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Effects of Lasers on the Human Eye

Abstract: In dealing with the relationship between human vision and lasers, this largely theoretical paper places particular emphasis upon the use of lasers within the normal operating range of the visual system, and upon the mechanisms by which laser radiation can cause threshold damage to the eye. Parallel but subordinate sections present some fundamentals of laser radiation, of the relevant aspects of the visual system, and of unit systems for the specification of laser output. A new approach to understanding laser radiation damage to the eye is developed by means of a model limited to conditions existing only at the threshold of damage. It is shown that such threshold damage to the visual system is primarily due to the effects of heat alone, but that photochemical effects and acoustic shockwaves can potentially be a cause of the threshold damage that cannot be entirely rejected under all conditions. A theoretical estimate of retinal irradiance for threshold damage is made and shown to be consistent with empirical findings. A survey of empirically determined damage thresholds is presented. A valid method of computing retinal irradiance from a laser is given, and the direction and magnitude of errors in earlier formulations are pointed out.

Introduction

The construction of the optical maser, or laser, in 1960 by Maiman brought into existence a light source having a unique set of properties and, consequently, unique ways of interacting with physical objects. Such uniqueness of these properties stimulates a host of questions about the effects of laser radiation on the visual system. Ophthalmologists ask whether and how lasers can be of practical service; physiologists ask whether lasers provide a useful tool for investigation of visual processes; engineers ask what implications of laser radiation exist for the design of devices using lasers either as a source of illumination or as a means of presenting visual information to the human operator; psychologists ask whether the special properties of laser radiation lead to unique phenomenological effects in visual perception. Further, virtually all having a professional concern with lasers must consider the potentially harmful effects of laser radiation to the eye. Hence, the purpose of this paper is to answer certain of these questions directly, to give the reader enough relevant information to enable him to answer others for himself and, finally, to direct the reader to sources of more detailed information.*

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* A goal of this paper is to address a wide variety of readers without requiring any to consult truly basic references. Hence, because of the great diversity of backgrounds of potential readers, the present paper is accompanied by basic and brief summaries on Structure of the Eye, and Characteristics of Laser Radiation.

The plan for the ensuing treatment is as follows: First, in the section *Some Visual Effects*, the special properties of laser radiation are considered individually with respect to their general implications for vision. Next, the several sections on *Damage Thresholds* form the major part of this paper. Although the original intent was simply to bring together into one paper all the scattered, relevant facts about damage thresholds, it soon became clear that the right kind of information was not available. The difficulty was that interpretation of past experimental investigations of damage thresholds have been hindered in two ways: *first*, the purpose of such experiments was to determine the threshold for permanent impairment of ocular function, but the conclusions were based solely on anatomical findings that bear an unclear relationship to functional impairment; *second*, practical interests seek a definition of those conditions that certainly *do not* lead to damage to the eye, but experiments have defined, instead, those conditions that certainly do lead to damage. The conditions defined differ by an unknown amount from the conditions for which a definition is sought.

Consequently, the published literature has offered neither experimental facts nor theoretical models that are sufficient to allow detailed coverage of the problem of defining the threshold of damage to the eye by laser radiation. We have greatly simplified matters, however, by limiting consideration to conditions of fundamental interest here, namely the

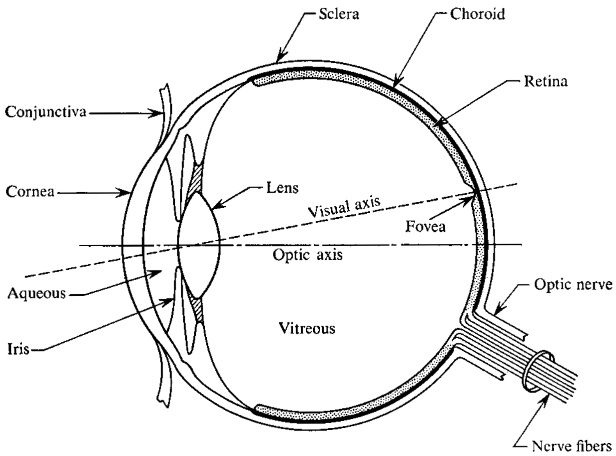

Figure 1

Structure of the Eye

Figure 1 is a drawing, not to scale, of the relevant ocular structures. The anterior surface of the eye directly exposed to the air is formed by the cornea and the conjunctiva. The transparent, laminar *cornea* is the central convexity on which light destined ultimately for absorption by the photoreceptors first impinges. Most of the refraction which brings an image to focus on the photoreceptor layer is done here, the cornea having a power of 43 diopters. (The reciprocal of the focal length expressed in meters gives the power of a lens in diopters.)

The *conjunctiva* is a transparent, mucous membrane attached to the *limbus*, or perimeter of the cornea. From the limbus the conjunctiva extends back over the eye, covering the "whites" of the eye, and then folds back to form the inside layer of the eyelid. Secretions from the conjunctiva keep the cornea clean and moist. The conjunctiva contains the blood vessels which are especially noticeable in "blood shot" eyes.

Behind the cornea is the anterior chamber filled with the *aqueous humour*, which, as its name suggests, is very nearly water; it has a refractive index of 1.336. The posterior boundary of the chamber is formed by the lens, which is partially covered by the iris.

The pigmented *iris*, which gives the eye its distinctive color, regulates the amount of light entering the eye. Through the processes of constriction and dilation, the iris controls the size of the circular aperture, or *pupil*, that it surrounds. The entrance pupil, which is the image of the real pupil as seen through the cornea, normally varies from 3 to 8 mm in diameter, but under some conditions limits of 1.3 and 10 mm are attainable. Theoretical considerations involving retinal irradiance should take into account the fact that pupillary size, although primarily determined by the amount of light entering the eye, is also affected by such factors as drugs, emotional state, and changes of viewing distance. In addition, the pupil undergoes spontaneous changes in size. The pupillary response to changes in illumination is relatively slow; for example, more than 100 msec elapse between the onset of a stimulus and the beginning of a pupillary response.

Delicately suspended just behind the iris is the crystalline *lens*, the posterior surface of which is the last important refractive interface in the optical system of the eye. *Accommodation* is the process by which the shape of the lens is changed, resulting in the decreased focal length that must follow a decrease in viewing distance if the image of the viewed object is to be in focus on the retina. Variables other than viewing distance also can influence accommodation. People tend to lose the power to accommodate, however, as they age. No matter what an individual's age, if he has normal vision (is emmetropic), his lens has a power of 20 diopters in the unaccommodated state. On the average, if the individual is 20 years old, he can increase the power of his lens through accommodation to 31 diopters; if he is 60 years old, he will probably be able to increase it through accommodation to only 22 diopters.

The inside of the eye is filled with a transparent, gelatinous substance called the *vitreous humour*, which has a refractive index equal to that of the aqueous humour (1.336).

Three distinct coats or layers surround the vitreous humour and form the outside shell enclosing most of the eye. The outermost coat is the tough, protective covering called the *sclera;* the "white" of the eye is that part of the sclera which is visible through the conjunctiva at the front of the eye. The middle coat, the *choroid*, is a thin, darkly pigmented membrane, containing many blood vessels that nourish adjacent structures. This heavy pigmentation and rich vascularization of the choroid give rise to some of the characteristics of radiation damage.

The innermost coat, the *retina*, is the most important for the purposes of this paper, for it is the sensitive layer containing the anatomical structures that detect light, encode in nerve impulses the information contained by the image, and relay this information to the brain. An understanding of retinal anatomy is especially important for an understanding of the effects of radiation damage.

As can be seen in Figure 1, light entering the eye first passes through the optical system described above, then passes through the

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conditions under which the first signs of ocular damage appear. By further limiting interest to the "worst case" (i.e., those conditions that are most likely to cause damage), any error or change in conditions is likely to be a conservative error, in that it over-estimates the danger and tends to leave a large margin of safety.

The following treatment of *Damage Thresholds* followed this philosophy. Most of the effects treated in *Possible Mechanisms* have had previous treatment in the literature, but not in just this way. The section on *Variables* begins with standard formulas, available elsewhere, but then departs to show the changes produced by an incremental process of definition and simplification. Various formulas give a range of options, varying in generality and in necessary assumptions, which can serve as the basis for generalizing from experimental tests of damage thresholds or as a basis for theoretical estimates of damage threshold under differing conditions. Finally, the limits on the value of each variable are evaluated on the basis of empirical data, and a theoretical estimate of the damage threshold, and a probable range of error, is given. This entire treatment, and the resulting theoretical estimate, have not previously been published.

Next, the previously published experimental tests of damage threshold are summarized and compared against the theoretical estimate we have derived. Confidence in our estimate is further increased by its correspondence to the conditions under which the eye begins to cease normal functioning, as determined by psychological effects of very intense stimuli (*Relations to Brightness*). This way of at-region near the center of the spheroid formed by the eye, and finally propagates outward from this central region to the retina. Within the retina, light passes through several very thin anatomical layers before it arrives at the receptors, the *rods* and the *cones*. The retina is in a sense inverted, for the first layer through which light passes as it traverses the retina is a layer containing the nerve fibers carrying information from various parts of the retinal image toward the optic nerve. Only after passing through these fibers and several other layers of interconnecting nerve cells does the light reach the vicinity of the photoreceptors (and even here the light passes through most of each receptor cell before reaching the photosensitive part of that cell). Any light not absorbed by the receptors is absorbed by the adjacent pigmented epithelial layer, which lies between the receptors and the choroid, and by the choroid itself. It should be noted that in discussing anatomical location in the eye, the terms "inner" and "outer" are referenced at the center of the eye and not, for example, referenced from the front of the eye or center of the head.

A consequence of these structural relations is that damage to the nerve fibers of the retina lying between the photoreceptors and the source of radiation can functionally disconnect from the brain an area of intact retinal photoreceptors that indeed may be much larger than the area of actual damage and that may lie at great distances from the damaged area. Damage to these fibers can occur by direct absorption of radiation if the wavelengths are long; by conduction of heat from the pigment epithelial layer, where most of the energy in the visible wavelengths is absorbed; or by any of a number of the mechanisms we take up below.

The two types of photoreceptors in the human eye, the *rods* and the *cones*, differ from each other in size, in shape, in distribution within the retina, and in connections with the brain. Although the rods and the cones do not form the bases for two anatomically independent visual systems, under some conditions they may form two functionally distinct systems. The rod receptors of the retina (about 10^8 in number) have diameters of about 2μ, are most sensitive to light of 510 $m\mu$ wavelength, and provide the input to that part of the visual system which responds to very small amounts of light but lacks the highest acuity. On the other hand, the cone receptors (10^7 in number) provide the input to that part of the visual system which is most sensitive to wavelengths of 555 $m\mu$, subserve color vision, are relatively insensitive to low levels of illumination, and provide for the finest form discrimination.

Located about 5 to 6 degrees off the optic axis of the eye, as shown in Figure 1, is the *fovea centralis*, a very small part of the retina containing only very slender, closely packed cone receptors each of which is about 2 μ in diameter. Except at very low illumination levels, this is the region of the retina that mediates the finest acuity, and this is the location of the image of the point in the visual field that is fixed by the eye. With increasing distance from the fovea, the proportion of cones to rods decreases and the size of the cones increases to about 2.5 μ.

It is of interest to relate the size of the receptors to the size of the images perceived. According to geometric optics, the angle subtended by a receptor corresponds to a visual angle of about 25 sec, or that of a penny at a distance of about 125 meters. This, however, is misleading, for even under optimum conditions (Westheimer, 1963), diffraction induced by the pupillary aperture distributes the image of even the smallest source in the form of an Airy disc subtending 1.5 min, which covers an area of 40 μ^2 on the retina, or 10 to 13 foveal receptors. Optical imperfections in the eye further increase this spread.

For further information on the basics of vision that are relevant to this paper, the reader is referred to works by Bartley, 1951; Brindley, 1960; Davson, 1963; Fry, 1959; Graham, 1951; Graham, 1965; Judd, 1951; and LeGrande, 1957.

Characteristics of laser radiation

In conventional sources of light, such as incandescent filaments, the atoms of the source emit light independently of one another. Thus, for example, light from an incandescent source is emitted at all wavelengths of the visible spectrum (420 to 710 $m\mu$), in all possible directions, and in all possible phase relations. Under certain conditions, however, when a laser is operating in the lowest order mode, atoms are stimulated to emit light quanta having nearly the same frequency, the same direction of propagation, and the same phase. Thus, laser radiation can be nearly *monochromatic*. The degree of monochromaticity can correspond to frequency stabilities of one part in 10^{13} for periods of seconds (Townes, 1965) and of one part in 10^9 for periods of months (Schawlow, 1965). Further, laser radiation is usually *collimated*. Deviations from exactly parallel rays are limited, in some lasers, to that caused by diffraction at the exit aperture of the laser. Laser radiation can have *uniform phase*. If a laser is operating in the lowest order mode, the phase is the same at all points in any plane perpendicular to the direction of propagation. Phase is always well defined in a laser beam, no matter in what mode it is operating, but its expression may be complicated. Lasers are the source of the most nearly perfectly *coherent* light, i.e., light that is monochromatic and has a well defined phase relationship through space and time.

Lasers are also capable of *high energy* output. A single pulse may deliver 10^3 J (Schawlow, 1965). In addition, lasers are capable of *great power*. A laser pulse may be shorter than 10^{-9} sec and the power can reach values of 10^9 watts (Schawlow, 1965; Townes, 1965).

tempting to define a damage threshold, by three fundamentally different techniques, is also new to this literature.

A discussion of *Temporal Factors* follows because of its great significance for the thresholds under consideration. Finally, attention is directed to certain kinds of errors that occur in the literature, some of which have previously escaped recognition.

Some visual effects of the special properties of laser radiation

Since laser radiation has special properties, the question naturally arises as to whether or not these special properties lead to special visual effects when the radiation stimulates the eyes.

• *Monochromaticity*

Because the intensity of monochromatic light from lasers can be much greater than that from monochromators, the laser is likely to be a useful research tool in visual physiology.* However, the human eye almost certainly cannot detect the difference in monochromaticity between light from a laser and light from an ordinary monochromator. It is unlikely, therefore, that special visual effects are to be expected from the high monochromaticity of laser radiation.

* It is possible that a photopigment molecule could be activated by absorbing two quanta of half the energy normally required, if the second quantum were absorbed before the electron excited by the first quantum had returned to the ground state. This may account for recent reports that light from a laser may appear green even though its wavelength is twice that which normally looks green.

• *Collimation*

When a highly collimated, monochromatic laser beam is directed into the unaccommodated eye, the optical system of the eye brings the rays into focus on the retina, forming what corresponds to an image of a point source. The resulting increase in irradiance, from cornea to retina, can exceed a factor of 10^6. Although the rays in any collimated beam are focused in a small retinal spot, the image formed by a laser beam may be smaller than is otherwise possible because the rays of a laser beam are better collimated than is typically possible for other sources of comparable power. The small retinal image formed under these conditions and the high power available from some lasers lead to especially hazardous conditions for the eye.

• *Uniform Phase*

The most striking effect of the well-defined phase relationships in laser radiation is the "sparkling effect" referring to the fact that a surface illuminated by a laser has a speckled or unevenly illuminated appearance that fluctuates over time. In addition, this property of laser radiation can now be used as an improved means for producing on the retina interference fringes that are smaller than the resolution normally permitted by diffraction at the pupil (Campbell & Green, 1965; Westheimer, 1960).

• *Energy*

The energy obtainable from lasers can easily cause permanent damage to the eye. That potentiality is treated in considerable detail below.

• *Short Duration*

The very short duration of laser pulses sometimes prompts the question whether a flash can be so brief as not to be visible. This is not possible because the eye responds to the integral, with respect to time, of energy absorbed by the photoreceptors. The time interval over which this integration occurs is referred to as the integration time of the eye. In most cases the integration time can be considered on the order of 10 msec. Thus, if Q is the luminous energy in a flash, $P_{\lambda t}$ is the spectral radiant flux of the source at time t, and K_λ (i.e., 685 V_λ) is the luminous efficiency of light at wavelength λ, then

$$Q = \int_0^{10^{-2}} \int_0^\infty P_{\lambda t} K_\lambda d\lambda dt \,.$$

The effects of the great power generated when the energy from a pulsed laser is compressed into a pulse of short duration are considered in the next section.

Damage thresholds for laser irradiation of the eye

To propose a general model for predicting or describing the effects of laser irradiation of the eye is not an aim of this paper; such attempts are already in the literature (Felstead & Cobbold, 1965; Vos, 1962). Rather, the purpose is to consider a theoretical model restricted to the conditions holding at the *threshold* for permanent ocular damage by laser radiation. So restricting the range of conditions to be considered permits use of a simpler model which is also more likely to be valid within these restricted conditions. (For example, it is so lethal to heat cells to the boiling point that conditions pertaining to tissue temperatures above 100°C need not be considered.)

In the approach adopted below, the various mechanisms whereby laser radiation could conceivably lead to ocular damage are individually considered, and it is shown that, with two exceptions that hold only under special conditions, the temperature of the tissue will rise to the boiling point before the particular mechanism under consideration can cause tissue damage. Therefore, the threshold for tissue damage, under most conditions, is determined only by thermal effects.

The threshold in question here can be defined as a multidimensional surface that separates those combinations of values of the many relevant variables that lead to permanent, detectable visual deficit from those that do not. The threshold must be taken to be multi-dimensional because a systematic change in any one of a number of variables could change the effect of the radiation from harmless to damaging. The validity of the many operational definitions used in experimental investigations of damage thresholds and, consequently, the validity of the conclusions reached in those investigations, depends ultimately upon how close these operational definitions come, in practice, to the definition given above; this is generally unknown.

Some mechanisms inducing threshold tissue damage

The following mechanisms are capable of inducing damaging effects on tissue by laser irradiation but, as will be seen, many have negligible effects under conditions holding near the damage threshold.

Thermal effects—The greatest damage at threshold, as well as at higher levels, is almost inevitably the consequence of heat. Heat can cause damage either by chemical or by mechanical means: to heat tissue above critical levels initiates certain destructive chemical reactions, such as the denaturation of proteins that occurs above 55°C, and it increases the rates of other reactions to a destructive extent (Henriques, 1947); the mechanical effect of heating tissue very rapidly is that the tissue is caused to expand very rapidly, which produces stresses that can damage the tissue structure.

The following discussion leads to the conclusion that the tissue damage caused by *long-lasting* irradiation at threshold values is a consequence of the chemical effects of the resulting heat, but that tissue damage caused by *short pulses* of radiation at threshold levels is a consequence of the vaporization of fluids. Whether expansion caused by temperature increases that do not reach the boiling point can cause damage is not clear, but, if so, it would result only from the shortest pulses and only under the worst conditions.

The threshold for the production of damaging chemical effects in tissue by transient increases in temperature, when these increments in temperature take place within the short periods corresponding to the duration of laser pulses, remains unknown. (The classical theory for thermally induced tissue damage (Henriques, 1947) can be applied with confidence only to durations greater than a second.) Attempts to measure the temperatures reached by tissues at the threshold of damage by lasers (Campbell, Rittler, & Koester, 1963; Noyori, Campbell, Rittler, & Koester, 1963; Jacobson, Najac, & Cooper, 1964; Najac, Cooper, Jacobson, Shamos, & Breitfeller, 1963) have indicated that a temperature rise between 12 and 30°C is sufficient to cause ophthalmoscopically visible lesions. The technical and the theoretical problems associated with such measurements, however, leave these results open to question.

The (temporal) rates of the chemical reactions that cause damage to the tissue increase as the temperature of the tissue increases; therefore, a higher temperature is required to cause a given amount of damage if the duration of supranormal tissue temperatures decreases. For example, if the threshold for damage during an increment lasting 1 sec is 20°C, then the threshold for damage during an increment lasting only 0.5 sec may be 30 or 40°C. This trade-off between duration and temperature occurs only within certain limits. Duration of supranormal tissue temperature is limited only by the rates of the processes whereby the absorbed energy degenerates into heat, which are extremely rapid, and by the rate of cooling, which may be very rapid or very slow. But the temperatures that can enter into this trade-off are limited to those below 100°C, because any excess energy present when the temperature of the tissue reaches that point goes into an explosive vaporization, the destructiveness of which dwarfs that of the chemical reactions. Shock waves produced by the explosion resulting when powerful laser bursts hit the skin, for example, can even cause serious damage to internal organs lying deep beneath the skin without necessarily producing a very great effect on the skin or interposed tissues through which the shock waves passed (Fine & Klein, 1964).

Whether mechanical damage can be caused by the expansion produced by rapidly heating tissue to temperatures below boiling is less clear. The stresses depend upon the time course of the expansion, but this is not given directly by the time course of the laser pulse because the pulse durations of many *Q*-switched lasers are well within the lifetime of excited atoms in many substances (Basov, 1965); thus, an unknown amount of time can elapse between the absorption of light and the conversion of the absorbed energy into forms that lead to thermal expansion. These temporal factors are further complicated by the possibility of photon absorption through indirect transitions (Basov, 1965). A final complication is that little is known about the threshold of tissue damage from these causes.

It can be shown (see Appendix), for example, that the maximum pressure that can be developed by thermal expansion, no matter how rapidly the heat were to be delivered to the tissue, is about 1000 atmospheres, or about 90 dB above the 1 dyne cm^{-2} reference level. Contrary to intuition, pressures of this magnitude may not be very damaging to tissue, for auditory experiments have been reported in which human subjects were exposed to acoustic stimuli of this magnitude (see Licklider, 1951). No damage was reported, but the subjects did report pain beginning at pressures between 50 and 90 dB (Licklider, 1951; Davis, 1951). It may not be valid, however, to draw comparisons between the pressures developed by thermal expansion following absorption of a laser pulse and the pressures within acoustic waves. Whatever mechanical damage is caused by acoustic waves results from successive compression and expansion of the tissue as the waves pass, but in the case considered here, the expansion of the tissue layer that absorbs the radiation produces a shearing action against the directly adjacent, stationary layers.

Electric field—A 10^9 W pulse focused to a spot 10 μ in diameter, such as would exist, in the worst-case, on the retina of an eye into which a laser beam were directed, would create an electrical field of approximately 10^9 V cm^{-1} (Schawlow, 1965; Townes, 1965). Field strengths of this size, which approach that by which valence electrons are held in atoms, disrupt all materials (Prokhorov, 1965; Townes, 1965) even, for example, materials that are normally transparent to the radiation (Bruma, 1964; Hercher, 1964). But a 10^9 W pulse that is of only 10 nsec duration contains enough energy to boil a cylinder of water having a diameter equal to that of the spot formed by the focused beam, and a length of over 500 m. Clearly, the temperature of the tissue that absorbs the energy in laser pulses would be raised to damaging levels by far weaker pulses than those necessary to erect damaging electric fields, and, consequently, the effects of electric fields need not be considered in connection with the damage threshold.

Electrostrictive stress—Electrostrictive stresses of approximately one dyne per kilowatt are also associated with the electric fields (Schawlow, 1965), which indicates that 10^9 W cm^{-2} is required to produce a pressure of one atmosphere. Clearly, the thermal effects take precedence over those of electrostrictive stress too; in fact, it is shown above that pressures produced by thermal expansion under these conditions exceed those that could be caused by electrostrictive stresses.

Radiation pressure—A considerable pressure is also developed on the retina by the momentum of a high energy pulse focused to a spot 10 μ in diameter. The momentum in a laser pulse is equal to the energy contained in it, divided by the speed of light; thus, a 1000 J pulse has a momentum of 0.33 gm cm sec^{-1}. The absorption coefficient of the retina can be assumed to be 10 percent per micron or 10^3 cm^{-1} (see p. 265); thus a 1000 J pulse directed into the eye has sufficient momentum (mass times velocity) to impart a speed exceeding 4000 km sec^{-1} to absorbing tissue having approximately the density of water. But only a 2-erg pulse is sufficient to raise the temperature of the same tissue to the boiling point (see p. 264), and the momentum of a 2-erg pulse corresponds to an imparted speed of less than 1 mm sec^{-1}. Thus, it is seen that the effects of radiation pressure at threshold are also negligible.

Photochemical effects—A more subtle consideration is the photochemical damage caused by laser radiation (see, for example, Rounds, 1965; Rounds, Chamberlain & Okigaki, 1965). Absorption of a photon or of several photons by a molecule that has a sharp absorption bond corresponding to the frequency of the laser radiation can lead to changes in the chemical bonds of the molecule or to their complete rupture. Although the effects of these chemical changes can be hard to detect, they have been observed and reported (Geeraets, Burkhart, & Guerry, 1963). So little is known about photochemical effects on physiological compounds that practically nothing useful can be predicted about these effects. On the other hand, empirical investigation of these effects are hindered and complicated by the difficulty of observing the effects, and by the wavelength specificity. Direct behavioral tests of the functional loss that defines damage may be required to detect the most subtle of these chemical effects.

Conclusion—All the mechanisms that have been proposed in the literature as causes of tissue damage by laser radiation have now been considered above. Although photochemical damage may occur apart from thermal damage, it is likely that the damage that occurs at threshold under most conditions is thermal in nature. If the pulse duration is *relatively long*, the threshold damage would probably be caused by the initiation of damaging chemical reactions or by the increase in rates of other reactions to damaging levels, both of which accompany rises in temperature. If the pulse duration is *relatively short*, however, threshold damage would probably be caused by the mechanical effects of tissue vaporization. And if the pulse duration is *extremely short*, threshold damage may possibly be caused by the mechanical effects of the rapid thermal expansion produced by temperature rises that do not even reach the boiling point.

Variables and relationships among them

In this section the various variables that affect the thresholds for damage to the human eye by lasers are described, and some general formulas are presented that indicate how these variables determine the damage thresholds. These formulas will serve other functions as well. They provide a context for discussion of considerations entering into the uses of lasers involving the human eye. They are also useful in arriving at a theoretical estimate of damage thresholds for a set of circumstances for which an estimate is both simple to compute and of reasonable validity. They also provide a ready approximation of the effects on the eye that result in many situations that commonly arise with the use of lasers. These formulas, of course, are not accurate in every situation without special consideration of the peculiarities of each. The same statement applies to the formulas that have been advanced by others.

To illustrate, it is not yet feasible to provide a completely general and accurate formula for retinal irradiance. On the one hand, geometric optics are not adequate, particularly for treating cross-sectional variations in flux density within the laser beam, such as those that often exist at the exit aperture of the laser (Ham, Williams, Geeraets, Ruffin, & Mueller, 1963), those caused by diffraction, and those caused by aberrations of the eye. On the other hand, the extra effort required by a more rigorous treatment is hardly warranted because of the imperfections and complexity of real eyes, because of our incomplete knowledge of the optical system of the human eye, because of the large variations among eyes of different individuals, and even because of variations within the same eye at different times and at different retinal locations.

For the sake of logical clarity, then, the most general formulas applicable are presented first. Then these formulas are simplified by successively limiting their range of applicability in discrete steps, the justification for each change and the limitations imposed on the generality of the formulas by each simplification being given at each step. Ultimately, this process leads to the specification of a particular set of circumstances for which the estimate of the damage threshold is simple to compute and of reasonable validity. This theoretical estimate is then compared with empirically determined thresholds obtained under the conditions specified, and a probable range of error in the theoretical estimate is given.

General formulas—According to the conclusions of the section on mechanisms, the magnitude of the temperature increment determines to a large extent whether or not damage is likely to occur. The rate of this rise in temperature ($\partial T/\partial t$) at a given point in a tissue is determined by the properties of the tissue and by the balance between the rate at which energy is absorbed (P_{+r}) and the rates at which conduction (P_{cd}), convection (P_{cv}), and radiation (P_{-r}) remove the absorbed energy. Nothing is added to this statement by expressing these facts as a formula, but so doing does succinctly summarize them. In addition, it simplifies future reference to this relationship and the terms involved. Thus,

$$\frac{\partial T}{\partial t} = f_1(P_{+r}) - f_2(P_{cd}) - f_3(P_{cv}) - f_4(P_{-r}) . \quad \text{(I)}$$

The properties of lasers, and of the radiation emitted by them, that constitute variables having relevance here are wavelength (λ), pulse duration (t), diameter of the exit aperture of the laser (D_L), radiant flux (P_L) at the exit aperture, and beam divergence (ϕ). In the case of interest here, in which the laser beam is centered on the eye, spectral transmission of the ocular media (p_λ), pupil size (D_E), distance for which the eye is accommodated (r_a), and point of fixation are all relevant variables.

The most general and most nearly accurate of the formulas that have been proposed for the computation of retinal irradiance, H_r, produced by lasers is one set forth by Solon, Aronson, and Gould (1961):

$$H_r = \frac{4D_E^2 p_\lambda P_L}{\pi(\theta\phi\xi rf)^2}[\min(\theta, \phi, \xi)]^2 ; \quad \text{(II)}$$

if $D_E > (r\phi + D_L)$,

then let $D_E = r\phi + D_L$.

In this: $\theta = D_L/r$; ξ is the angular subtense of the smallest retinal image possible; r is the distance of the exit aperture of the laser from the primary focal point of the eye (the latter is approximately 1.4 cm in front of the cornea; Fry, 1959); f is the focal length of the unaccommodated eye (1.67 cm); and min (θ, ϕ, ξ) is equal to the least of the three angles within the parentheses. The restriction of Formula II simply indicates that if the pupil size is larger than the beam width at the plane of the pupil, then the pupil size should be considered equal to the beam width at this plane.

The angles ϕ and ξ bear further discussion. As indicated in the section on laser properties, beam divergence, ϕ, is usually determined empirically, though in some lasers it approaches the lower theoretical limit determined by diffraction occurring at the exit aperture of the laser; thus, $\phi \geq 2.44\lambda/D_L$.

The angle ξ enters into Formula II only under conditions in which the rays entering the eye can be considered collimated. Under these conditions, the diameter of the smallest possible retinal image is determined by three effects: pupillary diffraction, optical aberrations, and defocusing of the rays by accommodation. The greatest of these three effects determines the value of ξ. If the eye is unaccommodated, ξ is determined by pupil size. When the pupil is large, the effects of optical aberrations are greatest. The aberrations, however, decrease as the pupil constricts, but diffraction increases. In a normal eye, diffraction is the largest effect when the pupil constricts to diameters approaching 2 mm. In that case, $\xi = 2.44\lambda/D_E$.

If the eye is accommodated for an object less than 6 m away, its focal length, f_a, is less than the unaccommodated focal length, f, by an amount Δf.* By using the thin lens formula, the change in focal length may be related to the accommodated focal length, and to the distance between the object and the primary focal point of the eye, r_a, by the equation,

$$r_a \Delta f = f_a^2 . \quad (1)$$

Since the parallel rays of the incident laser beam are focused at the accommodated focal point, the diameter, D_I, of the blur image cast on the retina behind the accommodated focal point is proportional to the distance of the accommodated focal point from the retina. (Accommodation is significant, of course, only if the blurring effect of accommodation is less than that from all other causes.) This proportion, $D_I/\Delta f$, is approximated by the proportion formed by dividing the pupillary diameter by the accommodated focal length, D_E/f_a; i.e.,

$$D_I/\Delta f = D_E/f_a . \quad (2)$$

Combining the relationships of (1) and (2) yields

$$D_I = (D_E f_a)/(r_a) . \quad (3)$$

Then,

$$\xi = D_I/f = (D_E f_a)/(r_a f) . \quad (4)$$

Because the human eye is limited in its power to accommodate, the ratio, f_a/f, never is less than 0.8, and when the eye is not accommodated for objects closer than reading distance (33 cm), this ratio is between 0.97 and 1.0. Therefore, in most situations, the limits imposed on ξ by defocusing of the retinal image can be well approximated by the ratio, D_E/r_a.

Then, in a normal eye with pupillary diameter near 0.2 cm,

$$\xi = \max(2.44\lambda/D_E, D_E/r_a) ; \quad (5)$$

that is, ξ is equal to the greater of the two ratios. It is obvious from these relationships that ξ is limited by diffraction whenever the eye is accommodated for distances exceeding 3 m.†

* Technically, a distinction between primary focal length, f, and secondary focal length, f', should be made. The two are related by the equation, $f = f'/n'$, where n' is the refractive index of the vitreous humour. Wherever f' enters into the equations, however, it is divided by n'. Therefore, use of f serves the same purpose.

† Of course, since the refractive index of the ocular media varies somewhat with wavelength, the unaccommodated focal length will vary also with wavelength. This is to say, eyes will differ in the wavelength to which they are emmetropic in the unaccommodated state. The above discussion treats only emmetropic conditions. Any error introduced by deviations from emmetropia (ammetropias) will be in the conservative direction. That is, emmetropia is the worst case.

Now that the variables having to do with the laser and its radiation have been introduced, it is appropriate to go into the effects of this radiation on tissue temperature. In Formula I, the rate at which energy is absorbed (P_{+r}) at a depth (z) depends upon the irradiance at the surface (H_r) and upon the spectral absorption coefficient (β_λ) of the tissue at the point in question. The effect of the absorbed energy on the rate of change of temperature depends upon the density (ρ) and specific heat (c_p) of the tissue. Then, in Formula I, assuming the tissue is optically homogeneous,

$$f_1(P_{+r}) = \frac{H_r \beta_\lambda e^{-\beta_\lambda z}}{\rho c_p}. \tag{6}$$

The rate of heat conduction (P_{cd}) depends upon the thermal gradients at the point in question ($\nabla^2 T$), upon the thermal conductivity of the tissue (k), and upon the density and specific heat of the tissue. Thus, in Formula I,

$$f_2(P_{cd}) = \frac{k \nabla^2 T}{\rho c_p}. \tag{7}$$

The expressions for heat loss through convection and radiation will be taken up below; the former is known only under certain conditions, and the latter is shown below to be negligible so far as damage thresholds are concerned.

Simplifying assumptions—It has been assumed throughout this section that, so far as the practical implications of hazards to the human are concerned, it is necessary to consider only those conditions that lead to the lowest damage threshold. This permitted a number of simplifications in the presentation of the general formulas, such as the assumption that no optically dense elements lie within the path of the laser beam. The approach permits, as well, some further simplifications of the general formulas.

The first of these further simplifications is that, with the few but important exceptions listed below, of all the possible tissues that a laser might damage, only those within the retina need to be considered here. This is because the optical system of the eye brings the energy in a laser beam to a focus at a higher density in the retina than is likely to exist in any other tissue. It is unlikely, for example, that a pulse of laser radiation that is subthreshold for the retina would damage the iris or skin if it were to impinge there instead of entering the pupil, because the energy would be distributed over a much larger area of the iris or skin than of the retina.

However, serious exceptions to the last statement must be made in the case of the ocular media, i.e., the cornea, lens, and humours. Three facts account for this: (a) the media are cooled only through conduction, whereas the retina is also cooled by circulating blood; (b) the flux densities within the media can attain rather high values, even if not so great as densities within the retina; (c) the media have high optical densities to radiation at wavelengths near the ends of the visible spectrum, which both makes them more vulnerable to damage and at the same time shields the retina from damage.

When the area of the retinal image is less than that of the pupil, flux density within the ocular media is maximum at a point, close to or within the retina, that is called the secondary focal point; when the area of the retinal image exceeds that of the pupil, flux density within the ocular media is maximum at a point near the posterior surface of the lens, called the second nodal point (see Fry, 1959, for details on focal and nodal points of the eye). Of course, these high flux densities affect the media only if the energy is absorbed there. If a sufficient amount of energy in the visible wavelengths is absorbed to heat the media to damaging levels, its occurrence has escaped general notice in the literature (see, for example, Meyer-Schwickerath, 1960), but damaging effects of the ultra-violet and infrared wavelengths are well documented and should be scrupulously avoided.

Ultraviolet radiation shorter than 230 mμ does not penetrate even the "transparent" ocular tissues to depths greater than the thickness of a few cells (Cogan & Kinsey, 1946; Kinsey, 1948). These superficial cells are highly vulnerable to the lethal effects of ultraviolet light in these wavelengths. As little as 15 mJ cm^{-2} can cause temporary damage. This damage is probably due to a photochemical effect on the nucleic acids, which are optically dense to radiation at these wavelengths (Cogan & Kinsey, 1946).

The longer ultraviolet penetrates more deeply, up to 80 percent of the radiation incident on the cornea being absorbed by the lens (Kinsey, 1948). Absorption by the ocular media decreases as wavelength lengthens through the visible spectrum, until absorption rises again at wavelengths greater than 1 μ because of the high absorption in these wavelengths by the water comprising a high percentage of the tissue. The heating caused by the energy absorbed within the ocular media leads to opacities in the cornea and lens (cataracts). The thresholds are not known, partly because the effects may accumulate over long periods of time, perhaps even over years (Duke-Elder, 1954; Smart, 1965).

The *first* simplifying assumption served to limit consideration to events occurring within the retina. Formula I can be simplified now by a *second* assumption, also made and adequately justified in the section on mechanisms, that only temperatures below 100°C need to be considered. Because substances below 100°C radiate a significant amount of heat only at wavelengths longer than 3 μ, and because any aqueous suspension, such as that surrounding the retina, is opaque to these wavelengths, no heat is lost through radiation; thus, $f_4(P_{-r}) = 0$.

A *third* assumption, that heat loss through convection is negligible during pulses of less than 0.3 sec duration, further simplifies formula I by setting $f_3(P_{cv}) = 0$ for $0 < t < 0.3$ sec. This assumption is justified by empirical findings (Geeraets, Williams, Ham, & Guerry, 1962).

Employing those assumptions, Formula I now becomes

$$\frac{\partial T}{\partial t} = \frac{H_r \beta_\lambda e^{-\beta_\lambda z}}{\rho c_p} + \frac{k \nabla^2 T}{\rho c_p}; \quad 430 < \lambda < 700 \text{ m}\mu, \quad 0 < t < 0.3 \text{ sec}, \quad 30 < T < 100^\circ \text{C}. \quad \text{(Ia)}$$

The first assumption, the so-called worst-case assumption, permits a *fourth* simplification, this time in Formula II, which gives retinal irradiance, H_r. Even in its original form, this formula does not represent the general case, for it is based on two assumptions: first, that the axis of the laser beam passes through the center of the pupil; and, second, that the radiant flux is constant across the area of the exit aperture, instead of having the "hot spots" that typically exist in laser beams (Ham et al., 1963). Both of these simplifications, however, are among those permitted by restricting consideration to the worst-case. In the worst-case, all of the energy from the laser enters the eye and is focused into the smallest retinal spot possible. This occurs if a laser with a small exit aperture, equal to the size of the pupil, is positioned close to an unaccommodated eye, in perfect alignment with it. This is likely to happen, for example, if one were tempted to look into the end of the laser.

In this case, then, $\xi = \phi = 2.44\lambda/D_E < D_L/r$; consequently, formula II for retinal irradiance becomes,

$$H_r = \frac{4 D_E^2 p_\lambda P_L}{2.44^2 \lambda^2 \pi f^2}. \quad \text{(IIa)}$$

A *fifth* simplification is permitted by the worst-case approach. The first step in this approach was to limit consideration to that tissue which is most likely to receive the greatest thermal dose, namely, the retina; now, by the same approach, consideration can be further limited to that point within the retina that is most likely to sustain thermal damage. Two considerations lead to the conclusion that this point lies on the axis of the laser beam: (a) the tissue on the axis of the beam, being farthest from the cooler tissues surrounding the irradiated tissues, loses the least amount of heat through conduction; and (b) the irradiance at the center of the Airy disc is over four times the mean irradiance within the entire disc. Formula IIa actually gives mean irradiance for the entire area of the disc. Computation shows that irradiance at the center of the disc is 4.38 times greater.

Of the points on the axis of the beam, the point that attains and maintains the highest temperature depends upon a number of factors, such as the duration of the pulse, but that point always lies between the inner surface of the pigment epithelium ($z = 0$), where the rate of energy absorption is highest ($\beta_\lambda e^{-\beta_\lambda 0}$), and the center of the pigment epithelium, which loses heat through conduction at the lowest rate. In worst-case, however, practically no heat is conducted away from the pigment epithelium during the laser pulse; i.e., $f_2(P_{cd}) = 0$. Calculations have indicated (Ham, Williams, Mueller, Ruffin, Schmidt, Clarke, Vos, & Geeraets, 1965) that $f_2(P_{cd}) = 0$ when pulse durations are less than 25 μsec.

The *sixth and last* assumption to be made here is that the parameters of Formula Ia remain constant during the laser pulse. Then, by combining Formulas Ia and IIa and integrating, the expression for energy delivered is:

$$U = P_L t = \frac{\Delta T \rho c_p 2.44^2 \lambda^2 \pi f^2}{17.5 \beta_\lambda D_E^2 p_\lambda}; \quad 430 < \lambda < 700 \text{ m}\mu; \quad 30 < T < 100^\circ \text{C}; \quad t \leq 25 \ \mu\text{sec}. \quad \text{(III)}$$

Now it becomes simple to compute the minimum energy from a laser that is likely to damage the eye under the conditions specified here. All that remains to be done is to determine the values of the parameters.

Let $\Delta T = 20°$C, because this is below the threshold for chemical damage even for much longer periods of irradiation, and because a 20°C temperature rise distributed over 25 μsec does not lead to very high pressures: the volume of water increases by less than 1 percent as the temperature rises from 35° to 55°C (Hodgman et al., 1960); but at the end of the laser pulse, the boundaries of the pressure wave caused by the stresses of thermal expansion enclose a volume 7.5 cm in diameter, or 7500 times the thickness of the absorbing layer.

Let $\rho = 0.8$ gm cm^{-3} and $c_p = 0.8$ cal gm^{-1} °C^{-1}, representing the worst case of the range of values of 0.8 to 1.0 reported for both quantities.

Let $\lambda = 5 \times 10^{-5}$ cm, because the β_λ is highest for this wavelength. Furthermore, this wavelength is especially suitable as it is in the middle of the visible spectrum.

The focal length of the average unaccommodated eye, f, is about 1.70 cm.

Between 40 and 70 percent of light of 500 mμ wavelength that is incident on the retina is absorbed within the 10 μ thickness of the pigment epithelial layer (Geeraets, Williams, Chan, Ham, Guerry, & Schmidt, 1962). The remaining energy is absorbed within the choroid. Thus, in the worst case, β_{500} equals 10^3 cm^{-1}. These measurements, however, are complicated by the light absorbed by the photopigments. The fact that maximum retinal absorption occurs at 500 mμ suggests that rhodopsin may have accounted for a significant amount of the absorption measured. The exact proportion of the light absorbed by the photosensitive pigments, in contrast to the photostable, melanin pigments, is especially difficult to determine because the amount of photopigment bleached before the measurements and during the measurements is not known. The absorption by the photo-sensitive pigments, particularly in the normally functioning eye, would exceed what

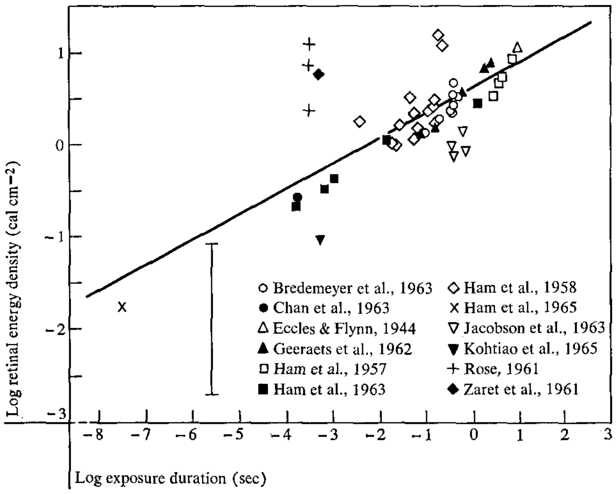

Figure 2 Thresholds for permanent damage to the retina by light irradiation of different durations. The symbols used to plot the data points refer, via the key, to the reports from which the thresholds were taken. Only published reports of experiments that yielded results in terms of areal density of energy on the retina, or that supplied enough information so that the results could be converted into these terms, were used. The bracketed, vertical line represents the range of probable error of the theoretical estimate of worst-case damage threshold (see text). The straight line was fitted by inspection. Justification for use of a straight line is given in the section (page 267) treating distribution of energy over time.

would be expected on the basis of their concentration in the retina because of the waveguide properties of some photoreceptors (cones). These properties funnel a disproportionately high amount of light through the photopigment. However, in the worst case, $\beta_{500} = 10^3$ cm^{-1}.

Let D_E equal 0.3 cm, because a pupil of this size produces the smallest retinal image under present conditions (Westheimer, 1963).

Transmission through the ocular media for this wavelength, p_{500}, in the human lies between 0.50 and 0.95 (Geeraets, Williams, Chan, Ham, Guerry, & Schmidt, 1960; Ludvigh & McCarthy, 1938; Prince, 1962). The validity of the most widely known results (Ludvigh & McCarthy, 1938) has been challenged on the basis of results with animals (see Westheimer, 1963), but wide species differences exist, and the validity of generalizing from work on animals in this matter has been questioned (Duke-Elder, 1954). Since the sensitive locus of damage is the retinal-choroid interface, it is evident the the worst-case situation is the one involving minimum absorption in the other ocular media, i.e., $p_{500} = 0.95$.

Theoretical estimate of damage threshold—Application of Formula III in the light of the preceding discussion leads to a value of 0.05 erg for U. This means that under the very worst conditions possible, the energy in a laser pulse 25 μsec long or longer would have to be less than 0.05 erg if all possibility of damaging the retina is to be avoided. This can be considered an estimated upper limit for completely harmless pulses.

What might be called an estimated lower limit for certainly damaging pulses can be obtained by determining the likely range of error in the above estimate. For the reasons given in preceding sections concerning the effects of boiling in tissue, the value for ΔT can be increased to 63°C; the maximum value of 1.0 can be given to ρ and to c_p; the value of p_{500} can be decreased to 0.5; the value of β_{500} can be decreased to 500 cm^{-1}, corresponding to 40 percent absorption within a 10 μ layer of tissue; and, finally, D_E can be decreased to 0.2 cm, because this is the smallest pupillary diameter that normally occurs, and because only negligible error is introduced by describing the retinal image produced by a pupil of this size as being an Airy disc (Westheimer, 1963).

These changes lead to a 40-fold increase in the estimate of damage threshold energy. Thus, a 2-erg laser pulse of 25 μsec or less at 500 mμ is predicted to cause some retinal damage in the worst case, though its extent might be so limited as to make its detection difficult. Its seriousness, therefore, would also be limited. This, then, we consider to be the lower limit of certain damage.

Application of Formula IIa shows that these two pulse energies correspond to peak energy densities on the retina of 13 and 123 mJ cm^{-2}, respectively, or 3×10^{-3} and 3×10^{-2} cal cm^{-2}.

Although the values just above correspond to worst-case conditions, the worst case has not actually been empirically investigated. Most empirical investigations of damage threshold for lasers have used ruby lasers, which emit light at a wavelength of approximately 690 mμ. Then, in order to bring the theoretical estimate more in line with the worst case actually tested, λ must be given a value of 690 mμ. β_{690} ranges from 160 to 500 cm^{-1} (derived from the data of Geeraets et al., 1962), and the lower limit of p_{690} is 0.7.

These changes increase the estimated lower limit of damaging pulse energy to 0.18 erg, and they increase the estimated upper limit for harmless pulses to 8.4 ergs. The corresponding limits on peak retinal energy density are 8 and 380 mJ cm^{-2}, or 0.002 and 0.090 cal cm^{-2}.

Empirically determined damage thresholds

Figure 2 provides a summary of the published data on radiation damage thresholds for the retina. Many of the points plotted on this graph do not represent the results of laser-induced damage, but those representing the shorter durations do. Most of the data plotted here represent results obtained with roughly comparable criteria for damage (typically, ophthalmoscopically visible blanching of the retina). Some of the deviant points can be traced to anomalies in the particular experiments they represent. For example, Zaret, Breinin, Schmidt, Ripps, Seigel, and Solon

(1961) did not localize the threshold within less than a log unit range (the point is at the upper end of the range), and Kohtiao, Newton, Schwell, and Resnick (1965) based their estimate on a larger retinal spot than has been typically used. The data points in the figure were fitted by inspection with a straight line. The vertical, bracketed line represents the range covered by the upper and lower theoretical limits determined in the preceding section.

Figure 2 shows that the theoretical range is lower than the empirical one. The threshold defined at the outset of this paper, however, might be expected to exist at lower levels than those defined by the criteria represented by these data points (see p. 257). Although some different criteria do yield the same results [e.g., opthalmoscopically visible lesions correspond closely to histologically identifiable lesions (Geeraets et al., 1963)] nevertheless, certain others yield widely differing results. For example, long-term alterations in the electroretinogram are detectable at half the energy required for an ophthalmoscopically visible lesion (McNeer, Ghosh, Geeraets, & Guerry, 1963), and alterations in enzyme activity have been reported at levels a full log unit below the thresholds defined by the most commonly used criterion, namely the ophthalmoscopical criterion (Geeraets et al., 1963).

Relations of damage threshold to brightness

Psychological experiments (Brindley, 1960) show that the brightness of a flash of light continues to increase as the energy in the flash increases, until the energy approaches a value of about 0.15 mJ cm^{-2} at the retina. Further increases in energy do not increase the apparent brightness, but differences in the after-image of the flash are observable until retinal energy density approaches a value of about 15 mJ cm^{-2}. This "saturation" of the after-image occurs in the region where the estimates made above suggest that incipient damage may be limiting retinal responsiveness.

These figures apply to yellow light of about 555 mμ wavelength, to which the eye is most sensitive. When the wavelength of the light is nearer to 630 mμ, the energy must be four times as great for an equivalent effect and when the wavelength is near 690 mμ, then the energy must be about 100 times as great for the same effect. This, of course, is because of the relative insensitivity of the eye to these wavelengths.

Two conclusions follow: On the one hand, nothing is gained, from the standpoint of vision, by attempting to stimulate the eye so strongly that there is risk of injury. All the brightness needed for any task—in fact, all the brightness obtainable—is available at levels well below the threshold for damage; further increases in power or energy above these levels do not increase the response of the eye to the stimulus. On the other hand, the brightness of radiation in wavelengths of 690 mμ or more is so low that brightness begins to lose its effectiveness as an indication of possible danger to the eye. Neodymium lasers afford a highly illustrative case in point. Radiation from these lasers, having a wavelength of approximately 1 μ, is invisible; yet 40 percent of this radiation does penetrate the ocular media and is ultimately absorbed by the retina, probably to a large extent by the neural layers, where damage may occur (Wohlbarsht, Fligsten, & Hayes, 1965).

Distribution of energy over time

The discussion so far has concerned laser pulse durations of approximately 25 μsec. As pulse durations of constant energy increase beyond this duration, not only does the effect of the conduction produced by the thermal gradients become increasingly apparent during the pulse itself, but the thermal gradients themselves become increasingly affected by the size and shape of the retinal image.

As the duration of irradiation increases further, to 0.3 sec or more, where the convective effects of blood flow become significant, even the location of the image on the retina becomes a factor. This is because the blood flow varies in different parts of the retina. For example, blood reaches the fovea only through the choroid, whereas other parts of the retina have a vascular supply on the inner surface of the retina. As a consequence, the fovea may be cooled less efficiently, and it would therefore be especially susceptible to burns. The clinical literature is consistent with this conclusion, but it is hard to tell to what extent the corroboration is due to the greater likelihood of detecting and reporting foveal burns. The effects of convection depend not only upon the vascularization of the tissue, but even upon the state of the circulatory system, which may change in response to thermal stimuli. The threshold for retinal damage by light from the sun during long periods of irradiation in the rabbit eye, approaches an asymptote between 30 and 120 sec at a power between 2.8 and 3.5 W cm^{-2} (Eccles & Flynn, 1944). In worst-case, this corresponds to a laser output of 0.015 mW (Formula IIa), but the comparison is misleading for a number of reasons. *First*, the retina would not be heated to so high a temperature by the small spot cast on the retina by a laser as it would be by an image of the size of that of the sun. Ham et al. (1965) estimate that the temperature in the retinal image of a laser would approach equilibrium within 10 msec and, according to the data of Kohtiao et al. (1965), between 2.5 and 10 times as great a power density would be required within an image of that size in order to produce a lesion. *Second*, after about 0.5 sec, the extent of involuntary eye movements (physiological mystagmus), and consequently the extent of movement of the retinal image, exceeds the retinal area typically covered by the focused beam of a laser (Riggs, Armington, & Ratliff, 1954). Therefore, the full amount of energy emitted by the laser over any particular period of time would not be concentrated within an area of the retina corresponding to the size of the image, but over the entire area of the retina on which the image had impinged during that period of time. Finally, only under very special circumstances would an observer be likely to fixate for a long enough time to provide a stationary image of sufficient duration to come to an equilibrium of the kind implicated in these steady-state studies.

The radiation in the visible wavelengths that reaches the retina, but is not absorbed by the receptor photopigment, is actually absorbed almost exclusively at the surface of minute pigment granules lying within the pigment epithelium and choroid. From the surfaces of these granules the heat is distributed to the surrounding material primarily by conduction. Because the distances between granules is small, the cytoplasm between them rapidly warms to temperatures close to that of their surfaces. The steepest thermal gradients thus typically exist at the boundary between the tissue containing the granules and the surrounding tissue, i.e., between the pigment epithelium and other layers of the retina. Nevertheless, even during the longest pulses, finite gradients exist between the granules. Such gradients must exist if heat flows from the granules to the surrounding cytoplasm whence it goes out to the surrounding tissue.

As pulses of constant energy but of decreasing duration are considered, the distance to which the heat is conducted during the pulse decreases, and the thermal gradients—all thermal gradients—increase. These changes in magnitude of thermal gradients may not cease when the gradients at the border of the pigment epithelium approach limiting values, but it is possible that changes in gradients continue within the cytoplasm surrounding the surfaces of the granules at which the energy is absorbed. These increases in gradients, as a given amount of energy is compressed into shorter and shorter pulses, would be expected to continue until limited by submolecular processes, such as those that determine the lifetimes of excited atoms.

But as the gradients increase with shorter pulses, the total amount of tissue heated decreases. Then, if the same amount of energy is absorbed, and if it is distributed within a smaller mass of tissue, the tissue in which it is contained must rise to a higher temperature. If this occurs even at an intracellular level within the pigment epithelium, one would expect that, as pulse duration continues to decrease, the concentration of energy within the tissue will increase even after the gradients at the boundaries of the pigment epithelium approach limiting values.

The consequence of all this is that the energy density at the threshold for damage may well continue to decrease as pulse duration decreases below 25 μsec. The precise dependence of threshold upon pulse duration cannot be estimated without knowledge of the size and distribution of the pigment granules and of the nature of the energy transferring processes within the absorbing molecules. The thermal gradients are essentially exponential in form, however, which leads to linear relationships between log threshold energy density and log pulse duration, such as are visible in Figure 2 of this paper and in Figure 1 of Ham et al. (1965).

Common errors made in evaluating laser hazards

Miscalculations of retinal irradiance from lasers—Since some serious errors have entered into earlier calculations of retinal irradiation by lasers, it will be worthwhile to examine their character and significance. Frequently, even the work of those well acquainted with optical phenomena has been subject to a class of error that leads to an underestimate of retinal irradiance. In one case (Smoyer, 1963), this error entered through the use of a formula which, on the one hand, depends on the assumption that the retinal image size continues to become smaller and smaller as the observer gets farther and farther from the laser, but, on the other hand, depends on the assumption that the retinal flux of the laser remains constant as distance changes. The former assumption is true for the coherent light of lasers only when $\phi > D_L/r > 2.44\lambda/D_E$, which is the case only when the laser is distant from the observer; the latter assumption, on the other hand, is true only when the laser is near the eye (see Solon et al., 1961). This error can lead to an underestimate of retinal irradiance by a factor of one million and also to the erroneous conclusion that the further the laser is from the eye, the more damaging it is (*Laser Focus*, 1965; Smoyer, 1963). Concerning the latter conclusion, it may be useful for those who tend to make this error to visualize the appearance of the laser aperture, in the near field of Solon et al. (1961), as being dark except for an extremely bright star in the center. It should be noted that the alleged magnitude of increase in retinal irradiance with an increase in distance between the observer and the laser certainly cannot be accounted for by accommodation. It was pointed out in the previous discussion of accommodation that there is little difference in focal length between an unaccommodated eye and an eye accommodated for a laser more than 33 cm away. Formula II shows that the maximum difference that could be expected on this basis is on the order of 6 percent.

Many of the errors in estimates of retinal irradiance are due to inaccuracies rather than to mistakes. Sometimes, however, the magnitude of an inaccuracy consciously incorporated is not fully appreciated. For example, it is common practice to assume the Airy disc is uniformly irradiated, though it is well known that the radiant flux is concentrated at the center; it is seldom appreciated that this practice introduces an error, or inaccuracy, of 438 per cent, as can be verified by straight-forward computation.

Less substantial inaccuracies are introduced by other approximations, such as ignoring the effects of variation in retinal irradiance produced by interference patterns (the sparkling effect) and failing to consider the effects that result when light passes through successive apertures, such as Formula 13a of Solon et al. (1961) and Formula II of this paper do for the near field case. Variations in laser power during the pulse should also be noted, not only because they certainly affect the damage threshold, but also because it is difficult to calibrate and measure these changes even with an accuracy of 20 percent (Killick, Batemen, & de la Perrelle, 1965).

It is worth repeating that Solon et al. (1961) give the general formula for retinal irradiance that involves least error. The formula, rewritten here as Formula II, is subsequently modified.

Biological factors—Considerable importance is often attached to the size of the pupil and to the effect of light and dark adaptation thereon. The full range of pupil sizes hardly affects retinal irradiance by more than a factor of 10, and normal variation seldom approaches the full range possible. Increases in pupil size, when the pupil is already large, are compensated to some extent by aberrations that spread the increased amount of light admitted to the eye over a larger retinal area. In the introductory section on the eye, it was pointed out that several other variables lead to variations in pupil size. This, however, probably does not lead to such serious conceptual errors as ignoring other facts of retinal anatomy, such as the fact that the neural connections of the receptors lie between the receptors and incoming radiation.

Light destined for stimulation of the receptors first passes through the nerve cells that transmit the information contained in the light. Fibers found at any given location in the retina may originate at great distances from that point. A lesion at one point in the retina, then, can blind a much larger portion of the eye; a lesion at the optic disc, or blind spot, where all of the nerve fibers leave the eye to form the optic nerve (Figure 1), can blind the entire eye. Perhaps it is also necessary to point out that one need not look directly at, or fixate on, a dangerous source, such as the sun or a laser, in order to sustain retinal damage, since objects in the visual field that are not fixated certainly do form retinal images in the periphery of the eye.

Safety precautions

A number of papers have given rather substantial attention to procedures for the prevention of ocular damage. Likewise, a number of protective devices for protecting against ocular damage are available, some of which have been shown to have rather severe shortcomings. (Straub, 1963, 1965a, 1965b; Goldman & Hornby, 1965; Swope & Koester, 1965). It should be noted that accidental retinal burns produced by lasers have been reported in papers by Rathkey (1965), and by Rounds (1965). A bibliography of papers concerned with safety precautions is contained within a larger bibliography of publications on vision and lasers. This listing is not included with the present paper, but will be attached to the reprints, available on request from the authors or the editors of this journal.

Conclusions

The purpose of this paper has been to present a set of organized facts and interpretation rather than to arrive at explicit conclusions. Nevertheless, it is desirable to present two culminating observations.

(1) Few novel visual effects would be expected from laser radiation except for the increased likelihood of retinal damage. Aside from the sparkling effect, laser radiation is likely to affect the visual system approximately the same as does any other colored light that has been collimated.

(2) It is fairly certain that laser pulses of less than 0.05 erg are incapable of producing permanent impairment of visual function. In general, one can suspect that if a very short duration (less than 25 μsec), well collimated pulse of 380 mJ or more from a ruby laser were to enter an emmetropic eye, some permanent damage would probably be done to the retina of the eye, but there is no telling how easily this damage could be detected.

Appendix

The following considerations lead to the conclusion that, so long as tissue temperatures do not rise above the vaporization point, the maximum pressures that could be produced within tissue by thermal expansion following extremely short pulses of irradiation, are about 1000 dyne cm^{-2}, no matter how rapidly the energy were to be delivered and converted into heat.

Assume that, except for optical density, tissue has approximately the characteristics of water. For the present, also assume that the proportion of the energy absorbed by the tissues that is expended in acoustic shock waves is so small relative to that necessary to raise the temperature of the tissue that it can be ignored in calculating changes in tissue temperature. Although this assumption begs the question to some extent, its justification is given below, where it is clear that the conclusions made here do not depend upon the circularity of this assumption.

Then, if a given volume of tissue were to absorb enough energy to raise its temperature to 100°C, and if that energy were instantaneously converted to heat, the increase in pressure would correspond to that which would be produced by an isothermal compression of the same mass of water by an amount equal to the difference in the volumes of the water, at constant pressure, at the two different temperatures.

The relationship between the isothermal changes in pressure (p) and volume (V) of liquids is expressed by the bulk modulus,

$$M_B = -Vdp/dV.$$

But M_B varies with pressure according to the relation,

$$M_B = (L + p)/C\,,$$

where L and C are empirical coefficients that are approximately constant over the temperature interval of concern here ($30° < T < 100°C$), wherein $L = 3.05 \times 10^9$ dyne cm^{-2}, and $C = 0.1368$ (Hodgman, Weast, & Selby, 1960). Then,

$$\int_{V_1}^{V_2} \frac{dV}{V} = -C \int_{p_1}^{p_2} \frac{dp}{L + p}\,.$$

The pressure, p_2, is equal to that produced by compressing, at a constant temperature of 100°C, a given mass of water from V_1, its volume, at atmospheric pressure, p_1, to V_2, the volume it would occupy at 30°C. Integration and rearrangement of terms yields

$$p_2 = (L + p_1)(V_1/V_2)^{1/C} - L\,.$$

But (see Hodgman et al., 1960)

$$V_1/V_2 = 1.039\,,$$

and

$$p_1 = 10^6 \text{ dyne cm}^{-2}\,;$$

consequently,

$$p_2 = 10^9 \text{ dyne cm}^{-2} = 10^3 \text{ atm}\,.$$

Thus, so long as the tissue constituents do not vaporize, the maximum pressure that can be developed by thermal expansion, no matter how rapidly the heat were to be delivered, is 1000 atmospheres, or about 90 dB above the 1 dyne cm^{-2} reference level.

It is possible now to demonstrate the validity of the assumption, made above, that the proportion of the energy absorbed into acoustic shock waves does not substantially affect estimates of temperature changes. The amount of energy required to heat tissue to 100°C is given by the formula,

$$U_H = \int_{30}^{100} \rho c_p V_0 dT = 70 V_0 \text{ cal} = 220 V_0 J\,,$$

where ρ is density, c_p is specific heat, and V_0 is initial volume when $T = 30°C$. The energy expended in acoustic waves is given by the formula,

$$U_A = \int_{r_0}^{r_1} psdr = \int_{V_0}^{1.039V_0} pdV\,,$$

where s is surface area, and dr is infinitesimal distance normal to the surface. Letting p assume a constant value equal to the maximum possible pressure under these conditions, (viz., 10^9 dyne cm^{-2}) simplifies the integration and, moreover, introduces a conservative bias into the result by overestimating U_A. Then

$$U_A = p \int_{V_0}^{1.039V_0} dV = 3.9 V_0 J\,.$$

Therefore,

$$U_A/U_H = 0.013\,;$$

i.e., the amount of energy expended in acoustic waves under the conditions considered here cannot exceed about one per cent of the energy required to heat the absorbing tissue to 100°C.

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