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Comprehensive Textbook of Cosmetic Dermatology, Laser & Energy-based Therapies



Zeina Tannous
Molly Wanner

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2

CHAPTER

Principles of Laser, Light, and Tissue Interactions

Omer Ibrahim, Jeffrey S Dover

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INTRODUCTION

Light is released, transferred, and absorbed in specific amounts of energy called quanta.[1] In dermatology, quanta of light have been harnessed and used in lasers and other light-based devices as integral sources of therapeutic energy in the treatment of many ailments, from photoaging to cutaneous neoplasia. Laser is an acronym for *light amplification by the stimulated emission of radiation*. In his "The Quantum Theory of Radiation," Einstein described the transfer of energy from light to atoms, ultimately leading to the emission of electromagnetic radiation. Most atoms exist in a ground-energy state, and upon absorbing light energy are converted to higher energy levels. Upon their return to their ground state, they release energy as photons or electromagnetic waves. When a photon of a specific wavelength collides with an excited atom, two photons are released concurrently with equal frequencies, a phenomenon known as "stimulated emission".[2,3] The first laser was created in 1960 by using an energy source to energize solid ruby crystals.[4] Since this crucial discovery, lasers and light-based devices have evolved rapidly, and their applications have spanned almost every field in modern medicine.

LASER FUNDAMENTALS

Light and other forms of electromagnetic radiation are composed of photons (particles or quanta of light) that travel in a vacuum at a constant velocity, also referred to as the speed of light ($c = 2.998 \times 10^8$ m/s).[5] The speed of light, c, is correlated to wavelength λ and frequency ν, by which: $c = \lambda\nu$. The energy of a photon is therefore proportional to its frequency, such that the energy of radiation increases with increasing frequency (and decreasing wavelength). Visible light, the only wavelengths visible to the human eye, ranges from 390 nm (violet) to 700 nm (red), and represents a relatively small portion of the electromagnetic spectrum.[5] Most lasers used in dermatology employ wavelengths within the visible light range and the near infrared spectrum.

Natural sunlight and spontaneously emitted light such as that from a light bulb are formed as a result of excited atoms that emit photons at different times, in any direction and usually with different wavelengths. This results in electromagnetic radiation that is broadband (multiple wavelengths), incoherent, and divergent. In contrast, lasers are fundamentally different (**Table 1**). A laser device creates energy in the form of a monochromatic,

TABLE 1: Comparison of three types of light sources.

	Lasers	Intense pulsed light	Ambient light
Properties	Monochromatic, coherent, collimated	Broadband, incoherent, divergent	Broadband, incoherent, divergent
Energy source	Flashlamp, electrical current, or another laser	Flashlamp	Pre-existing source (e.g., sunlight, lamp, flashlight, etc.)
Wavelengths	193–10,600 nm (monochromatic)	420–1,200 nm (broadband, but with filter capability to narrow wavelength band)	Depends on light source
Output	Continuous or pulsed	Pulsed	N/A
Pulse duration	Milliseconds to picoseconds	Milliseconds	N/A

coherent, and collimated beam of light that interacts with its target tissue. All laser devices contain an energy source and an optical resonator (**Fig. 1**). The function of the energy source is to excite atoms and electrons to an elevated energy state; the energy source can be in the form of an electrical current, flashlamp, or even another laser.[6] The resonator is comprised of a medium (solid, liquid, liquid crystal, or gas) that determines the resultant wavelength and properties of the laser. The medium is enclosed within a tube that contains two parallel mirrors. One of the mirrors is completely opaque and the other is partially transmissible. The laser's energy source pushes the electrons in the medium to an excited state, a process referred to as "pumping." After pumping, the excited atoms or molecules return from the excited to the ground state and in doing so release photons with a specific wavelength. The combination of multiple atoms and molecules undergoing this transition with resultant emissions of wavelengths of identical frequencies allows another process to occur: The stimulated emission of photons, as described by Albert Einstein.[2] When a photon of a certain wavelength collides with an electron in an excited state, a second photon of the same wavelength is emitted (**Fig. 2**). Therefore, one photon stimulates the next atom or molecule to emit a photon. The resulting two photons stimulate another two atoms or molecules, resulting in the release of four photons and so on (**Fig. 2**). The photons oscillate back and forth between the parallel mirrors, colliding with and exciting other atoms in the process. This chain reaction increases the number of photons exponentially, while maintaining their energy. Inversion occurs when the number of excited atoms outweighs the number of atoms in the ground state. Finally, the photons that are traveling in a perfectly parallel direction exit the resonator through the partially transmissible mirror at the distal end of the resonator as a laser beam (**Fig. 1**).

The resulting laser beam exiting the resonator is monochromatic, coherent, and collimated (**Table 1**).[5] Unlike natural or incandescent light that emits light across the entire visible spectrum, the monochromatic nature of a laser describes its temporal coherence—laser light is composed of a single wavelength. This property of laser light is important therapeutically as it allows for the selective absorption of the laser by specific chromophores (light-absorbing compounds). Laser light is also coherent, in that the waves are in phase in space and time. This allows the beam to travel larger distances with minimal

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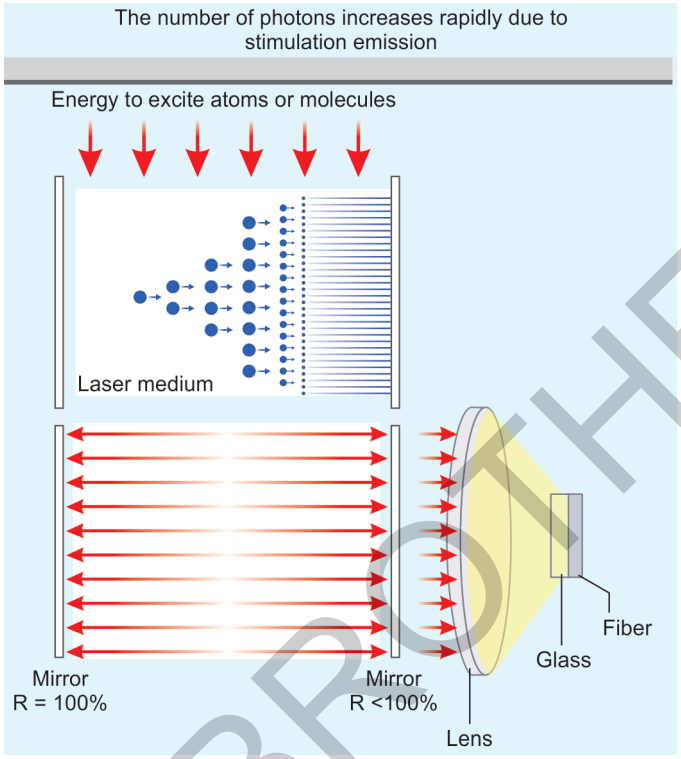

FIG. 1: A laser is composed of an energy source and an optical resonator. The energy source excites atoms and electrons to an elevated energy state. The resonator is comprised of a medium that determines the wavelength of a laser, and two parallel mirrors. Photons oscillate back and forth between the parallel mirrors. The photons that are traveling in a perfectly parallel direction exit the resonator through the partially transmissible mirror at the distal end of the resonator as a laser beam.

Source: Used with permission from Landthaler M, Bäumler W, Hohenleutner U. Chapter 239. Lasers and Flashlamps in Dermatology. In: Goldsmith LA, Katz SI, Gilchrest BA, Paller AS, Leffell DJ, Wolff K (Eds). Fitzpatrick's Dermatology in General Medicine, 8th Edition. New York, NY: McGraw-Hill. 2012.

divergence and without substantial loss of intensity. Finally, laser light is collimated, in that all the photons are parallel to each other. This allows the light to travel long distances with minimal distortion and dispersion. Collimation also allows the dense packing of energy into a relatively small volume of light.

Laser light can be continuous or shuttered, depending on the temporal behavior of the excitation of the atoms or molecules. When the energy supply to the laser medium is continuous, the emission of the laser is considered continuous-wave emission, and a continuous beam of laser light is produced with little or no variation in power output overtime. Shutters may be opened or closed to either permit or obstruct the laser beam from exiting the device.

Pulsed excitation occurs when the energy supply of the laser is limited to a certain time interval by gas discharge or flashlamp and the laser emits pulses accordingly. These laser pulse durations can vary from microseconds to milliseconds. Pulsed excitation allows for

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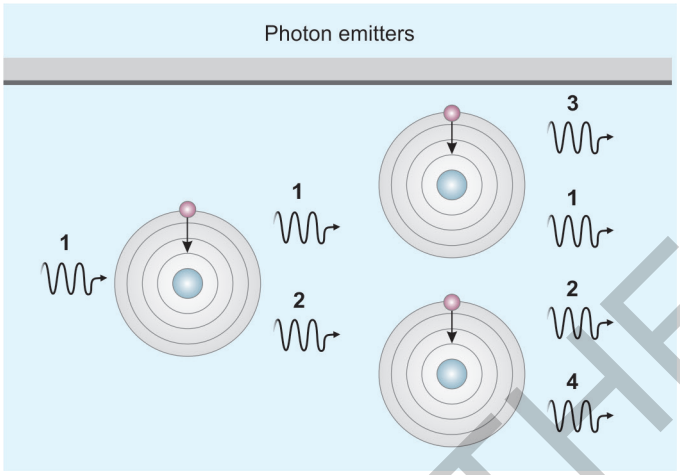

FIG. 2: The stimulated emission of radiation: When a photon (photon 1) of a certain wavelength collides with an electron in an excited state, a second photon (photon 2) of the same wavelength is emitted. Photon 1 then stimulates another excited electron, as does photon 2, both emitting photon 3 and photon 4, respectively. This chain reaction continues until a laser beam of coherent and parallel photons exits the resonator.

Source: Used with permission from Landthaler M, Bäumler W, Hohenleutner U. Chapter 239. Lasers and Flashlamps in Dermatology. In: Goldsmith LA, Katz SI, Gilchrest BA, Paller AS, Leffell DJ, Wolff K (Eds). Fitzpatrick's Dermatology in General Medicine, 8th Edition New York, NY: McGraw-Hill. 2012.

peak power outputs up to 100-fold greater than continuous wave lasers. To achieve very short pulse durations in the nanosecond or picosecond range, the resonator is additionally equipped with an optical polarizer and a non-linear crystal. The polarizer sets the linear polarization of the laser beam that passes through the crystal. However, if an appropriate electrical voltage is applied to the crystal, the polarization of the photons is changed inside the crystal and subsequently blocked by the optical polarizer. This prevents photons from running back and forth inside the resonator (oscillation), which hampers light amplification and which is equivalent to a low circuit quality of the resonator. In such a case, the stimulated emission is minimized and the pumping of the laser medium produces a maximum number of excited atoms or molecules in the laser medium. When this maximum number is achieved, the electrical voltage of the crystal is switched off within nanoseconds and the photons can easily pass the crystal, which is equivalent to a high circuit quality of the resonator. Now, the laser oscillation immediately starts and the light amplification yields a very high number of photons per time unit due to the high number of excited atoms or molecules in the laser medium. The outcome of this circuit quality switch (Q-switch) is a very intense laser pulse with pulse durations in the range of nanoseconds to picoseconds.

INTENSE PULSED LIGHT FUNDAMENTALS

Intense pulsed light (IPL) devices are not lasers; they contain a flashlamp that delivers noncoherent light ranging from approximately 420 nm to 1,200 nm. IPL devices can be modified to emit different bands of light, fluences, and pulse durations (**Table 1**).[7] The monochromatic light can be tuned

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using different cutoff filters and separate handpieces in order to maximize certain desired wavelengths depending on the target chromophore and the skin type of the patient. The filter cuts off emitted light, permitting only wavelengths longer than a certain threshold value to pass through. For example, a 560-nm filter allows only wavelengths longer than 560 nm to pass through. Therefore, given the vascular absorption peak of 585–595 nm, a 560-nm filter can be used to treat vascular lesions.[7]

LASER-TISSUE INTERACTIONS

When a laser light comes into contact with skin, it may undergo reflection, transmission, absorption, and scattering.[8] The fundamental action or effect of the laser depends on its absorption. Absorption describes the conversion of laser energy to heat when its photons strike the target chromophore. Reflection occurs when the laser beam strikes the skin's surface at an oblique angle and a portion of the beam bounces off the surface in a different direction. Of note, even laser beams positioned exactly perpendicular to the skin still undergo up to 5% reflectance due to the inherent reflective property of the stratum corneum; therefore, safety goggles are necessary when wielding lasers at all times.[5] Scattering of a laser beam describes the deviation of laser light once it passes through the stratum corneum into the tissue, due to contact with small molecules in the skin. Scattering occurs more readily with shorter wavelengths. The size of the scattering objects ranges from a few nanometers (small cell organelles and cell membranes) to a few microns (large cell organelles, cells, and collagen fibers) to hundreds of microns (hair follicle and sweat glands). As expected scattering increases the effective area treated by the laser due to aberrant scattered laser light, thereby increasing the effective heated area. Finally, transmission of the laser occurs when the beam is not absorbed by the target chromophore and is conducted to deeper structures. Transmission is more likely to occur with longer wavelengths and larger spot sizes.[5] Once the length of the wavelength surpasses approximately 1,100 nm, the increase in penetration depth reverses because radiation >1,100 nm is increasingly absorbed by water in skin.

For monochromatic lasers, photons of a single wavelength are absorbed by specific chromophores (targets) according to their absorption coefficient. Given the polychromatic nature of IPL radiation, photons of different wavelengths are absorbed by different chromophores, thereby necessitating the use of cut off filters.

The absorption of laser or IPL light by a chromophore can result in a several reactions: Chemical (photochemical), mechanical (photoacoustic or photodisruptive), or thermal.[5] A photochemical reaction is one in which a microscopic chemical or biological response is induced by the light energy. For example, low-level laser energy to the scalp is absorbed by the mitochondria, thereby activating the cellular respiratory chain to induce hair growth.[9] A photoacoustic or photodisruptive response describes the generation of high energy acoustic waves within the target chromophore, as exemplified by the use of nanosecond or picosecond lasers in the treatment of pigmented lesions and tattoos: High-energy photons absorbed by pigment particles generate acoustic waves within the particles leading to their break up.[10]

The photothermal response is the mechanism on which most lasers rely. This response refers to the transformation of light energy into heat when absorbed by a chromophore, causing direct damage to adjacent structures. The neodymium-doped yttrium aluminum garnet (Nd:YAG) laser (1,064 nm frequency doubled to 532 nm) and pulsed dye lasers (585–600 nm) are absorbed both by

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hemoglobin and melanin but they sit near peaks of absorption of oxyhemoglobin. Radiation of ruby lasers (694 nm), alexandrite lasers (755 nm), and diode lasers (around 810 nm) are less well absorbed by oxyhemoglobin and more suitable for pigmented chromophores in benign pigmented lesions and for hair removal. Radiation of the Nd:YAG laser (1,064 nm) is minimally absorbed by all skin chromophores; however, at high radiant exposures, the Nd:YAG laser can be used for pigmented or vascular lesions with either a short or long pulse. Longer wavelength lasers such as the erbium-doped yttrium aluminum garnet (Er:YAG) and carbon dioxide (CO_2) lasers are absorbed solely by water, and therefore heat up tissue for vaporization or ablation. These ablative lasers induce thermal damage to the epidermis and dermis, whereas nonablative lasers (1,440, 1,540, 1,550, 1,565, and 1,927 nm) target dermal tissue to induce bulk heating of dermal collagen and avoid damage to the epidermis. Unlike photochemical effects, photothermal reactions necessitate relatively high energy levels.[11]

At temperatures below 50°C, reversible thermal damage, local vasodilation, and inflammatory cascade activation are produced.[12] At temperatures between 50°C and 100°C, tissue and its proteins undergo irreversible coagulation or denaturation. At temperatures >100°C, tissue vaporization occurs, generating plume containing water vapor and tissue components.[12] At temperatures >100°C, and with sufficiently high energy levels, steam can be formed deep within tissue cavities, leading to explosion and mechanical damage to tissue.

By changing the intensity and pulse duration parameters, the temperature can be modulated to achieve coagulation, vaporization or explosion of a specific target in skin. In general, using higher intensities (W/cm^2) and shorter pulse durations results in higher maximum temperatures inside the target, more tissue vaporization and less surrounding coagulative necrosis. Using lower power and longer pulse widths results in less vaporization and more coagulative necrosis at the edge of the vaporized tissue.

SELECTIVE PHOTOTHERMOLYSIS

The model of selective photothermolysis was first described in the early 1980s by Anderson and Parrish, and since then, this model has revolutionized the use of lasers in the treatment of cutaneous disorders.[13] The two key concepts of this theory require that the selected wavelength of the laser must be absorbed by a specific target chromophore and the correct pulse duration (exposure time) must be used for spatial confinement of heat in the target. This was demonstrated using a 577-nm wavelength for vessels and 351-nm wavelength for pigment. By selecting a wavelength that can be absorbed by a specific chromophore with pulse duration shorter than the thermal relaxation time (TRT), the damage to the peripheral tissue is minimized. The TRT is the time needed for the irradiated tissue to cool down to half of its heated temperature.[14] The TRT is proportional to the diameter of the target. Therefore larger chromophores require longer pulse durations, and thereby demonstrate greater overall heating.

Wavelength

The main chromophores in skin on which many lasers exert their action are water, melanin, and hemoglobin. The CO_2 laser (10,600 nm) is absorbed by water in the tissue, and penetrates <0.1 mm deep.[15] The Er:YAG laser (2,940 nm) is also preferentially absorbed by water. Argon-ion lasers (488 and 514 nm), potassium titanyl phosphate (KTP) lasers (532 nm), and pulsed dye lasers (585–595 nm) may be absorbed by both hemoglobin and

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melanin. Ruby lasers (694 nm) are absorbed mainly by melanin pigment although a small amount of hemoglobin absorption also occurs. Alexandrite lasers (755 nm) are preferentially absorbed by melanin. As mentioned earlier, although there is no specific chromophore for the Nd:YAG laser (1,064 nm), this wavelength is absorbed to a small extent by hemoglobin and melanin and at higher fluences it can be used to treat pigmented or vascular lesions.

Pulse Duration

The pulse duration must correlate with the size of the target. Coagulation of blood vessels, including larger leg veins, requires pulse durations in the millisecond range.[16-19] With proper pulse duration selection, only the temperature in the vessel increases, coagulating the vessel. Longer pulse durations than necessary result in excessive heat flow to the adjacent dermis, which may cause unnecessary adverse reactions.

The pulse duration necessary to effectively treat pigment in hair, epidermal and dermal melanocytic lesions, and tattoos depends on the size of the target chromophore. Hair follicles are 0.02–0.2 mm in diameter and are targeted by pulses in the range of 10–50 ms.[20] Melanocytes are typically several micrometers in diameter (average 7 micrometers) and therefore require pulse widths ranging from milliseconds to nanoseconds. Melanosomes are smaller in size, with diameters ranging from 1 to 1.5 μm, and therefore can be targeted with nanosecond and picosecond pulses.[21] Tattoo pigment particles are on average 100 nm (0.1 μm) in diameter and are targeted by nanosecond and picosecond pulses.[22]

FRACTIONAL PHOTOTHERMOLYSIS

In the late 20th century, ablative lasers (CO_2 and Er:YAG) revolutionized the treatment of facial photodamage and skin laxity. By ablating aged, damaged skin, the induced heating and ensuing repair response made way for younger, tighter collagen, and skin. Although effective, fully ablative lasers carried high-risk profiles and prolonged recovery times, leading to their ultimate downfall. In addition, the nonablative lasers of the time fell short of the dramatic results that ablative ones offered. In order to fill the prominent gap between fully ablative and nonablative lasers, the concept of fractional photothermolysis emerged.[23] Fractional photothermolysis uses focused laser irradiation to create spatially distributed columns of microthermal injury within the treatment area, called microscopic treatment zones (MTZs). Each MTZ consists of a micro ablation zone (MAZ) and a surrounding area of coagulation known as the coagulation zone (CZ) (**Fig. 3A**). The depth and size of the MAZ and CZ depend on the laser and laser settings. For instance, the CZ of the Er:YAG laser is much smaller than the CO_2 laser. With proper technique and settings, there is enough energy in each column of MTZ to induce thermal heating or ablation without spread to adjacent tissue. Fractional irradiation allows relatively deeper penetration of the laser beam into the dermis, causing controlled vaporization or coagulation, transepidermal elimination of the coagulated debris, and reactive collagen remodeling. By maintaining the integrity of most of the intervening epidermis, fractional irradiation shortens healing times and decreases side effects.[15]

Fractional photothermolysis can be either ablative (CO_2 and Er:YAG) or nonablative (1,440, 1,540, 1,550, 1,565, and 1,927 nm). In ablative fractional photothermolysis, there is controlled vaporization and removal or ablation of thin columns of both epidermis and dermis. In nonablative fractional photothermolysis, the tissue is more gently heated yielding coagulated columns in the epidermis and dermis without any tissue ablation (**Fig. 3B**).[23,24] Intense heating of columns

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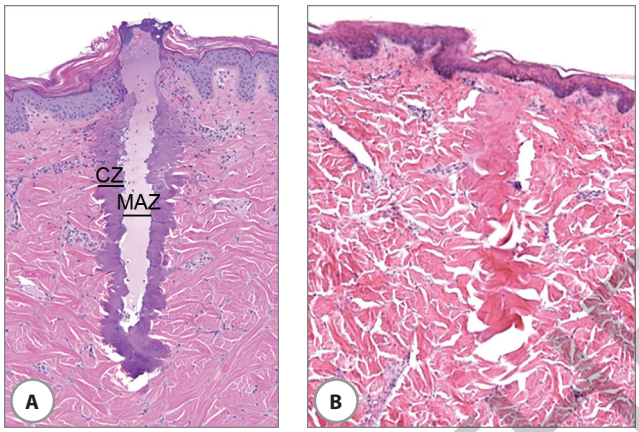

FIGS. 3A AND B: (A) Microscopic treatment zone (MTZ) of fractional ablative carbon dioxide laser, demonstrating the central ablated channel [micro ablation zone (MAZ)], and surrounding area of coagulation [coagulation zone (CZ)]. (B) microscopic treatment zone (mtz) of nonablative 1,540 nm, demonstrating coagulation within the treated dermis, and an intact overlying epidermis.

Courtesy: Dr. Emily Wenande, MD.

Source: Used with permission from Lapidoth Moshe, Adatto Maurice, Shlomit Halachmi. Treatment of actinic keratoses and photodamage with non-contact fractional 1540-nm laser quasi-ablation: An ex vivo and clinical evaluation. Lasers Med Sci. 2013;28(2):537-42.

within the dermis while leaving normal intervening skin intact can induce enough protein denaturation, release of heat shock proteins, and an inflammatory cascade that ultimately leads to collagen remodeling and synthesis. Nonablative fractionation allows for columns of thermal damage in comparison to nonfractionated, nonablated lasers that merely induce a homogenous layer of dermal heat.[23] Although, nonablative fractional lasers are much better tolerated, with fewer side effects and shorter healing time, their results may not be as impressive as ablative fractional lasers. Fractional photothermolysis has been used to treat many conditions including photoaging, acne scarring, traumatic scarring, skin laxity, and dyschromias.[24] Fractionation has also made laser-assisted topical drug delivery possible.[25-28]

CONCLUSION

Lasers and light-based devices harness the energy of light quanta in order to treat a myriad of cutaneous conditions in aesthetic and surgical dermatology. Prior to wielding these complicated and powerful devices, the clinician should have a solid knowledge base of the fundamentals and scientific basis of these devices. Herein we described the mechanism by which lasers exert their effect on their targets; we described the differences between lasers, IPL, and ambient light; and we elucidated the different types of interactions light can have with skin and tissues. Knowledge of these basic laser and light-based principles will allow the clinician to safely and effectively employ these devices in the treatment of a plethora of skin conditions.

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