.*, 1985), with subsequent recrystallisation (Lobene *et al.*, 1968; Kantola *et al.*, 1973), which leads to the formation of calcium phosphate minerals (alpha tri-calcium phosphate) other than apatite (Brune, 1980; Kuroda and Nakahara, 1984).