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Laser Principles

Inja Bogdan Allemann[a] · Joely Kaufman[b]

[a]Klinik für Plastische Chirurgie und Handchirurgie, Universitätsspital Zürich, Zürich, Switzerland;
[b]University of Miami Miller School of Medicine and Private Practice Coral Gables, Miami, Fla., USA

Abstract

Since the construction of the first laser in the 1960s, the role that lasers play in various medical specialities, including dermatology, has steadily increased. However, within the last 2 decades, the technological advances and the use of lasers in the field of dermatology have virtually exploded. Many treatments have only become possible with the use of lasers. Especially in aesthetic medicine, lasers are an essential tool in the treatment armamentarium. Due to better research and understanding of the physics of light and skin, there is now a wide and increasing array of different lasers and devices to choose from. The proper laser selection for each indication and treatment requires a profound understanding of laser physics and the basic laser principles. Understanding these principles will allow the laser operator to obtain better results and help avoid complications. This chapter will give an in-depth overview of the physical principles relevant in cutaneous laser surgery.

It was in 1959 that the first laser was invented and developed by the physicist T.H. Maiman at the Hughes Research Laboratory in Malibu, California [1]. He used a flashlamp-pumped ruby crystal, which produced red light with a wavelength of 694 nm. The clinical use of lasers in dermatology started in 1963, when Dr. Leon Goldman used the ruby laser for various dermatological conditions [2, 3]. In the following years, the continuous-wave argon, CO_2, and Nd:YAG lasers followed. In 1983, the theory of selective photothermolysis, proposed by R. Anderson and J.A. Parrish revolutionized the use of lasers in dermatology [4]. Based on this theory, pulsed lasers were able to specifically and selectively target and destroy structures within the skin, without damage to surrounding tissues. This theory can be regarded as a milestone in cutaneous laser treatments. Finally, a recent technological advance was the introduction of the concept of fractional photothermolysis in 2003, which led to further development and expansion of laser resurfacing in dermatology and will also be discussed below [5].

Spontaneous and Stimulated Emission of Radiation

The word laser is an acronym for 'light amplification by the stimulated emission of radiation'. In contrast to sunlight for example, which is emitted spontaneously, laser light is emitted by stimulated emission. The concept of stimulated light emission, which the generation of laser light is essentially based on, was originally conceived by Albert

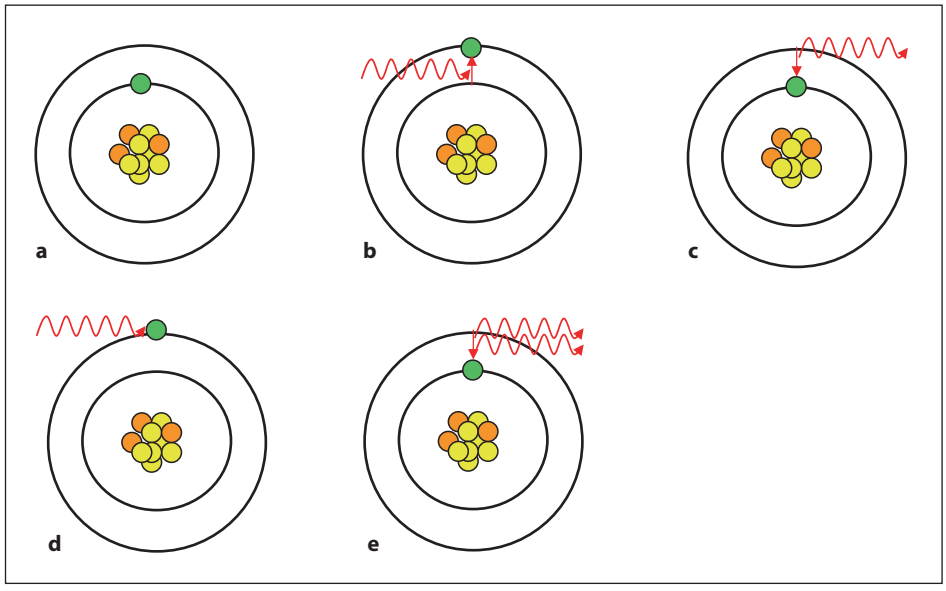

Fig. 1. Spontaneous and stimulated emission. **a** Resting state of the atom, with the electron in a low-energy orbiting position. **b** Absorption of a photon elevates the electron into an excited state. **c** The unstable excited electron falls back into the lower-energy resting state and releases the photon of energy. **d** If an excited electron absorbs another photon of energy, (**e**) it releases two photons with the same energy, direction, and wavelength when falling back into its resting state.

Einstein [6] in 1917, clearly long before the first laser was invented.

Spontaneous Emission of Radiation

Einstein based his theory on the original model of the atom, set forth by Neils Bohr in 1913.

Bohr's model explained that atoms are composed of a nucleus, containing positively charged protons and neutral neutrons. Negatively charged electrons orbit the nucleus, much like planets orbit the sun. The electrical attraction between the positive and the negative holds the entire complex together. Electrons that surround a molecule or atom orbit in a stable resting state. This state is the state of lowest energy and is close to the nucleus. If an electron absorbs energy, in the form of a photon (particle of light), it will jump to a higher energy level at a position further from the nucleus and will then be in an excited state. This state is a far less stable position. Atoms can only stay in an excited state for a short period of time and as the excited electrons return to their resting state, they release the energy previously absorbed in form of photons. This emission of photons is random and occurs in all directions. When it is random and occurs on a regular basis, it is referred to as spontaneous emission. In nature, spontaneous emission is dominant and the majority of electrons are in the resting state. Spontaneous emission is the source of sunlight and virtually any light observed in nature. However, this process of excitation followed by de-excitation and release of photons can also be induced, as proposed by Einstein in the early 1900s.

Stimulated Emission of Radiation

For this to happen, an already excited electron has to collide with yet another photon with the

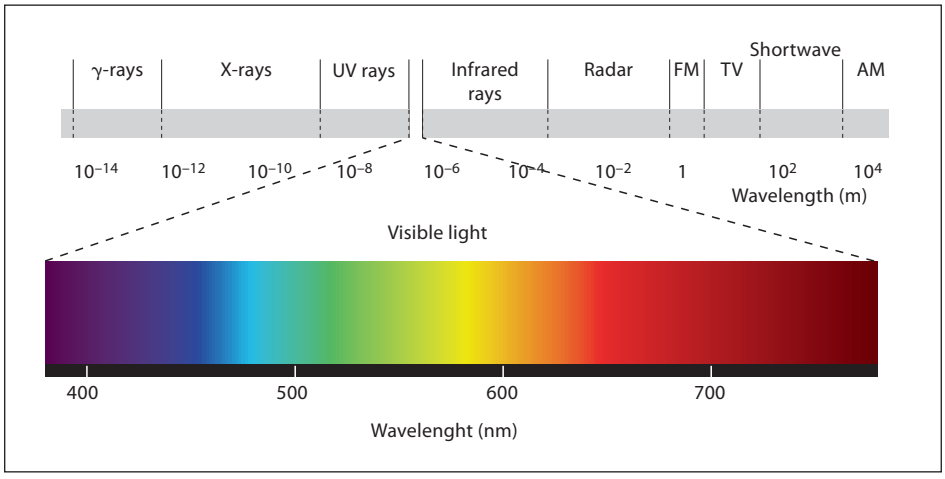

Fig. 2. The electromagnetic spectrum.

proper energy, which will lead to the emission of two photons when the electron returns to resting state. Importantly, these two photons will contain the same energy, frequency, and direction, which sets the basis for the unique properties of laser light discussed below (fig. 1). Finally, such emitted photons can again stimulate the emission of further photons. Stimulated emission needs a source of energy to induce the change. In the case of lasers, we have several sources of stimulation, referred to as the 'pump' portion of the device. This increasing stimulation of photon emission will ultimately lead to an environment where there are proportionately higher numbers of atoms in the excited state as compared to the resting state. This situation is called 'population inversion' and is a crucial prerequisite for the generation of laser light within an optical chamber.

Properties of Laser Light

Light is a form of energy, all forms of light are represented in the electromagnetic spectrum. Thus, laser light represents a part of the electromagnetic spectrum of energy. The electromagnetic spectrum encompasses all energy sources that travel through space in the form of waves, and ranges from the short wavelengths of X-rays and γ-rays to the long wavelengths of microwaves and radio waves. Most lasers that are used in medicine and dermatology generate light within the visible (400–760 nm), near-infrared (760–1,400 nm), mid-infrared (1.4–3 μm), infrared (>3 μm), and rarely the ultraviolet range (200–400 nm) of the electromagnetic spectrum (fig. 2). As the name states, laser light within the visible spectrum of the electromagnetic spectrum creates visible beams of various colors, depending on the respective wavelength. For example, the ruby laser at 694 nm generates red light, whereas the KTP laser at 532 nm generates green light. Lasers within the infrared section of the electromagnetic spectrum, such as YAG and CO_2, generate invisible beams of laser light.

Laser light has three main and unique characteristics that differentiate it from other light sources and represent the basis for its therapeutic use and effectiveness: monochromaticity, coherence, and collimation (fig. 3) [7].

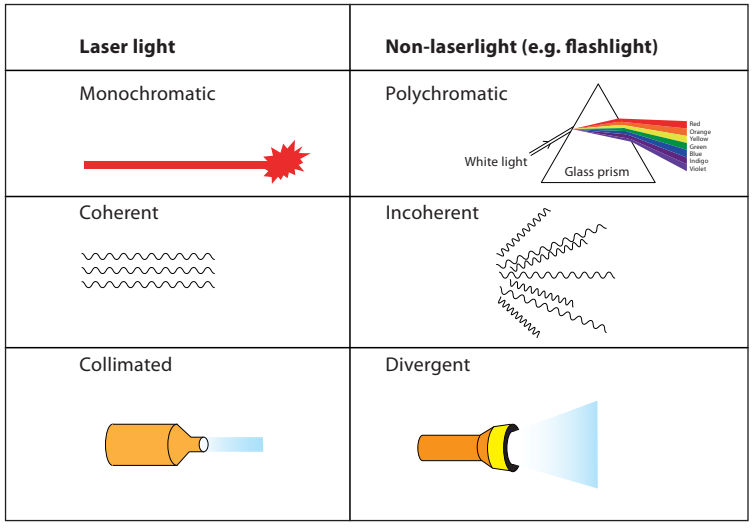

Fig. 3. Laser properties.

Monochromaticity
In contrast to sunlight, which encompasses a wide spectrum of wavelengths, laser light is monochromatic, which means it emits light of only one clearly defined single wavelength or a very narrow band of wavelengths. Laser light that is shone through a prism produces the same color. White light however that is shone through a prism produces the entire spectrum of colors. In the visible spectrum, every wavelength has its own color. The respective wavelength is defined by the lasing medium used in the laser, which will be discussed below.

Coherence
Laser light is also required to be coherent, which means that the light waves are in phase with respect to space and time. The coherence of laser light is based on the process of stimulated emission, where only light in the same phase and direction can be emitted.

Collimation
Finally, laser light is collimated. As the light waves are parallel, the laser beam has a defined narrow beam diameter with no divergence, which virtually does not increase even with increasing distance. This is best illustrated by a laser pointer, whose beam stays practically the same no matter at which distance. This fact allows laser light to not loose any of its energy even after travelling over longer distances. This is different from non-collimated light emitted by a flashlight, which with increasing distance, has an increasing divergent beam diameter and a decreasing intensity.

High Intensity
Finally, another crucial feature of laser light is that it can emit light with very high intensity.

The wavelength of lasers is usually measured in nanometers. One nanometer is 10^{-6} mm, which is 0.000001 mm.

Non-Laser Light Sources

Non-laser light sources, such as for example intense pulsed light (IPL) devices, which nowadays are used as frequently as lasers, have different features than true lasers. IPL, or broadband light

devices use flashlamps and computer-controlled capacitor banks to generate pulsed polychromatic high-intensity light. In contrary to lasers, IPL devices emit light of many wavelengths, encompassing a wide range usually from 500 to 1,300 nm within the electromagnetic spectrum. Many times, cutoff filters are built in, in order to tailor the wavelength range more precisely to the target structure, such as vascular or pigmented ones [8]. Thus, for example vascular lesions can be targeted with an IPL device with a wavelength range of 500–670 and 870–1,400 nm, whereas pigmented lesions would be targeted with another device of 525–1,200 nm. The clinical effect of IPL on the target tissue is photothermal, which means generation of heat within the target tissue with subsequent destruction the target structure.

For such photothermolysis to occur, monochromaticity (one single wavelength or a very narrow range of wavelengths) is not necessarily required, as the main chromophores in the skin (melanin, hemoglobin, water) show broad absorption spectra.

As IPL devices employ multiple wavelengths at the same time, various chromophores, such as for example melanin and hemoglobin can be targeted with the same light exposure. This can be a desirable effect, when for example performing the treatment of photoaged skin, which is composed of pigmentary and vascular alterations. Clearly, in other clinical indications, such as for example the treatment of facial redness, where only one chromophore is targeted, the other one (melanin) might be competing, actually hindering the treatment.

IPL devices have pulse durations within the millisecond range which can be set according to the clinical indication. There are indications, such as photoepilation, where pulse durations within the millisecond range or longer are needed. However, as with lasers, ideally the pulse duration should be lower than the thermal relaxation time (TRT) of the target structure to prevent unselective damage to the surrounding tissue [9]. The usually large spot sizes represent another advantage of IPL devices, again especially useful in photoepilation.

There are numerous clinical indications that can be treated with IPL devices such as vascular lesions, pigmented lesions, photoaged skin, excess hair growth, and acne vulgaris.

A detailed description of IPL technology can be found in a dedicated chapter [Schoenewolf et al., pp. 166–172].

Fractional Technology

The novel concept of fractional photothermolysis was introduced to the market by Dieter Manstein and Rox Anderson in the year 2003 [10]. It was an answer to the need for effective yet low-risk resurfacing methods. Unlike conventional ablative and non-ablative lasers, fractional ablative and non-ablative lasers treat only a fraction of the skin, leaving up to a maximum of 95% of the skin uninvolved – thus the name 'fractional'. This is achieved by inducing microscopic small three-dimensional zones of thermal damage or ablation, surrounded by undamaged tissue allowing for rapid epidermal repair.

The target chromophore for fractional photothermolysis devices is water. Hence, various water-containing structures at differing depths throughout the skin, such as, epidermal keratinocytes, collagen, and blood vessels can be targeted resulting in a wide array of clinical indications [11].

Based on the wavelength's affinity for water, fractional technologies can be divided into two main categories. Those with wavelengths that are highly absorbed by water are termed 'ablative', while those wavelengths that are less avidly absorbed by water are termed non-ablative. This novel technology will be described in detail in chapter on fractional photothermolysis [Bogdan Allemann and Kaufman, pp. 56–66].

Construction of Lasers

Extremely simplified, a laser consists of a laser chamber containing the lasing medium, two mirrors, and an excitation source, i.e. the external pump [12]. The chamber is a highly reflective optical cavity, with two parallel mirrors on each side, one of which is fully reflective and the other partially reflective. Within this chamber is the lasing medium, which can be a solid, liquid, or gas (table 1). It is the lasing medium which defines the wavelength emitted by a laser. Examples of solid lasing media are ruby, alexandrite, Er:YAG, Nd:YAG, and diode. A liquid lasing medium used is dye, and gaseous lasing media are CO_2, argon, and excimer. The lasing medium represents the supply of electrons needed to produce a laser beam by stimulated emission of radiation and defines the wavelength of the laser light emitted. Many lasers are named after the lasing medium contained within the laser. The photons within the lasing medium in the chamber need to be externally stimulated, in order to be amplified and generate the laser light. This external energy source is referred to as the pump or pumping system. Pumps can be any form of energy source, including a high-powered flashlamp, electrical current or another laser. Within the lasing medium, the energy gets absorbed by the atoms, which with stimulation get excited and then release their energy in the form of photons. The photons within the chamber are reflected back and forth from one mirror to the other one, colliding with other atoms and hence stimulating further emission within the same axis. The more excited atoms in the medium (= population of atoms) there are, the more photons get produced. Eventually, there will be more excited atoms than resting ones, which is called 'population inversion'. Remember, in a normal population of atoms, the majority resides in a resting state. One mirror within the chamber then becomes partially reflective, which allows for some of those stored photons within the chamber to get emitted as a beam of light. In fact, only a fraction of light is emitted from the laser cavity, while the rest remains in the cavity to maintain the lasing process, which continues, as long as the pump keeps exciting the atoms in the lasing medium [13]. Of the light emitted, only parallel waves of the same wavelength can be emitted, which sets the basis for the collimated, monochromatic nature of laser light. Depending on whether the pump is continuous or pumped, a continuous wave of light or a pulsed wave of light gets emitted.

Once the light is emitted from the chamber through the partially reflective mirror, it enters the delivery system which will ultimately transmit the light to the handpiece. Some delivery systems are fragile articulated arms that contain mirrors reflecting the light, other delivery systems are build from fiberoptic cables. The handpiece finally focuses the light onto the skin (fig. 4).

Table 1. Lasing media and lasers

Laser type	Lasing media
Liquid	dye
Gas	CO_2
	argon
	excimer
Solid	ruby
	alexandrite
	Er:YAG
	Nd:YAG
	diode

The lasing medium represents the supply of electrons needed to produce a laser beam by stimulated emission of radiation and defines the wavelength of the laser light emitted

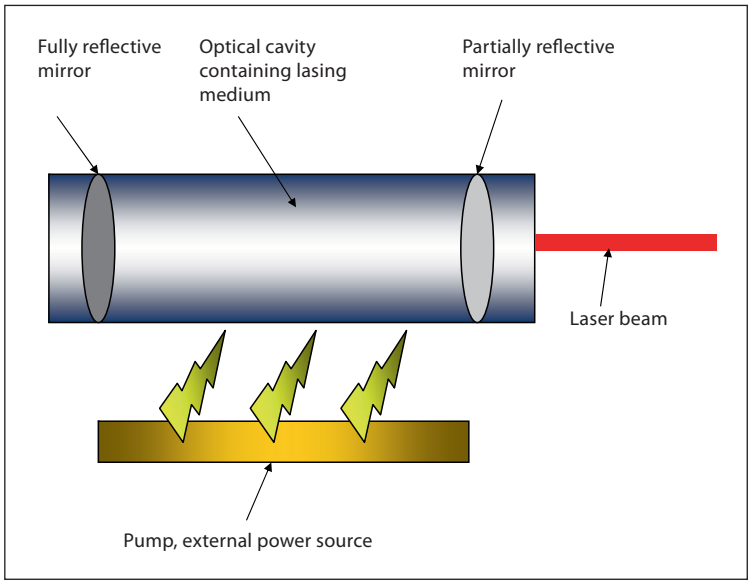

Fig. 4. Construction of lasers.

Tissue Optics

When laser light reaches the skin, it can interact with the tissue in four different ways: it can be absorbed, reflected, scattered, or transmitted (fig. 5). Most times, a combination of various interactions, each at different degrees, takes place at the same time [14].

Absorption

When using lasers for therapeutic purposes, the goal is for the laser light to be absorbed by a specific target. The Grothus-Draper law of photobiology states that in order to have a biological effect in the tissue, light must be absorbed by the target tissue. It is only this light which is absorbed that does real work on the tissue. When a target molecule absorbs a photon, the entire energy of the photon is transferred to the target molecule. The specific light-absorbing targets are known as chromophores. The three main chromophores in the skin are melanin, (oxy)hemoglobin, and water. Tattoo ink is the main external chromophore of importance in laser dermatology. The amount of light that gets absorbed by the specific chromophore depends on the wavelength used and whether it corresponds to the specific absorption spectrum of the respective chromophore.

Light that is not absorbed can either be scattered, transmitted, or reflected – all of which have no true biological effects [15]

Reflection

Roughly 4–6% of the light usually gets reflected when it hits the skin at a 90 degree angle. Reflection of the light mostly takes place at the stratum corneum and is the reason why protective eyeshields should be worn at all times when working with lasers. Reflection of light can be minimized by applying the incident laser beam perpendicularly to the tissue surface. With an increased angle of light incidence, the rate of reflection increases above the aforementioned 4–6% [16]. Furthermore, dry or scaly skin reflects even more light, which again can be minimized by applying a thin layer of clear

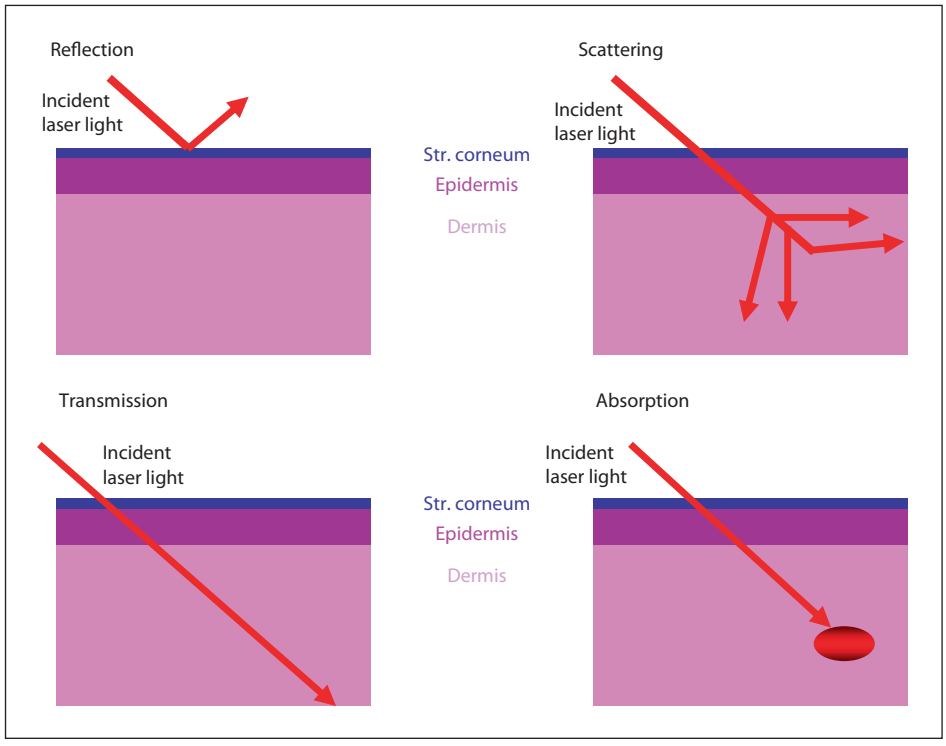

Fig. 5. Laser tissue interactions.

oil or gel onto the skin. It is crucial to try to keep the amount of reflection at the surface at its minimum as increased reflection means decreased fluence absorbed by the tissue.

Scattering

Once the light has passed the stratum corneum and into matter, it can then be scattered within the tissue. Scattering mainly occurs with shorter wavelengths – largely in the dermis, due to the collagen fibers – and predominantly in the forward direction. The amount of scattering of laser energy is inversely proportional to the wavelength of incident light, with shorter wavelengths scattered more and longer wavelengths less. With some exceptions, this also results in deeper penetration of longer wavelengths. The scattered light shows a different direction from the original direction of the incident light. This plays an important role in the spatial distribution of absorbed light energy. Hence, scattering defocuses the spot of light by spreading out the beam, which actually leads to the irradiation of a larger area. With larger laser beam diameters (spot size), less scattering occurs, while at the same time there is deeper penetration and less loss of energy with depth of penetration [17].

Transmission

Finally, residual light – which has not been reflected, absorbed, or scattered – will be transmitted into deeper structures, such as the subcutaneous tissue. The transmission of light is important for longer wavelengths to reach and target deeper structures

within the tissue. Shorter wavelengths of 300–400 nm will already have been scattered and only penetrate superficially, while scattering at 1,000–1,200 is minimal and hence penetration is great.

From 300 nm and higher, the depth of penetration of the laser light increases with increasing wavelengths, e.g. a 755 nm alexandrite laser beam will penetrate deeper than a 532 nm KTP laser beam [18]. At high water absorption wavelengths (2,940 nm), penetration is once again decreased by the trapping of light in the upper layers of the skin where water is the most abundant. Clinically, when choosing a laser for a certain indication, not only the absorption maximum of the target chromophore needs to be taken into account, but also the depth of penetration of the chosen laser wavelength in order to actually be able to reach the target within the tissue.

Light-Tissue Effects

As described before, a biological effect of laser light in the tissue can only be achieved if the light is absorbed and converted into thermal energy. The biological effect depends on the actual temperature achieved in the target chromophore or tissue, as well as on the period of time the target is at the respective temperature. Finally, tissue effects also depend on the conduction of the heat from the target to the surrounding tissue. This means the damage achieved in the tissue depends on the energy density, the pulse width/pulse duration, and the heat conduction [19].

Laser light that is absorbed by a chromophore can result in three biological reactions: photothermal, photomechanical, or photochemical.

Photothermal

Photothermal effects occur when the absorbed light energy within the chromophore is converted into thermal energy. This is the primary mechanism by which lasers function in skin. Depending on the actual temperature achieved within the target, various effects, such as coagulation or vaporization, can occur. A temperature increase in the tissue of only 5°C can lead to tissue injury with subsequent inflammation and repair. At temperatures above 60°C, a denaturation of proteins and DNA and a coagulation of the tissue occur. Finally, at temperatures above 100°C, the intracellular water exceeds the boiling point and vaporization occurs, which can be seen clinically as ablation of the tissue [16].

Photomechanical

Photomechanical reactions within the chromophore occur when extremely high energies get absorbed at short pulse durations, which lead to extremely rapid thermal expansion of the target and subsequent photomechanical destruction [20]. Such reactions play an important role in selective photothermolysis of melanin or tattoo ink particles treated with the nanosecond Q-switched lasers. Another example of photomechanical destruction of tissue is with the purpura induced by the high-fluence short-pulse-width pulsed dye laser.

Photochemical

Finally, photochemical reactions occur with endogenous or exogenous photosensitizers, such as those used in photodynamic therapy, where light-absorbing chromophores are introduced into the tissue and then elicit selective photochemical reactions by light absorption. Photosensitivity associated with porphyria can also be regarded as a photochemical reaction.

A more detailed description on light and tissue interactions can be found in a dedicated chapter [Weber et al., 24–34].

Selective Photothermolysis

The theory of selective photothermolysis, proposed by Anderson and Parrish in 1983, is one of the most important concepts to explain laser-

tissue interactions and why laser light can be used for targeted therapeutic purposes [4]. It states that laser energy can be absorbed by a defined target chromophore, leading to its controlled destruction without significant damage to the surrounding tissue. This means that we can selectively destroy tissue, for example melanosomes or tattoo ink particles within the skin, without damaging the surrounding tissue, such as vessels or collagen. In order for this concept to hold true, a number of principles need to be applied:

Wavelength
First, the wavelength of the laser light needs to correspond to the absorption maximum or lie within the absorption spectrum of the respective target chromophore.

Pulse Duration
Second, the pulse duration, also called 'pulse width' of the laser beam must be equal to or shorter than the thermal relaxation time (TRT) of the target chromophore. The TRT is defined as the time needed for the target chromophore to dissipate 63% of its peak temperature. This time is directly proportional to the square size of the chromophore. Hence, small objects cool faster than large ones, while larger chromophores have a longer TRT than smaller chromophores. The pulse width is thus determined by the size of the target chromophore.

Energy Density
Third, the energy density delivered by the laser beam, also referred to as fluence, must be high enough to actually destroy the target chromophore within the defined pulse duration. Energy densities are measured in joules per centimeter squared (J/cm^2).

Based on these concepts, the wavelength, pulse duration, and fluence must all be tailored to the properties of the target chromophore and clinical indication in order to produce a desirable clinical outcome and avoid complications.

In most cases, the clinical target of the treatment is also the chromophore used in the treatment. However, in many cases, the chromophore and the target are not the same. In these cases, the pulse width selected may actually differ significantly from the TRT of the chromophore. Consider, for example, in laser hair removal, where the target and the chromophore are not the same. The target chromophore in hair removal is melanin; however, the clinical targets seem to be the hair matrix and the stem cells. The hair matrix contains melanin and hence can be destroyed by direct heating. However, the stem cells do not contain pigment and are found at a distance from the pigment-containing target chromophore, namely the melanosomes within the hair shaft. If we heat only long enough to destroy the hair itself (nanosecond pulse width), the hair itself will fragment, but no permanent removal will result. In order to destroy the non-pigmented stem cells, and achieve (semi)permanent hair removal, the heat must diffuse from the pigmented area (hair) to the target. Thus, the clinical target will be destroyed by heat diffusion rather than by direct heating. This can only be achieved if the TRT is longer than the TRT of the chromophore. This mechanism has been proposed as the concept of thermal damage time as an extended theory of selective photothermolysis for non-uniformly pigmented targets [21]. It basically states that thermal destruction of a clinical target at a distance from the chromophore can be achieved by heat diffusion. Importantly, the pulse width chosen must be longer than the TRT of the chromophore, in order to allow heat diffusion. Using an alexandrite laser with a nanosecond pulse width – shorter that the TRT of the melanosomes – and one with a several millisecond pulse width results in very different treatments (table 2). The short nanosecond pulse width (high, short peak heat) will result in destruction of melanin (or tattoo pigment),

whereas the millisecond pulse width (with cooling) will result in hair removal. Much of this can also be attributed to epidermal cooling, which will be discussed below.

Summary

Principles of selective photothermolysis:
- Wavelength: preferentially absorbed by target chromophore
- Pulse width: shorter than TRT of target chromophore
- Fluence: high enough to destroy target chromophore

Parameters to adjust according to target:
- Wavelength
- Pulse width
- Fluence

Table 2. Thermal relaxation times of important laser targets

	Size, μm	Thermal relaxation times (approx.)
Tattoo ink particle	0.5–4	10 ns
Melanosome	0.5–1	1 μs
Erythrocyte	7	2 μs
Blood vessel	50	1 ms
Blood vessel	100	5 ms
Blood vessel	200	20 ms
Hair follicle	200	10–100 ms

Chromophores in the Skin

There are various chromophores in the skin that are able to absorb light; however, the three main endogenous chromophores in the skin are melanin, hemoglobin, and water. Each of these chromophores has its own absorption spectrum and absorption peaks, detailing their relative absorption for each wavelength (fig. 6). This figure is of major importance in laser dermatology and should be studied and understood in detail, as from this curve the corresponding wavelengths and hence laser devices to target the respective chromophore can be identified.

Melanin shows a decreasing absorption spectrum ranging from 400 to 750 nm. The absorption spectrum of hemoglobin reaches from 400 to 600 nm, with absorption peaks that can be targeted preferentially. Oxyhemoglobin shows its maximum absorption peak at 418 nm, followed by smaller peaks at 548 and 577 nm. These absorption peaks can be targeted specifically, in order to minimize absorption by competing chromophores.

Water shows increasing absorption, starting at mid-infrared and increasing towards the infrared portion of the electromagnetic spectrum.

Thus, visible and UV light are mainly absorbed by melanin and hemoglobin, while infrared light is mainly absorbed by water.

Tattoo ink is the main exogenous chromophore of importance in laser dermatology. Its corresponding wavelengths are based on the color of the ink particle and will be discussed in detail in the chapter 'Tattoo removal' [Adatto et al., 97–110].

Choosing the optimum wavelength for the absorption of the respective target chromophore is one important thing. However, for clinical results, one has to bear in mind that within the visible range, penetration depth increases with increasing wavelengths. Hence, although the highest melanin absorption is at short wavelengths within the visible range, those wavelengths will not penetrate deeply enough to, for example, reach melanin located in the dermis and consequently treat a dermal pigmented lesion. Thus, wavelength has to be considered with regards to absorption maximum as well as depth of penetration. For example, a Q-switched 532 nm laser may easily remove superficial pigmented lesions such as a lentigo, but will not be able to penetrate far enough to treat a nevus of Ito due to the deep location of the pigment.

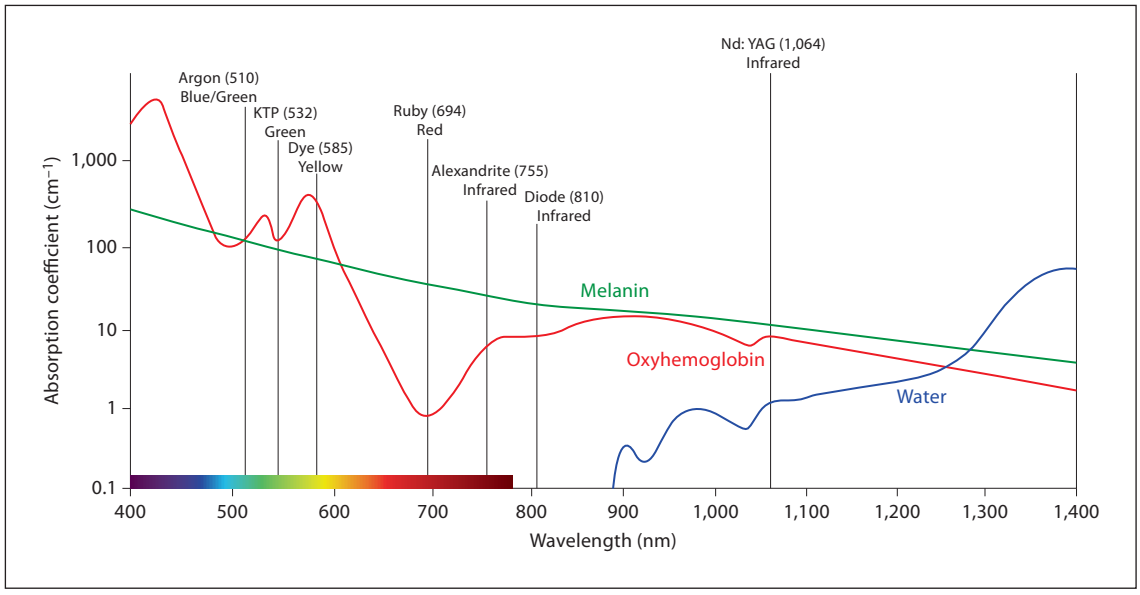

Fig. 6. Absorption spectra of the three main skin chromophores.

Summary

- The specific light-absorbing targets are known as chromophores.
- The three main chromophores in the skin are melanin, (oxy)hemoglobin, and water.
- Tattoo ink is the main external chromophore of importance in laser dermatology.
- The peak absorptions of chromophores are: melanin = 400–750 nm; hemoglobin = 400–600 nm; water = mid- and far-infrared.

Laser Properties and Parameters

When working with lasers, there are a few definitions that need to be clearly understood (table 3).

For continuous-wave lasers, time, power, and spot size are essential. For pulsed lasers, energy per pulse, pulse duration, fluence, and spot size are the most important parameters.

The energy of laser light refers to the number of photons delivered in a single pulse and is measured in joules. Joules are thus suitable to describe the energy of pulsed lasers. The energy per area is the fluence or energy density, which is expressed in joules per cm^2. The power of a laser is measured in watts and expresses the amount of energy the laser releases per unit time, i.e. how many joules are delivered per second (W = J/s). Watts are mainly used for continuous-wave lasers. Irradiance describes the power density, i.e. watts per square centimeter. It describes the intensity of a continuous-wave laser beam.

The time over which energy is delivered – i.e. the time of actual lasing, which is especially important in pulsed lasers – is the pulse duration or the pulse width. It can reach anywhere from nanoseconds to seconds. The frequency (pulse repetition rate) at which the single pulses are delivered is measured in hertz and 1 Hz equals 1 pulse per second. The wavelength characterizes the type of laser light and is measured in nanometers. It

refers to the distance between two peaks of the light waves.

Finally, the spot size is the diameter of the laser beam, and is measured in millimeters.

Table 3. Laser properties and parameters

Energy	joules = watts × seconds
Fluence	energy density = joules/cm^2 = watts × seconds/cm^2
Power	watts = joules/second
Irradiance	power density = watts/cm^2
Pulse duration	seconds, milliseconds, nanoseconds
Frequency	hertz = pulses per second
Wavelength	nanometers
Spot size	millimeters

Fluence describes the energy density of a pulsed laser beam:

$$\text{Fluence} = \frac{\text{intensity} \times \text{time}}{\text{area}} = \frac{\text{watts} \times \text{seconds}}{\text{cm}^2} = \frac{\text{joules}}{\text{cm}^2}$$

Irradiance describes the power density of a continuous-wave laser beam:

$$\text{Irradiance} = \frac{\text{intensity}}{\text{area}} = \frac{\text{watts}}{\text{cm}^2}$$

Spot Size

The spot size of a laser is an important parameter, which in many lasers can be chosen within a range of possibilities. Spot size is not simply based on the size of the target area. In fact, the spot size can have a great deal of influence on the depth of penetration of the laser, regardless of wavelength. Clearly, it is easy to understand that the larger the area to be treated, i.e. when performing photoepilation of hairs on an entire male back, a larger spot size is more convenient and faster. The opposite would be true when removing unwanted hair from an upper lip. However, one needs to be aware that the spot size directly influences the fluence and the irradiance of the laser beam as well as the depth of penetration and the scattering of the laser light. The fluence and the irradiance of the laser beam, which measure energy and power density, respectively, are inversely proportional to the square of the radius of the spot size. Thus, decreasing the spot size by 50% will increase the fluence or irradiance by a power of 4. Hence, to maintain the same energy density with a spot size only half the diameter, one would have to reduce the fluence or irradiance by a factor of 4. Further, small spot sizes allow for greater scattering of the laser beam, thus limiting its penetration, while larger spot sizes penetrate more deeply into the tissue due to less scattering [18]. This increase in penetration with larger spot sizes does not hold true when considering microscopic spot sizes, as is seen with fractional resurfacing. Spot sizes of one hundred to several hundred micrometers somehow penetrate very deeply into the skin. Very high energies can be safely delivered without significant epidermal damage via micrometer spot sizes, allowing for deep penetration and rapid recovery.

Beam Profiles

The beam profile represents how the intensity of the light produced by a laser is distributed across the beam diameter. This special distribution of power is also referred to as transverse electromagnetic mode (TEM). Most commercial lasers used in dermatology produce a beam with a gaussian profile. This beam profile is called the fundamental mode or TEM_{00} of the laser. In this beam profile, the intensity is not the same across the beam diameter but its intensity peaks at the centre of the diameter. Roughly 86% of the power is contained at that impact spot. From the center of peak intensity, the intensity then falls off to both sides with a gaussian distribution (bell shaped). When focused through a lens, this mode of lasing gives the smallest focal point. However, such a distribution of light intensity clinically requires that the target be treated with some degree of overlap, in order to achieve more uniform energy throughout the tissue.

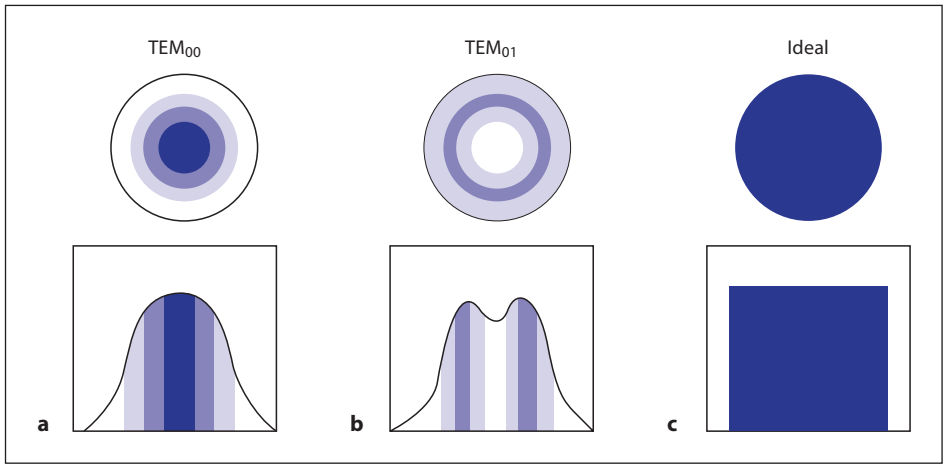

Fig. 7. Beam profiles. **a** Gaussian distribution with central hot spot which contains approximately 86% of the total energy. **b** Donut distribution with cold spot in the center. **c** Ideal energy distribution across the entire spot.

Other modes, so called higher order modes ($TEM_{01,}$ TEM_{02}) of laser operation are for example doughnut-shaped or target-shaped modes. There, the intensity is greatest at the edge of the beam diameter (doughnut-shaped) or fluctuates across the beam diameter (target-shaped). The advantage of these modes is that they deliver a more constant intensity across the beam diameter and overlapping should not be performed [22, 14]. Ideally of course, the power would be distributed evenly over the entire surface area (fig. 7).

Pulse Duration

There are different modes of how laser light can be delivered. It can either be delivered in a continuous wave or a pulsed wave. The continuous-wave light is emitted over an uninterrupted period of time with a constant beam of light (fig. 8a). These continuous waves are of rather low power with limited peak energies. They are mostly gaseous lasers, such as the CO_2 laser. Continuous beams can be mechanically shuttered to produce beams with short emission times, which then are referred to as quasi-continuous-wave lasers. In fact these are just interrupted emissions of the usually continuous-wave laser energy (fig. 8b). Such mechanical shutters are able to generate pulses with durations between 1 ms and 1 s. However, those are not truly pulsed laser beams and the peak power is still low. Truly pulsed lasers can be either short pulse with nanosecond (10^{-9} s) pulse durations, or long pulse with pulse durations within the millisecond (10^{-3} s) range. There are also investigations into creating picosecond (10^{-12} s) pulsed lasers. Truly pulsed lasers are high-power lasers that emit ultra-short single pulses with extremely high energies (fig. 9). They are mostly solid state lasers, such as the Nd:YAG. Quality switching or Q-switching is the method of producing a short pulse of light with high peak powers by using an electro-optical switch with two polarizers within the laser chamber, suddenly creating a population inversion and emitting the stored energy in extremely short highly energetic pulses. Hence, Q-switched lasers can emit ultra-short pulses in the range of nanoseconds with extremely high peak power.

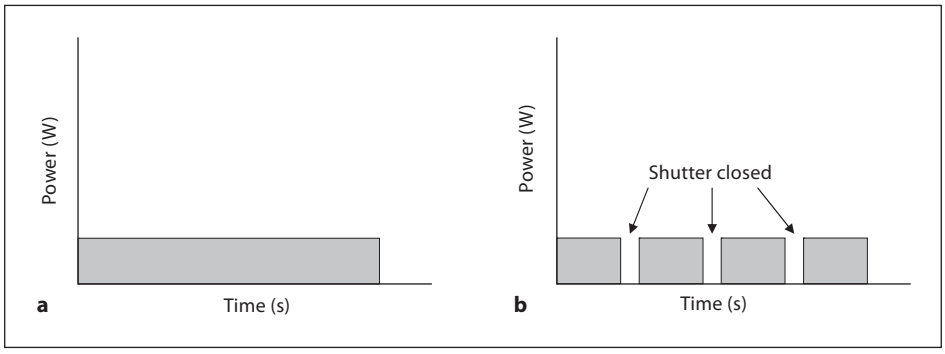

Fig. 8. **a** Continuous-wave lasers. **b** Quasi-continuous-wave lasers.

Continuous laser light leads to bulk heating and nonselective tissue damage. The concept of selective photothermolysis applies only to pulsed lasers which can emit light in times shorter than the TRT of the target chromophore and as such allow for selective damage. The pulsed mode further allows for the tissue to cool between pulses, reducing the spread of thermal damage.

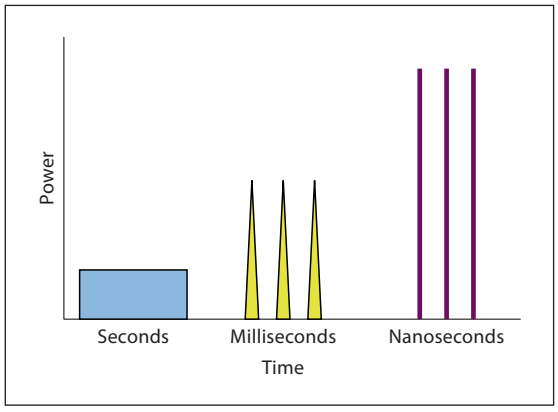

Fig. 9. Pulsed lasers.

Surface Cooling

When we aim at a target located in the dermis or subcutaneous tissue, the laser must first pass through the epidermis. If the target is not located in the epidermis, the epidermis needs to be protected from this transient passage of light. This can effectively be achieved by cooling. In many laser and light procedures, especially during photoepilation and the treatment of vascular lesions, cooling of the epidermis is the one crucial factor in avoiding adverse events.

Cooling can be performed before, during, and after treatment, and is hence referred to as pre-cooling, parallel cooling, or post-cooling. Pre-cooling is performed shortly before laser treatment in order to reduce or prevent excess heating of the epidermis during treatment. It is usually performed by a cryogen spray built into the laser handpiece, an air cooling device, contact cooling, or even ice. Parallel cooling is performed simultaneously with laser treatment, usually by cooled sapphire tips within the handpiece. Post-cooling helps to extracted excess heat from the tissue and prevent side effects such as post-inflammatory hyperpigmentation due to bulk heating and damage to surrounding tissue structures. Cooling at any stage of the

treatment reduces pain and edema and patients appreciate its use.

However, care has to be employed when cooling during the targeting of epidermal structures. Too much cooling of the epidermis can result in ineffective treatments. Cryogen burns or post-inflammatory hyperpigmentation from over-aggressive cooling is also a danger. The proper balance between laser energy and cooling is always critical.

Summary

When using lasers in clinical applications, it is important to remember all aspects of the laser, including the wavelength, pulse duration, fluence, and cooling. These should always be viewed and adjusted according to the biology of the target structure and the tissue. Understanding the basic principles of lasers will allow you to expand your treatment indications and success.

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I. Bogdan Allemann, MD
Klinik für Plastische Chirurgie und Handchirurgie, Universitätsspital Zürich
Rämistrasse 100
CH–8091 Zürich (Switzerland)
Tel. +41 44 255 1111, E-Mail Inja.bogdan@usz.ch