Wave Theory

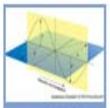
Figure 1-3


Figure 1-2

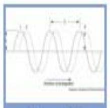
Figure 1-1


Figure 1-4

Water waves provide a good analogy for understanding light waves. When a wave travels along the water's surface, particles at the surface move up and down as the wave is propagated, but they do not move along with the wave. In the case of light, no material substance moves as the light wave propagates. Rather, at each point the electric field increases, decreases, and reverses direction in a sinusoidal manner as the wave passes (Fig 1-1). The electric field is always perpendicular to the direction of propagation.

Among the principal characteristics of a wave, as illustrated in Figure 1-1, are its wavelength (λ) and amplitude *(A)*. *Wavelength* is determined by the distance between crests of the wave. *Amplitude* is the maximum value attained by the electric field as the wave propagates. It determines the intensity of the wave. A third characteristic of a wave, not shown in Figure 1-1, is the *frequency*, which is the number of wave crests that pass a fixed point per second. Finally, multiple waves of the same amplitude may be described as "in phase," which means the light intensity is doubled; "out of phase," meaning they cancel each other; or at some level in between, resulting in an intermediate level of intensity (Fig 1-2).

In addition to an electric field, a light wave has a magnetic field that increases and decreases with the electric field. As indicated in Figure 1-3, the magnetic field *(H)* is perpendicular both to the direction of propagation of the light and to the electric field. The magnetic field is less important than the electric field and is often omitted in descriptions of a light wave.

Figure 1-4 illustrates the electromagnetic wave spectrum, including the very small portion occupied by visible light. In common usage, the term *light* refers to the visible portion of the electromagnetic wave spectrum, but it can be applied to radiation in the infrared and near-UV portions of the spectrum as well. Although the region of visible light is normally defined as 400–700 nm, the boundaries are not precise, and under certain conditions the eye's sensitivity extends into the infrared and UV regions. For example, in aphakia, without the UV absorption of the natural lens, the retina is able to detect wavelengths well below 400 nm. X-rays also produce a response in the retina, but these waves are not focused by the optical components of the eye.

The speed of light in a vacuum *(c)* is one of the fundamental constants of nature, almost exactly 3×10^8 m/sec. The wavelength of light in a vacuum (λ) is related to its frequency (ν) by the equation

$$\lambda . \nu = c$$

When light travels through any transparent medium *(m)* other than a vacuum, its velocity *(V)* is reduced, but its frequency does not change. The index of refraction *(n)* of the medium is defined as the ratio of the speed of light in a vacuum to the speed of light in the given material and is written as

$$n_m = c/V$$

Lens materials have unique indices of refraction. The index of refraction of typical CR-39 plastic lenses is 1.50, whereas that of a typical high-index lens is 1.66. The higher the index of refraction, the thinner the lens. This is important for patients with higher refractive errors who prefer "thin lenses."

Given that the frequency of a wave does not change on traveling through a transparent medium, the wavelength (λ_m) becomes shorter, as governed by the relationship

$$n_m = c/V = \lambda_v/\lambda_m$$

where λ_v is the wavelength of light in a vacuum.

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Photon Aspects of Light


Figure 1-4

When light interacts with matter, individual quanta of energy (photons) are emitted or absorbed. The amount of energy *(E)* per photon is equal to the Planck constant multiplied by the frequency and is written as

$$E = h\nu$$

where ν is the frequency of the light wave and h is the *Planck constant:* 6.626×10^{-34} J/sec. Because the frequency of blue light is greater than that of red light (see Fig 1-4), a photon of blue light has greater energy than a photon of red light.

The diagnostic use of fluorescein demonstrates a practical application of this principle. For example, a photon of blue light is absorbed by an individual fluorescein molecule. When the molecule reemits light (fluoresces), the emitted photon has a lower energy, lying in the yellow-green portion of the spectrum. The remaining energy is converted into heat or chemical energy. As a rule, light emitted through fluorescence has a longer wavelength than the excitation light.

The particle–wave duality extends to other fundamental concepts as well. The electron, for example, behaves like a wave with a wavelength much shorter than that of light. Because diffraction effects are much reduced at shorter wavelengths (see the section Diffraction, later in this chapter), extremely high resolution can be obtained with the electron microscope.

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Interference and Coherence

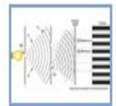

Figure 1-5

Interference occurs when 2 light waves originating from the same source are brought together. Interference occurs most readily when the light is monochromatic; that is, it lies within a narrow band of wavelengths. But interference can also be obtained with white light under optimum conditions.

In Figure 1-5, the curved lines represent the crests of the waves at a particular instant. Where the crests coincide (eg, at *A*), a maximum of intensity is produced because the energy of the electromagnetic fields is added together *(constructive interference)*. Where the crest of 1 wave coincides with the trough of the other wave *(B)*, the 2 electromagnetic fields cancel each other, and intensity is minimized *(destructive interference)*. If the 2 waves are exactly equal in amplitude, the destructive interference will be complete, and the light intensity will be zero. Thus, in Figure 1-5, the screen displays a series of light and dark bands corresponding to areas of constructive and destructive interference.

The term *coherence* describes the ability of 2 light beams, or different parts of the same beam, to produce interference. *Spatial,* or lateral, coherence refers to the ability of 2 separated portions of the same wave (*P* and *Q* in Fig 1-5) to produce interference. *Temporal,* or longitudinal, coherence is the ability of 1 wave of a beam to interfere with a different wave within the same beam (*P* and *R*). A large, white light source has a coherence close to zero. However, if the light is passed through a narrow slit, as in Figure 1-5, the spatial coherence between *P* and *Q* improves, approaching unity as the slit approaches zero width. Temporal coherence is improved when a filter is used to select a narrow band of wavelengths, thereby making it highly monochromatic. Laser light is highly coherent. Most gas lasers approach perfect temporal coherence, meaning that a portion of the beam can be made to interfere with a much later portion of the beam.

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Applications of Interference and Coherence

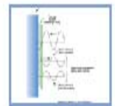

Figure 1-6

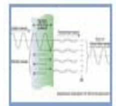

Figure 1-7

Interference resulting from the high degree of coherence in laser light can lead to serious problems in some laser applications. However, interference effects can also be put to practical use, as in *laser interferometry,* a technique for evaluating retinal function in the presence of a cataractous lens. In laser interferometry, a laser beam is split into 2 beams, which then pass through different parts of the pupil. Where the beams again overlap on the retina, interference fringes are formed, even if the beams have been diffused by the cataract.

One of the most important applications of interference is in *antireflection films* (Fig 1-6) and *interference filters* (Fig 1-7). If the 2 reflected beams in Figure 1-6 are equal in amplitude but exactly half a wavelength out of phase, the resulting destructive interference will cause the beams to cancel each other and thereby prevent reflection for a given wavelength. Modern low-reflection coatings consist of several thin layers of transparent materials designed to give a reflection of only a few tenths of a percent over the visible spectrum. Films are typically prepared by evaporation of the material in a vacuum chamber and deposition on the glass surface.

The *interference filter* (see Fig 1-7) is designed so that successive rays transmitted through the filter are exactly in phase and therefore interfere constructively. This condition applies exactly for only 1 wavelength; as a result, the filter transmits only that wavelength and a narrow band of wavelengths on either side. Other wavelengths are reflected by the interference filter. The reflecting layers can be thin films of metal such as silver or aluminum. More frequently they consist of multiple thin layers of transparent materials, with the thickness of each layer chosen to give the desired reflectance.

Thin layers can also be designed so that the transmission (or reflection) has the characteristic properties of a sharp cutoff filter. For example, a so-called *cold mirror* has a multilayer coating designed to reflect the visible (cold) light and transmit the infrared wavelengths. The excitation filter used in fluorescein angiography transmits short wavelengths, below about 500 nm, that cause fluorescein to fluoresce. The barrier filter used in the fundus camera transmits only the long wavelengths, above about 500 nm. Therefore, the fluorescent emission is received by the film, but all excitation light is excluded.

Optical coherence tomography (OCT), introduced into ophthalmology for in vivo imaging of the retina and optic nerve head, relies on low-coherence tomography, in which the signal carrying light returning from the eye is allowed to interfere with light that has traveled a path of known length. OCT is discussed in Chapter 8, Telescopes and Optical Instruments.

van Velthoven ME, Faber DJ, Verbraak FD, van Leeuwen TG, de Eye Res. Recent developments in optical coherence tomography for imaging the retina. *Prog Retin Eye Res.* 2007;26(1):57–77.

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Polarization

In general, the human eye is not sensitive to polarization of light. Nevertheless, polarization has a number of applications in visual science and ophthalmology; these are discussed in the following section.

A good analogy for polarization is light waves moving through a picket fence. The fence lets through only waves of a certain direction, blocking the rest of the waves. *Plane-polarized*, or *linearly polarized*, light consists of waves that all have their electric fields in the same plane.

In a different analogy, we could turn one end of a rope in a circular motion. The wave would then travel along the rope as a circular oscillation. Similarly, in *circularly polarized light*, the electric field at any point rotates rapidly. In *elliptically polarized light*, the electric field both rotates and changes amplitude rapidly as the wave passes.

Unpolarized light consists of a random mixture of various plane-polarized beams. *Partial polarization*, as the name implies, produces a mixture of unpolarized light and polarized light (plane, circular, or elliptical).

One way to produce plane-polarized light is to pass a beam of unpolarized light through a polarizing filter (eg, sheet plastic). This is analogous to passing a vibrating rope through a picket fence so that only the vertical vibration is transmitted. Certain crystals, particularly calcite, can be used to polarize light. As will be seen later, reflection can also cause complete or partial polarization. Even the sky acts as a partial polarizer by means of the scattering properties of air molecules.

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Reflection

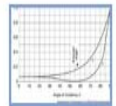

Figure 1-9

The laws of reflection as they affect light rays and the formation of images are discussed in Chapter 2, Geometric Optics.

The magnitude of the reflection at an interface between 2 media depends primarily on the difference in index of refraction between the first and second media. An air–glass interface reflects approximately 4% (at normal incidence). The air–cornea interface reflects about 2%, whereas the cornea–aqueous interface reflects only about 0.02%.

Reflection from an interface also depends strongly on the angle of incidence. As illustrated in Figure 1-9, polarization becomes important for oblique incidence. At 1 particular angle (known as the *Brewster angle*) for every interface, only 1 polarization is reflected. The fact that 1 polarization is reflected more strongly than the other enables polarizing sunglasses to block reflected light, as explained in the earlier discussion of polarization and in the following paragraph.

Total reflection occurs when light from a medium with a high index of refraction encounters a medium with a lower index at oblique incidence (see Total Internal Reflection in Chapter 2). The basis for transmission of light in fiber optics is total reflection at the internal surface of the fiber. The fiber usually consists of a high-index core glass surrounded by a lower-index cladding glass.

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Applications of Reflection

This topic is discussed in greater detail in Chapter 2, Geometric Optics.

As discussed in the previous section, total reflection occurs at the interface between a high-index glass and a lower-index glass. This interface must remain free of dirt, contamination, and contact with any other material that might degrade the total reflection. Reflection from metals such as silver or aluminum can be as high as 85%–95%. As with other materials, the reflectivity increases with the angle of incidence. Mirrors used in ophthalmic instruments usually consist of an aluminum layer that has been vacuum-evaporated on a glass substrate and then overcoated with a protective thin film of transparent material such as silicon monoxide to prevent oxidation and scratching of the aluminum surface.

Semitransparent mirrors, sometimes used as 1-way mirrors, often consist of a metallic layer thin enough to transmit a fraction of the incident light. They are not 100% efficient in that a substantial fraction of the light is also absorbed. In critical applications, partially reflecting mirrors can be made of other materials so that only a negligible fraction is lost to absorption.

Metallic reflection *partially* polarizes the reflected light. As with other materials, the perpendicular component is more strongly reflected than the parallel component. However, with metals there is no angle at which only 1 polarization is reflected; therefore, the polarization of the reflected light is never complete.

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Transmission and Absorption

Transmission is the passing of radiant energy through a medium or space. It is measured in terms of transmittance, the percentage of energy that can pass through a particular medium. For absorbing materials, the transmittance is often a function of wavelength.

Absorption is usually expressed as an *optical density* (OD). An OD of 1 represents a transmittance of 10%; an OD of 2, a transmittance of 1% (0.01); and an OD of 3, a transmittance of 0.1% (0.001).

In general, the expression for optical density is $OD = \log 1/T$, where T is the transmittance. See also Chapter 4, Clinical Refraction, for a discussion of absorptive lenses.

Duke-Elder S, Abrams D, eds. *System of Ophthalmology*. Vol V, *Ophthalmic Optics and Refraction*. St Louis: Mosby; 1970:30–36.

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Light Hazards

Although the eye requires light in order to function, it has long been recognized that light itself in excess, particularly at certain wavelengths, can be hazardous to various parts of the eye:

- The cornea and lens are particularly susceptible to UV injury in the wavelength range of 180–400 nm, from which photokeratitis and cataract can result.
- The retina is susceptible to photochemical injury from blue light in the wavelength range of 400–550 nm (310–550 nm for an aphakic eye). This is the basis for the incorporation of UV-blocking and blue-blocking chromophores in certain intraocular lenses.
- The retina is susceptible to thermal injury from optical radiation in the wavelength range of 400–1400 nm.
- The lens of the eye is susceptible to thermal injury from near-infrared radiation in the wavelength range of 800–3000 nm.
- The cornea and lens of the eye are susceptible to thermal injury from radiation in the wavelength range of 400–1200 nm.
- The cornea is susceptible to thermal injury from optical radiation in the wavelength range of 1400 nm–1 mm.

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Laser Fundamentals

Laser is an acronym for *light amplification by stimulated emission of radiation*—a phrase that highlights the key events in producing laser light. In the most simplified sequence, an energy source excites the atoms in the *active medium* (a gas, solid, or liquid) to emit a particular wavelength of light. The light thus produced is amplified by an optical feedback system that reflects the beam back and forth through the active medium to increase its coherence, until the light is emitted as a laser beam. This process is described in greater detail in the following sections.

Although Einstein had developed the basic theory of laser emission more than 40 years earlier, it was not until 1960 that Theodore Maiman built the first successful laser with a ruby crystal medium.

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Properties of Laser Light

Lasers are only one of many sources of light energy. The unique properties of laser light, however, make it particularly suitable for many medical applications. These properties are *monochromaticity*, *directionality*, *coherence*, *polarization*, and *intensity*.

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Monochromaticity

Lasers emit light at only 1 wavelength or sometimes at a combination of several wavelengths that can be separated easily. Thus a "pure," or monochromatic, beam is obtained. Although the wavelength spread is not infinitesimally small, a gas laser emission line can be as narrow as 0.01 nm, compared with the 300-nm span of wavelengths found in white light. At best, a filter might reduce the transmission of white light to a color range (bandwidth) of 5 nm at the expense of most of the white light's energy. For medical purposes, the color of light can be used to enhance absorption or transmission by a target tissue with a certain absorption spectrum. The wavelength specificity of a laser greatly exceeds the absorption specificity of pigments in tissues. In addition, monochromatic light is not affected by chromatic aberration in lens systems. Thus, monochromatic light can be focused to a smaller spot than can white light.

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Directionality

The second property of laser-emitted light is *directionality*. Lasers emit a narrow beam that spreads very slowly. As explained later in this chapter, lasers amplify only those photons that travel along a very narrow path between 2 mirrors. This process serves as a very efficient mechanism for collimating light. In a typical laser, the beam increases by approximately 1 mm in diameter for every meter traveled. Directionality makes it easy to collect all of the light energy in a simple lens system and focus this light to a small spot.

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Coherence

Coherence, meaning that all the propagated energy from the source is in phase, is the term most often associated with lasers (see Fig 1-5 and the earlier discussion, Interference and Coherence). Laser light projected onto a rough surface produces a characteristic sparkling quality known as *laser speckle.* This phenomenon occurs because the irregular reflection of highly coherent light creates irregular interference patterns, or speckle. Coherence of laser light is utilized to create the interference fringes of the laser interferometer. In therapeutic ophthalmic lasers, coherence, like directionality, is important because it improves focusing characteristics.

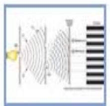

Figure 1-5

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Polarization

Many lasers emit linearly polarized light. *Polarization* is incorporated in the laser system to allow maximum transmission through the laser medium without loss caused by reflection.

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Intensity

In most medical applications, the most important property of lasers is *intensity.* Intensity is the power in a beam of a given angular size, and the physical correlate of the perception of "brightness" is the intensity per unit area. In medical laser applications, the most important radiometric terms are *energy* (J), *power* (W), *radiant energy density* (J/cm^2), and *irradiance* (W/cm^2) (see Table 1-2). The laser output is fixed in either joules (J) or watts (W). Recall that energy is work, and power is the rate at which work is done. One joule = 1 watt × 1 second, or 1 W = 1 J/s. The tissue effect is then determined by the focal point spot size, which determines energy density and irradiance (or, less properly stated, "power density"). In ophthalmic lasers, spot size is conventionally given as the diameter. Thus, a 50-μm spot size has an area of $\pi (25 \times 10^{-4})^2$ cm^2, or about 2×10^{-5} cm^2.

Figure 1-14

Table 1-2

Directionality, coherence, polarization, and, to some degree, monochromaticity enhance the most important characteristic of lasers, which is light intensity. The sun has a power of 10^{26} watts but emits energy in all directions at a great distance from the earth. Thus, a simple 1-mW helium neon laser has 100 times the radiance of the sun. Their intense radiance, combined with monochromaticity that can target selected tissues and avoid others on the basis of spectral absorption, makes lasers a unique tool in medicine. This is particularly true in ophthalmology, as the eye is designed to allow light transmission to most of its structures. Figure 1-14 summarizes the major properties of laser light in comparison with a conventional light source.

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Elements of a Laser


Figure 1-15

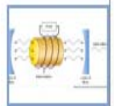
Figure 1-16

All ophthalmic lasers currently in use require 3 basic elements: (1) an *active medium* to emit coherent radiation; (2) energy input, known as *pumping;* and (3) *optical feedback,* to reflect and amplify the appropriate wavelengths.

In 1917, Albert Einstein explained the mathematical relationships of 3 atomic transition processes: *absorption, spontaneous emission,* and *stimulated emission.* According to the fundamental principles of quantum physics, certain atomic energy transitions are highly probable, or "allowed." Light energy can readily induce such an allowed transition, causing the energy of the atom to move from its ground state (E_0) to an excited state (E_1). The atom absorbs a quantum of energy at a predictable frequency appropriate to cause the specific transition. If the source of illumination is white light, a discrete frequency (line spectrum) will be subtracted from the illuminating beam. Each atomic element has a characteristic line spectrum. This process is known as *absorption* (Fig 1-15A).

Because the lowest energy state is the most stable, the excited atom soon emits a quantum of energy at the same frequency in order to return to the ground state. This process can occur without external stimulation (*spontaneous emission;* Fig 1-15B) or as a result of stimulation by a photon of light at the same frequency (*stimulated emission;* Fig 1-15C). Spontaneous emission occurs randomly in time, whereas stimulated emission is in phase with the stimulating wave. Therefore, stimulated emission is coherent. After absorption, the majority of energy release is through spontaneous emission occurring incoherently in all directions, and only a small fraction of the energy is normally released as coherent stimulated emission. The laser environment, however, amplifies only the stimulated emission.

As indicated in Figure 1-15C, stimulated emission occurs when an incident photon of the proper frequency interacts with an atom in the upper energy state. The result is the emission of a photon of the same wavelength and the return of the atom to its lower energy state. The emitted photon also has the same phase and direction of propagation as the incident photon.

The *active medium* is an atomic or molecular environment that supports stimulated emission. The active medium allows a large number of atoms to be energized above the ground state so that stimulated emission can occur. Recall that $v = c/\lambda$; hence, the particular atomic energy transition determines the wavelength of the emission ($E = hv = hc/\lambda$). Lasers are usually named for the active medium. The medium can be a gas (argon, krypton, carbon dioxide, argon-fluoride excimer, or helium with neon), a liquid (dye), a solid (an active element supported by a crystal, such as neodymium supported by yttrium-aluminum-garnet [Nd:YAG] and erbium supported by yttrium-lanthanum-fluoride [Er:YLF]), or a semiconductor (diode).

The second requirement for a laser is a means of imparting energy to the active medium so that a majority of the atoms are in an energy state higher than the ground state. This condition is known as a *population inversion* because it is the inverse of the usual condition in which the majority of atoms are in the ground energy state. The energy input that makes possible population inversion is known as *pumping.* Gas lasers are usually pumped by electrical discharge between electrodes in the gas. Dye lasers are often pumped by other lasers. Solid crystals are usually pumped by incoherent light such as the xenon arc flashlamp.

Once population inversion in an active medium has been achieved, *optical feedback* is required to promote stimulated emission and suppress spontaneous emission. The laser cavity acts as an optical resonator. Mirrors are placed at each end of a beam path to reflect light back and forth through the active medium, in which pumping maintains a population inversion (Fig 1-16). Each time the light wave resonates through the active medium, the total coherent light energy is increased through stimulated emission. Spontaneous emission, which occurs randomly in all directions, rarely strikes a mirror and therefore is not amplified.

The last element in this schematic laser design is a mechanism for releasing some of the oscillating laser light from the cavity. This is achieved by making one of the mirrors fully reflective and the other mirror only partially reflective. A portion of the light waves striking the second mirror is emitted from the cavity as the laser beam. The reflectivity of the mirror is selected to satisfy the requirements for efficient amplification in a particular system. For example, if a laser has a 98% reflective mirror, the light waves are coherently amplified by stimulated emission during an average of 50 round-trips through the active medium before they are emitted as the laser beam.

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Laser Sources

Solid-state laser sources commonly used in medical applications are ruby and Nd:YAG. Refractive surgery uses excimer lasers (ablative procedures) and, less commonly, infrared holmium:YLF (IntraLase, Advanced Medical Optics, Santa Ana, CA) and holmium:YAG lasers (laser thermal keratoplasty [LTK], laser in situ keratomileusis [LASIK]). Argon, krypton, carbon dioxide, and argon-fluoride excimer are the most important gas laser sources used in medicine. The dye laser is the only liquid laser used in ophthalmology.

In 1975, it was shown that rare gas atoms in metastable excited states could react with halogens to form diatomic rare gas halides in a bound excited dimer *(excimer)* state. Decay of these excimer molecules to a weakly bound or unbound ground state is accompanied by emission of a photon with UV frequency. Excimer lasers efficiently produce high-power UV irradiation. A number of different excimer molecules can be created, and each is associated with a specific transition and emission wavelength: argon fluoride, or ArF (193 nm); krypton fluoride, or KrF (249 nm); and xenon fluoride, or XeF (351 nm).

Semiconductor diode lasers are solid-state lasers that are extremely compact and highly efficient. These laser sources are commonly used in communications applications and in digital information and audio systems. The increased power output of semiconductor diode lasers makes them feasible for retinal photocoagulation and for some glaucoma applications.

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Photocoagulation

Even before the invention of lasers, light energy had been employed therapeutically to heat and permanently alter target tissue. This early phototherapy had its origins in observations of solar retinitis and was used in the treatment of numerous retinal disorders and glaucoma. A laser could now achieve similar effects in a more controlled manner. The term *photocoagulation* refers to the selective absorption of light energy and conversion of that energy to heat, with a subsequent thermally induced structural change in the target. These processes and their therapeutic results depend on laser wavelength and laser pulse duration. A variety of photocoagulating lasers are currently in clinical use: argon, krypton, dye, holmium, and the solid-state gallium arsenide lasers.

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Photodisruption

A second category of laser–tissue interaction uses high-peak-power pulsed lasers to ionize the target and rupture the surrounding tissue. In clinical practice, this process (known as *photodisruption*) uses laser light as a pair of virtual microsurgical scissors, reaching through the ocular media to open tissues such as lens capsule, iris, inflammatory membranes, and vitreous strands without damaging surrounding ocular structures. Currently, the Nd:YAG and Er:YAG lasers are the principal photodisruptive lasers used in clinical ophthalmology.

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Photoablation

A third category of laser–tissue interaction, called *photoablation,* arose from the insight that high-powered UV laser pulses can precisely etch the cornea in the same manner that they etch synthetic polymers. The high energy of a single photon of 193-nm UV light exceeds the covalent bond strength of corneal protein. The high absorption of these laser pulses precisely removes a submicron layer of cornea without opacifying adjacent tissue, owing to the relative absence of thermal injury. Over a decade of laboratory and clinical investigation has brought excimer laser photoablation to clinical use in refractive surgery and corneal therapeutics. (See also BCSC Section 13, *Refractive Surgery.*)

Figure 1-17

Figure 1-17 shows some typical laser wavelengths.

Azar DT, ed. Gatinel D, Hoang-Xuan T, associate eds. *Refractive Surgery.* 2nd ed. St Louis, MO: Elsevier-Mosby; 2007.

Campbell CJ. *Physiological Optics.* Hagerstown, MD: Harper & Row; 1974.

Rubin ML, Walls GL. *Fundamentals of Visual Science.* Springfield, IL: Charles C Thomas; 1969.

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Laser–Tissue Interactions

Three laser–tissue interactions are exploited for keratorefractive surgery. *Photothermal* effects are achieved by focusing a holmium:YAG laser with a wavelength of 2.13 µm into the anterior stroma. The laser beam is absorbed by water, causing collagen shrinkage from heat. This technique is approved by the FDA for treating low hyperopia.

The femtosecond laser is approved by the FDA for creating corneal flaps for LASIK and may be used for lamellar or PKP. It uses a 1053-nm infrared beam that creates *photodisruption,* a process by which tissue is transformed into plasma, and high pressure and temperature create rapid tissue expansion, leading to microscopic cavities within the corneal stroma. Contiguous photodisruption allows for creation of the corneal flap or keratoplasty incision.

Photoablation, the most important laser–tissue interaction in refractive surgery, breaks chemical bonds using excimer (for "*exc*ited d*imer*") lasers or other lasers of the appropriate wavelength. Laser energy of more than 4 eV per photon is sufficient to break carbon–nitrogen or carbon–carbon tissue bonds. Argon-fluoride (ArF) lasers are excimer lasers that use electrical energy to stimulate argon to form dimers with the caustic fluorine gas. They generate a wavelength of 193 nm with 6.4 eV per photon. The 193-nm light is in the ultraviolet C (high ultraviolet) range, approaching the wavelength of x-rays. In addition to having high energy per photon, light at this end of the electromagnetic spectrum also has very low tissue penetrance and thus is suitable for operating on the surface of tissue. Not only is the laser energy capable of great precision, with little thermal spread in tissue, but its lack of penetrance or lethality to cells makes the 193-nm laser nonmutagenic, enhancing its safety. (DNA mutagenicity occurs in the range of 250 nm.) Solid-state lasers have been designed to generate wavelengths of light near 193 nm without the need for toxic gas, but the technical difficulties in manufacturing these lasers have limited their clinical use.

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Types of Photoablating Lasers

Photoablating lasers can be divided into broad-beam lasers, scanning-slit lasers, and flying-spot lasers. *Broad-beam lasers* have larger-diameter beams and slower repetition rates and rely on optics or mirrors to create a smooth and homogeneous multimode laser beam of up to approximately 7 mm in diameter. These lasers have very high energy per pulse and require a small number of pulses to ablate the cornea. *Scanning-slit lasers* use excimer technology to generate a narrower slit beam that is scanned over the surface of the tissue to alter the photoablation profile, improving the smoothness of the ablated cornea and allowing for larger-diameter ablation zones. *Flying-spot lasers* use smaller-diameter beams (0.5–2.0 mm) that are scanned at a higher repetition rate, but to create the desired pattern of ablation, a tracking mechanism is required for precise placement. Broad-beam lasers and some scanning-slit lasers require a mechanical iris diaphragm or ablatable mask to create the desired shape in the cornea. The flying-spot lasers and some of the scanning-slit lasers use the pattern projected onto the surface to create the desired laser ablation profile without masking.

Wavefront-Optimized and Wavefront-Guided Laser Ablations

In the past, conventional treatments used laser profiles with smaller blend zones and created a more oblate corneal shape postoperatively, with induced higher-order aberrations, especially spherical aberration and coma. Wavefront-optimized laser ablations try to preserve the prolate shape of the cornea by increasing the number of peripheral pulses, which often results in better quality vision and fewer night vision complaints. This method also compensates for the decreased effect of the conventional laser in the periphery of the cornea due to the angle of the beam. As in conventional procedures, the refraction is used to program the wavefront-optimized laser ablation. This technology does not attempt to address preexisting higher-order aberrations but rather to minimize induction of spherical aberration caused by the laser ablation itself. It has the advantage of being quicker than wavefront-guided technology and avoids the additional expense of the aberrometer. There are currently few published studies comparing wavefront-optimized results with conventional or wavefront-guided laser results.

In wavefront-guided laser ablations, information obtained from a wavefront-sensing aberrometer (which quantifies the aberrations) is transferred electronically to the treatment laser to program the laser ablation. This is distinct from conventional excimer laser and wavefront-optimized laser treatments, where the subjective refraction is used to program the laser ablation. The wavefront-guided laser attempts to treat both lower-order (myopia or hyperopia and/or astigmatism) and higher-order aberrations.

Wavefront-guided lasers apply complex ablation patterns to the cornea to correct wavefront deviations from a desired final corneal shape. The correction of higher-order aberrations requires non–radially symmetric patterns of ablation (which are often much smaller in magnitude than ablations needed to correct defocus and astigmatism). Based on the difference between the desired and the actual wavefront, a 3-dimensional map of the ablation is generated. To match the intended ablation pattern with that which is ultimately delivered to the cornea, registration must be achieved using either marks at the limbus prior to obtaining the wavefront patterns or iris registration, which matches reference points in the natural iris pattern to compensate for cyclotorsion and pupil centroid shift. The wavefront-guided laser then uses a pupil-tracking system, which helps to maintain centration during treatment and allows the accurate delivery of the customized ablation profile.

The results of wavefront-guided ablations for low and moderate myopia are excellent, with well over 90% of eyes achieving 20/40 or better UCVA. Early results for wavefront-guided hyperopic and astigmatic corrections appear promising as well. Only a few published papers in the literature compare this technology to conventional treatments, however, and these do not show clear-cut superiority when the conventional procedures are performed with modern technology, including larger ablation zones, pupil-tracking systems, and anatomical registration. In general, fewer induced higher-order aberrations are found in eyes treated with wavefront technology compared with those treated with conventional treatments (although there is an increase postoperatively in both groups), but Snellen visual acuity parameters are similar. For enhancements, however, early results indicate that customized treatments appear to be better than conventional re-treatments. Investigations are ongoing in comparing wavefront-guided technology platforms and determining what constitutes the ideal optical correction.

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Optical Coherence Tomography

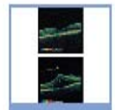
Figure 2-3

Optical coherence tomography (OCT) is a noninvasive, noncontact imaging modality that produces micrometer-resolution, cross-sectional images of ocular tissue. OCT is based on imaging reflected light. The technique produces a 2-dimensional, false-color image of the backscattered light from different layers in the retina, analogous to ultrasonic B-scan and radar imaging. The only difference is that OCT, using the principle of low-coherence interferometry, measures optical rather than acoustic or radio wave reflectivity. With the use of light instead of sound, the resolution is enhanced and the speed is much greater. First- and second-generation OCT scanners produced cross-sectional images of the retina with an axial (depth) resolution of approximately 12–15 µm; current commercial OCT scanners offer a resolution of 8–10 µm that is at least 10 times better than ultrasound. Because axial resolution depends on the "coherence length" of the light source, ultrahigh-resolution images using a femtosecond titanium:sapphire laser light source can deliver resolutions of 1–3 µm, which approaches the theoretical limit of OCT imaging. To further improve OCT imaging, Fourier-domain, or spectral-domain, technology is now available and delivers a 100-fold improvement in speed over current time-domain OCT scanners. For example, Fourier-domain scanners show greater detail (1125 scans vs 512 scans) than time-domain scanners in a shorter period of time (0.072 vs 1.23 seconds). This dramatically decreases motion artifact. Moreover, the faster scanning time allows a larger area to be scanned and offers more precise image registration.

OCT single-scan cross-sectional views (tomograms) of the retina appear similar to histopathologic specimens and have been termed "optical biopsies" (Fig 2-3). Tissues with higher reflectivity, such as the RPE, appear in brighter colors (red-white), and less dense structures, such as the vitreous and intraretinal fluid, appear in darker colors (blue-black). OCT is useful for differentiating lamellar from pseudo- and full-thickness macular holes, diagnosing vitreomacular traction syndrome, differentiating traction-related diabetic macular edema, monitoring the course of central serous chorioretinopathy, making age-related macular degeneration (AMD) treatment decisions, and evaluating for subtle subretinal fluid that is not visible on FA. An obvious benefit of higher-resolution systems is the ability to better delineate retinal layers, including the external limiting membrane and the junction between the inner and outer photoreceptor segments. This improves our ability to localize retinal pathology and subtle anatomical changes.

OCT can also produce a retinal thickness map. The OCT software automatically determines the inner and outer retinal boundaries and produces a false-color topographic map, with areas of increased thickening in brighter colors and areas of lesser thickening in darker colors. An assessment of macular volume can also be obtained from the retinal thickness map. By evaluating differences in retinal volume over time, the clinician can evaluate the efficacy of therapy. Time-domain OCT produces retinal thickness maps from 6 × 6-mm radial scans centered on the fovea, with interpolation between the scan lines, to produce a map of the macula. In contrast, Fourier-domain OCT can image the entire macula due to increased scanning speed and improved accuracy of thickness and volume measurements; it also offers the capability of improving registration, so imaging the same area from visit to visit is now possible.

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Quantitative measurement of the optic nerve head and retinal nerve fiber layer

Figure 3-17

Since the 1850s, the appearance of the optic nerve head has been recognized as critical in assessing the disease status of glaucoma. However, optic disc assessment can be quite subjective, and interobserver and intraobserver variation is greater than desirable, given the importance of accurate assessments. Thus, the need for reliable and objective measures of optic disc and associated retinal nerve fiber layer morphology is clear. A number of sophisticated image analysis systems have been developed in recent years to evaluate the optic disc and retinal nerve fiber layer. These instruments give quantitative measurements of various anatomic parameters.

Confocal scanning laser ophthalmoscopy (Fig 3-17A) can be used to create a 3-dimensional image of the optic nerve head. The optical design of instruments using confocal scanning laser technology allows for a series of tomographic slices, or optical sections, of the structure being imaged. The images acquired by this method are stored as a computer data file and manipulated to reconstruct the 3-dimensional structure, display the image, and perform data analysis. Parameters such as cup area, cup volume, rim volume, cup–disc ratio, and peripapillary nerve fiber layer thickness are then calculated. Software to evaluate the images for the statistical likelihood of glaucoma damage as well as identify areas of possible progression over time are available.

Techniques such as scanning laser polarimetry and optical coherence tomography have been used to acquire images of the retinal nerve fiber layer. The *scanning laser polarimeter* (Fig 3-17B) is basically a scanning laser ophthalmoscope outfitted with a polarization modulator and detector to take advantage of the birefringent properties of the retinal nerve fiber layer arising from the predominantly parallel nature of its microtubule substructure. As light passes through the nerve fiber layer, the polarization state changes. The deeper layers of retinal tissue reflect the light back to the detector, where the degree to which the polarization has been changed is recorded. The acquired data can then be stored, displayed, and manipulated by computer programs, just as with the unmodified confocal scanning laser ophthalmoscope. The fundamental parameter being measured with this instrumentation is *relative* (not absolute) retinal nerve fiber layer thickness. The addition of a variable corneal compensator (VCC) to include analysis of potential anterior segment birefringence has improved the quality of the information available by this technique.

Optical coherence tomography (OCT) (Fig 3-17C) uses interferometry and low-coherence light to obtain a high-resolution cross section of biological structures. The resolution of OCT instrumentation in the eye is approximately 10 µm, and OCT has the potential to yield an absolute measurement of nerve fiber layer thickness. In vivo OCT measurements appear to correlate with histologic measurements of the same tissues.

Quantitative measurement of the optic disc and retinal nerve fiber layer is a promising nascent science. The instrumentation and techniques used to acquire quantitative imaging and analysis of nerve head and nerve fiber layer anatomic parameters are rapidly evolving. Both for single measurements directed at detecting the presence of glaucoma and, especially, for serial measurements necessary to determine clinical progression of glaucoma, these technologies have great potential. The clinician must remember that no system of measurement and observation is currently more useful or has proven more reliable than good-quality stereophotographs combined with detailed and careful clinical examination.

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