Optics in Medicine

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13.1 Introduction

13.1.1 Why Optics in Medicine?

Unlike the present time, medical practitioners of the ancient world did not have the benefit of sophisticated instrumentation and diagnostic systems, such as X-rays, ultrasound machines, or CT scanners. Visual and manual auscultations were the tools of the day. Hence, since the early days of medicine, optics has been a useful and powerful technology to assist doctors and all forms of healthcare practitioners carry out examination and diagnosis of their patients. This is so because one of the fundamental aspects of medicine is observation and physical examination of the patience's general appearance. Hence, anything that can help "see" better the condition of a patient will be of aid. As such, optics, as the science that studies the behavior and manipulation of light and images, is an ideal tool to assist doctors gain better visual examination capabilities by providing improved illumination, magnification, access to small or internal body cavities, among others. But it is in reality light and its interaction with living tissues that is at the center of what makes optics in medicine possible. Light possesses energy and is capable of interacting with biological cells, tissues, and organs. Such interaction can be used to probe the state of such living matter for diagnostics and analytical purposes or, it could be used to induce changes on the same living systems and be exploited for therapeutic purposes. The science of light generation, manipulation, transmission, and measurement is known as *photonics*. The application of photonics technologies and principles to medicine and life sciences is known as *biophotonics*.

Nowadays, it is not only optics but also photonics that are used extensively in a myriad of medical applications, from diagnostics, to therapeutics, to surgical procedures. Hence, when we use the term medical optics, we are referring to biomedical optics and biophotonics as well. The interrelation between optics and light in medicine is ever present and it could be said that more significant advances in biophotonics are now due to the availability of more powerful, concentrated, and multi-spectral light sources which have been available only in the last 50 years. Historically, ambient light was the illumination source, which precluded performing exams late in the day or during certain hours in the winter time. Oil candles in the ancient world gave way to wax ones and alcohol burning lamps in the fifteenth through the nineteenth centuries until the development of electricity and the introduction of the electric lamp by Edison. Then, in the 1960s, with the development of semiconductor lasers, light emitting diodes (LEDs) and lasers, modern medical optics began to take shape and, coupled with the availability of optical fibers, a new generation of medical instruments and techniques began to be developed.

Fiber optics has been used in the medical industry even before their adoption and subsequent explosion as the technology of choice for long haul data communications [1]. The advantages of optical fibers have been recognized by the medical community long ago. Optical fibers are thin, flexible, dielectric (non-conductive), immune to electromagnetic interference, chemically inert, non-toxic, and of course, small in size. They can also be sterilized using standard medical sterilization techniques. Their major advantage lies in the fact that they are thin and flexible so they can be introduced into the body for both remotely sense, image and treat. Their initial and still most successful biological/biomedical application has been in the field of endoscopic imaging. Prior to the development of such devices, the only method of inspecting the interior of the body was through invasive surgery. Many patients owe their lives today to the existence of fiberoptic endoscopes. Optical fibers are not only useful for endoscopes, but can also be used to transmit light to tissue regions of interest either to illuminate the tissue so that it

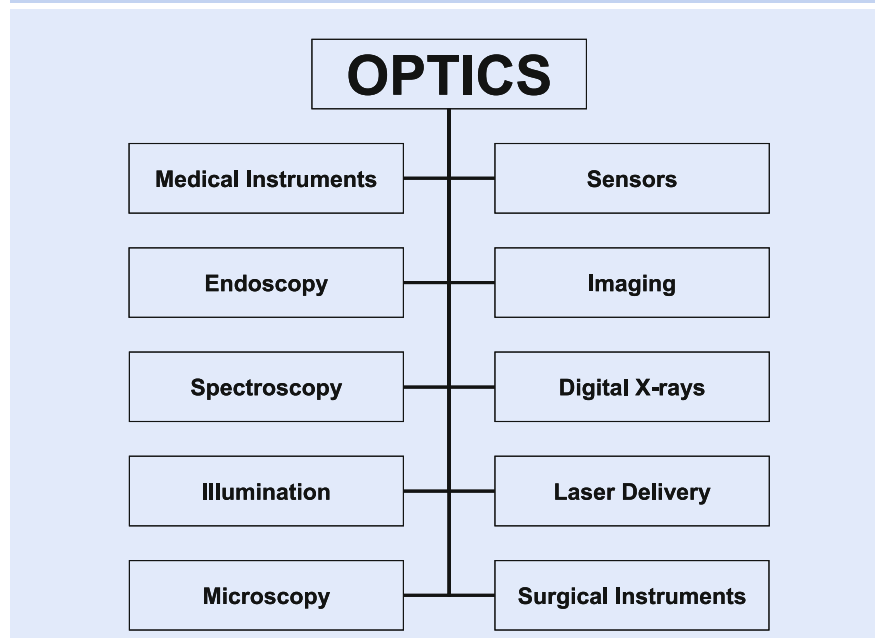| ▣ **Table 13.1** Medical industry trends that promote the use of optical fibers |
| --- |
| • Drives towards minimally invasive surgery (MIS)→Need for disposable probes and catheters
 • Miniaturization, Automation and Robotics→Need for instrumented catheters |
| • Sensors compatible with MRI, CT, PET equipment as well as thermal ablative treatments involving RF or microwave radiation→Need for fiber sensors |
| • Increased user of lasers→Need for fiber delivery devices
 • Increased use of optical imaging and scanning techniques→Need for fiber OCT probes |

can be inspected, or if much higher power laser light is used, to directly cut or ablate it. Hence, they are used extensively as laser-delivery probes, as well as imaging conduits in optical coherence tomography (OCT).

Optical fibers have revolutionized medicine in many ways and continue to do so thanks to the advent of new surgical trends, as summarized in ▣ Table 13.1. One such trend is the advent of minimally invasive surgery (MIS) where the trend now is to avoid cutting open patients and instead, perform small cuts and incisions through which a variety of different surgical instruments, such as catheters and probes, are inserted through these small opening, thus minimizing the postoperative pain and discomfort. Furthermore, there is today growing use of surgical robots where a surgeon operates them remotely using control arms to do a surgical procedure from the comfort of his office while the patient is at a remote hospital location. However, one of the issues with these types of systems is the fact that the surgeon loses the actual manual feedback and does not have sensitivity of the force needed to apply to a scalpel or other surgical tools. This is called *haptic* feedback. These "robotic surgeons" operate using very small tools and catheters, and in order to make sensing elements compatible with such slender instruments, fiber optics represent an ideal solution to provide shape, position, as well as force-sensing information to the remote surgeon's controls.

Fiber optic and photonic devices are also being exploited as sensing devices for patient monitoring during medical imaging and treatment using radiation devices such as MRI, CT, and PET type scan systems that involve the use of high-intensity electromagnetic fields, radiofrequencies, or microwave signals. Because the patient's risk of an electric shock conventional electronic monitoring devices and instrumentation cannot be used in these applications. Instead, patient monitoring is performed using optical fiber sensors.

Based on the above arguments it becomes evident the need for and benefits of optics (and photonics) in medicine. ▣ Table 13.2 summarizes the key general applications for optics in medicine. In general, it could be said that optics has been and will continue to be an enabling technology to further the development and advancement of medicine and the healthcare industry as a whole.

Table 13.2 Typical applications of optics in medicine

OPTICS	
Medical Instruments	Sensors
Endoscopy	Imaging
Spectroscopy	Digital X-rays
Illumination	Laser Delivery
Microscopy	Surgical Instruments

13.1.2 Global Healthcare Needs and Drivers

We all need medical care, from the day we are born, until the day we die. However, this need for medical care has now been affected and accentuated by a convergence of social, demographic, economic, environmental, and political global trends that have been developing over the last few decades. It is a world full of challenges that impact how to effective deliver healthcare in an effective, affordable, and sustainable fashion. On one hand, average lifestyles have changed drastically in the past century resulting in a more sedentary lifestyle with lack of exercise, poor diet, smoking, and excessive alcohol consumption that have resulted in a growing number of chronic diseases such as obesity, arteriosclerosis, diabetes, and cancer that have become leading causes of death and disability. On the other hand, the entire global population keeps growing. According to a recent United Nations Department of Economic and Social Affairs (DESA) report the world's population is estimated to be in excess of 7.3 billion and growing at ~1.1 % annual rate, and expected to reach 8.5 billion by 2030, 9.7 billion in 2050, and 11.2 billion in 2100 [2]. As illustrated in ◘ Fig. 13.1, the world population has experienced continuous growth since the end of the Great Famine and the Black Death back in 1350, when the total population stood at merely 370 million. Nowadays, total annual births are approximately 135 million/year, while deaths are around 56 million/year, but expected to increase to 80 million/year by 2040.

Add to this the fact that in certain parts of the world the population is aging, including the USA, Japan, and parts of Europe. Globally, the number of persons aged 65 or older is expected to reach to nearly 1.5 billion by 2050. An aging population puts additional demands on healthcare since older people are more vulnerable to illness and chronic diseases. Furthermore, life expectancy at birth has increased significantly. The UN DESA estimates a 6-year average gain in life expectancy among the poorest countries, from 56 years in 2000–2005 to 62 years in 2010–2015, which is roughly double the increase recorded for the rest of the world. Another key trend and global challenge is the expected shortage of medical doctors and physicians available to meet the healthcare needs of a growing world population.
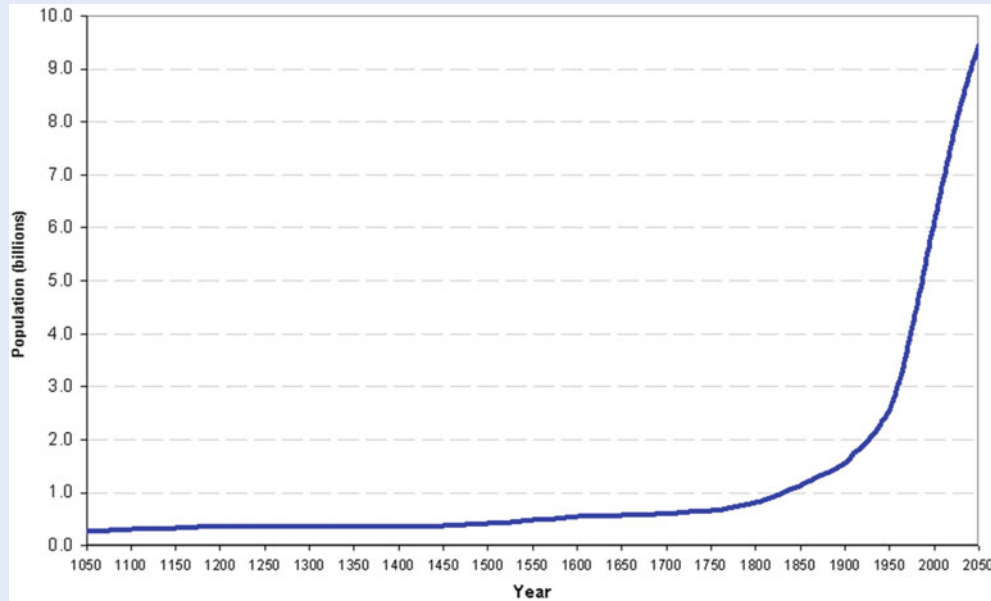
◘ Fig. 13.1 Historical global population growth

A large global population requires more doctors, medical devices, medical supplies, clinics, hospitals, and overall healthcare infrastructure to address the needs of people needing immunizations, or getting sick or injured. Hence, there is and will continue to be an overall growth and expansion of the health care industry on a global basis, that continuous to demand more medical instruments and technical innovations that can facilitate and expedite medical examinations, while reducing costs. Historically, optics has been an enabling technology for the design and development of such medical devices and instruments.

Another relevant and converging present trend is how biomedical devices and instruments are so extremely pervasive across the healthcare industry today. We may not realize it, but whenever we get our blood pressure tested, monitor our blood sugar, or when a expectant mother is being monitored by her doctor, an instrument or sensing device is needed which, often times, is based on the use of an optical technique or based on the use of optical components. Couple this with the fact that in many parts of the underdeveloped world there is not enough doctors, hospitals, clinics, and instrumentation available to support local populations. Hence, it becomes critically important to develop simple, practical, effective, and inexpensive medical devices that can be used in rural and remote areas by non-professionals to examine and treat patients.

13.1.3 Historical Uses of Optics in Medicine

Mankind has always been fascinated with light and the miracle of vision, dating back to the first century when the Romans were investigating the use of glass and how viewing objects through it, made the objects appear larger. However, most of the significant developments of optics for medical diagnosis and therapy started occurring in the nineteenth century. Before that, the vast majority of the known published works on optics and medicine dealt mostly with the anatomy and physiology of the human eye. For instance, the Greek anatomist, Claudius Galen (130–201) provided early anatomical descriptions of the structure of the human

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Fig. 13.2 Photograph of the Jansen compound microscope (c. 1595)

eye, describing the retina, iris, cornea, tear ducts, and other structures as well as defining for the first time the two eye fluids: the vitreous and aqueous humors. Subsequently, Arab scholars Yaqub ibn Ishaq al-Kindi (801–873) and Abu Zayd Hunayn ibn Ishaq alIbadi (808–873) provided a more comprehensive study of the eye in the ninth century in their *Ten Treatises on the Eye and the Book of the Questions of the Eye*. In the eleventh century Abu Ali al-Hasan ibn al-Haytham (965–1040)—known as Alhazen—also provided descriptions of the eye's anatomy in his Book of Optics (Kitab al-Manazir).

It is around this time that the so-called reading stones are being used as magnifying lenses to help read manuscripts. The English philosopher Robert Bacon (1214–1294) described in 1268 in his *Opus Majus* the mechanics of a glass instrument placed in front of his eyes. Then, in the thirteenth century, Salvino D'Armate from Italy made the first eye glass, providing the wearer with an element of magnification to one eye.

With the advent of the optical telescope optics took a significant step forward towards the development of one of the first early medical instruments—the microscope [3]. The compound microscope was developed around the late 1590s by Hans and Zacharias Janssen, a father and son team of Dutch spectacle makers, who experimented with lenses by placing them in series inside a tube and discovered that the object near the end of the tube appeared greatly enlarged (see Fig. 13.2).

A seminal optical medical instrument development came in 1804 when the German born physician Philipp Bozzini (1773–1809) developed and first publicized his so-called light conductor (*Lichtleiter*), which enabled the direct view into the living body [4]. The *lichtleiter* was an early form of endoscope which consisted of an open tube with a 45° mirror mounted at the proximal end with a hole in it. Illumination was provided by a burning alcohol and turpentine lamp was shone to a speculum mounted on the distal end and made to fit to the specific anatomy of the desired body opening to be inspected (see Fig. 13.3). In December 1806 Bozzini's light conductor was presented to the professors of the Josephinum, the "Medical-Surgical Joseph's Academy" in Vienna.

A period of significant activity and innovation in medical optics occurred from the mid-1800s through the early 1900s, when a variety of early medical instruments such as otoscopes, ophthalmoscopes, retinoscopes, and others, as well as improved illumination systems were developed. In 1851 German scientist and physician Hermann L. F. von Helmholtz (1821–1894) used a mirror with a tiny aperture (opening) to shine a beam of light into the inside of the eyeball [5]. Helmholtz found that looking through the lens into the back of the eye only produced a red reflection. To improve on the image quality, he used a condenser

⊡ **Fig. 13.3** Bozzini's original light conductor with specula (c. 1806)

⊡ **Fig. 13.4** Helmholtz ophthalmoscope (c. 1851)

lens that produced a 5× magnification (⊡ Fig. 13.4). He called this combination of a mirror and condenser lens an *Augenspiegel* (eye mirror).

The term ophthalmoscope (eye-observer) did not come into common use until later. Helmholtz also invented the ophthalmometer, which was used to measure the curvature of the eye. In addition, Helmholtz studied color blindness and the speed of nervous impulses. He also wrote the classic Handbook of Physiological Optics.

In 1888 Prof. Reuss and Dr. Roth of Vienna used bent solid glass rods to illuminate body cavities for dentistry and surgery. This would be the earliest idea to

use a precursor of an optical fiber for medical applications. Decades later, in 1926, J. L. Baird of England and Clarence W. Hansell of the RCA Rocy Point Labs, propose independently of each other fiber optic bundles as imaging devices. A few years later, German medical student Heinrich Lamm assembles the first bundles of transparent optical fibers to carry the image from a filament lamp, but is denied a patent. Then in 1949, Danish researchers Holger M. Hansen and Abraham C. S. van Heel begin investigating image transmission using bundles of parallel glass fibers. Prof. Harold H. Hopkins from Imperial College in London begins work in 1952 to develop an endoscope based on bundles of glass fibers. University of Michigan Medical professor Basil Hirschowitz visits Imperial College in 1954 to discuss with Prof. Hopkins and graduate student, Narinder Kapany, about their ideas for imaging fiber bundles. Hirschowitz hires undergraduate student Larry Curtis to develop a fiber optic endoscope at the University of Michigan. Curtis fabricates the first clad optical fiber from a rod-in-tube glass drawing process. Prof. Hirschowitz tests first prototype fiber optic endoscope using clad fibers in February of 1957, and then introduces it to the American Gastroscopic Society in May of the same year.

The first solid-state laser was built in 1960 by Dr. T. H. Maiman at Hughes Aircraft Company. Within the year, Dr. Leon Goldman, chairman of the Department of Dermatology at the University of Cincinnati, began his research on the use of lasers for medical applications and later established a laser technology laboratory at the school's Medical Center. Dr. Goldman is known as the "father of laser medicine." He is also the founder of the American Society for Lasers in Medicine and Surgery [6]. However, the first medical treatment using a laser on a human patient was performed in December 1961 by Dr. Charles J. Campbell of the Institute of Ophthalmology at Columbia–Presbyterian Medical Center, who used a ruby laser that is used to destroy a retinal tumor. Since then, lasers have become an integral part of modern medicine [7].

During the 1980s and 1990s, extensive research was conducted to develop fiber-optic-based chemical and biological sensors for diverse medical applications [8].

OCT is a newer optical medical imaging technique, first introduced in the early 1990s, that uses light to capture micrometer resolution, three-dimensional images from within biological tissue based on low-coherence, and optical interferometry [9]. OCT is a technique that makes possible to take sub-surface images of tissues with micrometer resolution. It can be thought of as the optical equivalent of an ultrasound scanning system. This is an active area of medical research at the moment.

13.1.4 Future Trends

Optics and photonics, as mentioned earlier, are powerful, versatile, and enabling technologies for the development of present and future generations of medical devices, instruments, and techniques for diagnostic, therapy, and surgical applications.

Given the present R&D activity worldwide based on optical and photonic techniques it should be no surprise to expect a broader utilization of optically based solutions across the healthcare industry and medical profession. In the future, advances in the development of ever smaller and thinner medical probes and catheters should be expected, as well as broad utilization of OCT devices to become as common as ultrasound scanning devices are in today's society. There will also be a proliferation of laser-based treatments and therapies. Endoscopy, for its part, will continue to evolve and more sophisticated and smaller devices will be
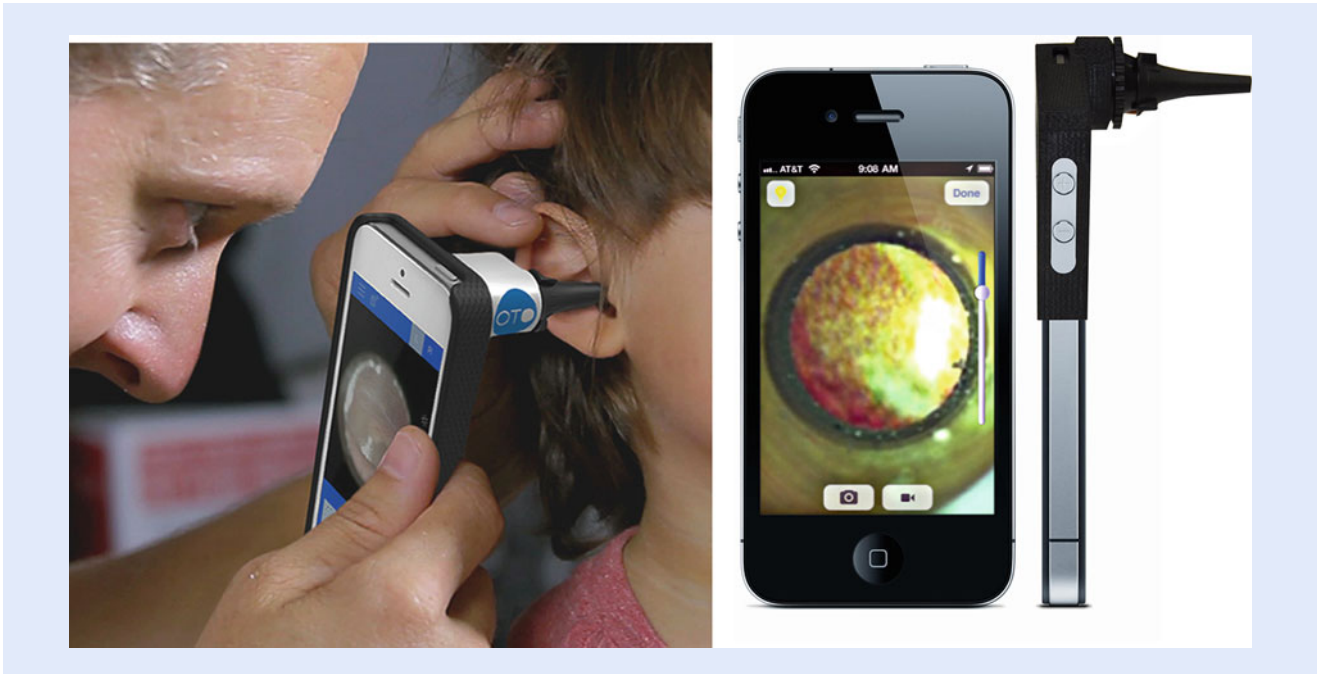
▣ **Fig. 13.5** A smartphone otoscope

developed that will combine more functions (from the standard illumination and visualization) with direct tissue analysis and laser treatment. Optical imaging techniques will continue to advance along with digital X-rays to make non-invasive examination and diagnosis safe, fast and with greater resolution and pinpoint accuracy.

Other future capabilities brought on by optics will be in the form of the so-called lab-on-a-fiber or LOF for short [10], where optical fibers are combined with micro- and nano-sized functionalized materials that react to specific physical, chemical, or biological external effects and can thus serve as elements to build multi-function, multi-parameter sensing devices. Light would remotely excite the functionalized materials which are embedded in the fiber's coating material. These materials in turn will react to specific biological or chemical substances (*analytes*) and induce an optical signal change proportional to the given analyte concentration.

Some future innovations can already be witnessed today in the form of optical devices used in combination with smart portable cell phones [11, 12]. For example, several new companies have now developed accessories for attachment to smartphones, which turn them into electronic video equivalents of conventional medical examination instruments such as otoscopes (to view inside ears), ophthalmoscopes (to view the inside of eyes), or even simple microscopes. Such devices are passive, optical elements that couple images from the patient to the video lens onto the smartphone's digital camera transform it into a fully function-ing, network-connected medical instrument, capable of sending images and video remotely to a consulting doctor. ▣ Figure 13.5 depicts a cell phone otoscope in use, while ▣ Fig. 13.6 depicts a smartphone version of an ophthalmoscope and a dermal loupe.

Another such smartphone innovation is the so-called CellScope developed by researchers at the University of California at Berkeley [13]. The CellScope is a microscope that attaches to a camera-equipped cell phone and produces two kinds
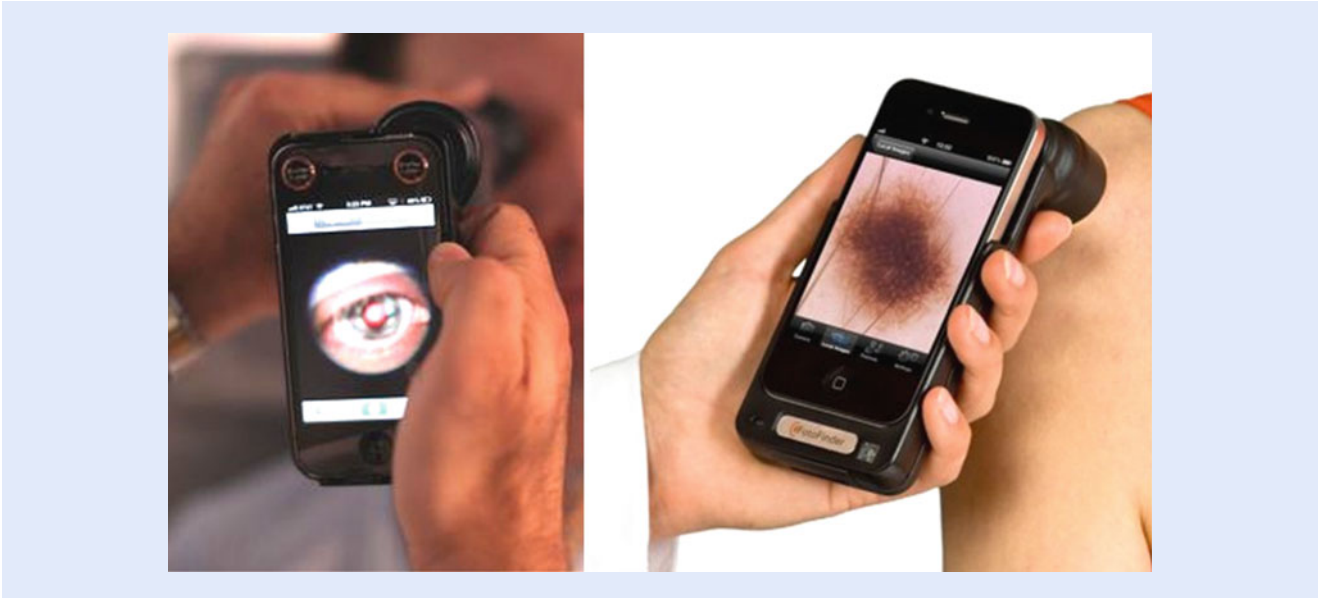
Fig. 13.6 Examples of an ophthalmoscope and a dermal loupe attached to a smartphone

of microscopy imaging: brightfield and fluorescence. The idea is that such device can then be used in the field (on remote locations or those where little medical infrastructure is available) and take snap magnified pictures of disease samples and transmit them to medical labs via mobile communication networks, and screen for hematologic and infectious diseases in areas that lack access to advanced analytical equipment.

13.2   Early and Traditional Medical Optical Instruments

As discussed earlier, optics has been used throughout the centuries as a technology to assist medical doctors perform examinations of patients. Many of the medical instruments in use today rely on optics and optical components to perform their intended function. In particular, there a set of very basic but very popular and common medical instruments that were developed in the nineteenth century and continue to be used in the medical profession of today. Among these optical instruments we have the *otoscope*, the *ophthalmoscope*, *retinoscope*, *laryngoscope*, and even basic devices such as the head mirror.

In general, many of the basic optical medical instruments have in common the goal to provide both a more direct illumination and optical magnification of the area under examination. Conceptually, these optical instruments are similar to a telescope or microscope, but their optical design is different. Typically, a medical instrument consists of a tubular structure fitted with an objective lens on the distal (patient) end, and an objective lens on the viewing (doctor) end, represented as (1) and (2) in ◘ Fig. 13.7.

This lens arrangement produces a magnification of the object under inspection on the objective side (distal end), which has a size Y, and is positioned a distance P from the entrance pupil of the objective lens. The visual magnification factor M_v is calculated as Eq. (13.1):
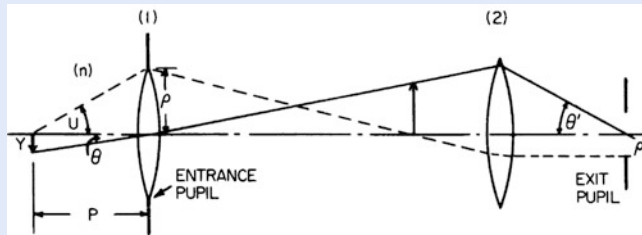
$$M_\mathrm{v} = \theta' D / Y \tag{13.1}$$

Fig. 13.7 Schematic of a basic medical optical instrument
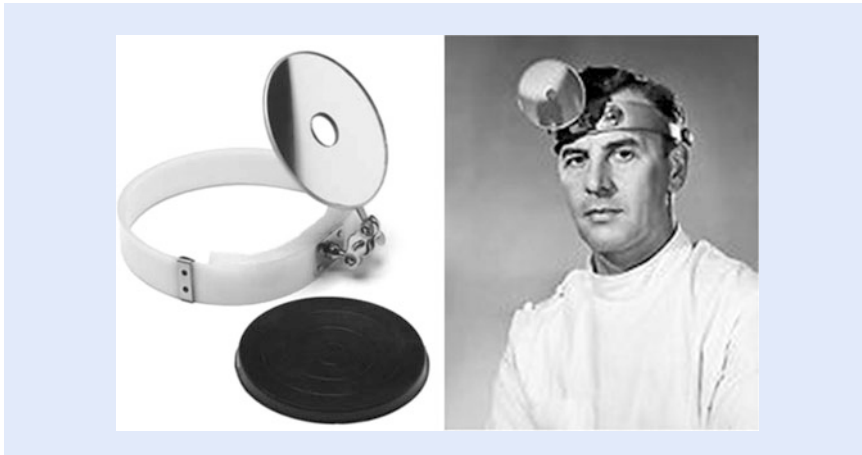

Fig. 13.8 A medical head mirror and common placement on a doctor's head

where θ' is the angle of the light ray from the eyepiece, D is the viewing distance from the observer to the eyepiece. Hence, the magnification factor M is inversely proportional to the working distance P.

In the sections to follow, we shall describe the basic optical operating principles and uses of such devices. Our discussion of these devices is by no means exhaustive, but is intended to provide the reader with an overall idea on the utilization of optics in medicine and brief introduction on the subject of medical optical instruments [14].

13.2.1 Head Mirror

The most basic optical medical instrument is the so-called head mirror (see Fig. 13.8). A head mirror has historically been used by doctors since the eighteenth century for examination of the ear, nose, and throat. It consists of simple circular concave mirror—made of glass, plastic, or metal—with a small opening in the middle, and mounted on an articulating joint to a head strap made of leather or fabric. The mirror is positioned over the physician's eye of choice, with the concave mirror surface facing outwards and the hole directly over the physician's eye.

In use, the patient sits and faces the physician. A bright lamp is positioned adjacent to the patient's head, pointing towards the physician's face and hence towards the head mirror. The lamp's light gets concentrated by the curvature of the mirror and reflected off it towards the area of examination, and along the line

of sight of the doctor, thus providing shadow-free illumination. When used properly, the head mirror thus provides excellent shadow-free illumination.

A French obstetrician named Levert, who was fascinated with the intricacies of the larynx and dabbled with mirrors, is credited with conceiving the idea for the head mirror back in 1743. Today's head mirror has withstood the test of time and is still routinely used by ophthalmologists and otolaryngologists, particularly for examination and procedures involving the oral cavity.

13.2.2 Otoscope

An *otoscope* is a hand-held optical instrument with a small light and a funnel-shaped attachment called an ear speculum, which is used to examine the ear canal and eardrum (tympanic membrane). It is also called *auriscope*. The otoscope is one of the medical instruments most frequently used by primary care physicians [15]. Health care providers use otoscopes to screen for illness during regular check-ups and also to investigate ear symptoms. Ear specialists—such as otolaryngologists and otologists—use otoscopes to diagnose infections of the middle and outer ear (otitis media and otitis externa).

The design of a modern otoscope is very simple [16]. It consists of a handle and a head (◘ Fig. 13.9). The handle is long and texture for easy gripping and contains batteries to power an integrated light. The head houses a magnifying lens on the eyepiece with a typical magnification of 8 diopters; a cone-shaped disposable plastic speculum at the distal end; and an integrated light source (either lamp bulb, LED, or fiber optic). The doctor inserts a disposable speculum into the otoscope, straightens the patient's ear canal by pulling on the ear, and inserts the otoscope to peer inside the ear canal. Some otoscope models (called pneumatic otoscopes) are provided with a manual bladder for pumping air through the speculum to test the mobility of the tympanic membrane.

The most commonly used otoscopes in emergency rooms and doctors' offices are monocular devices. They provide only a two-dimensional view of the ear canal. Another method of performing *otoscopy* (visualization of the ear) is use of a binocular microscope, in conjunction with a larger metal ear speculum, with the patient supine and the head tilted, which provides a much larger field of view and depth perception, thus affording a three-dimensional perception of the ear canal.

◘ **Fig. 13.9** Otoscope for visual inspection inside the ear canal
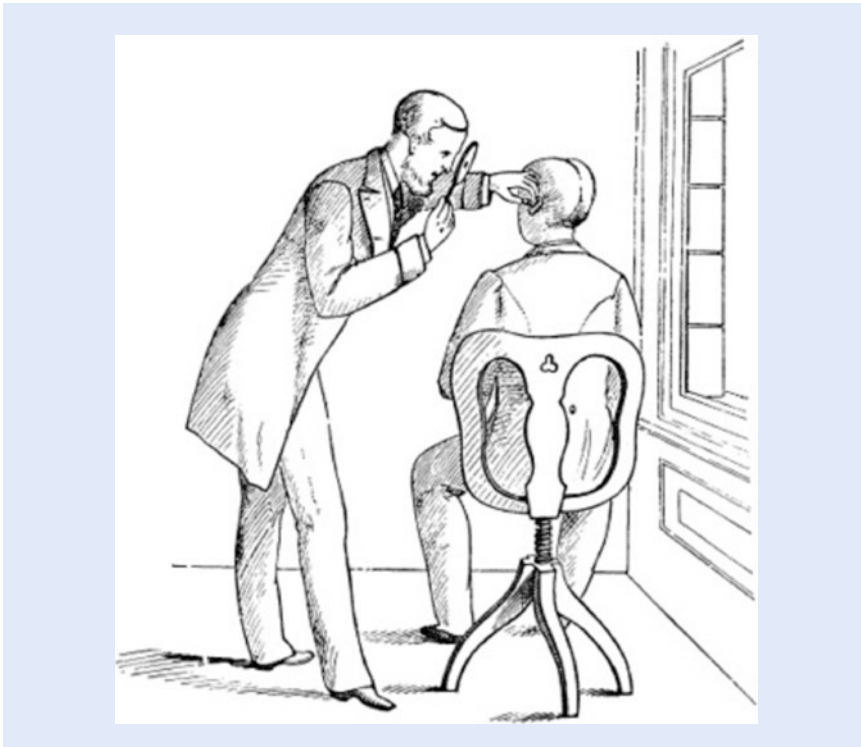
🔲 **Fig. 13.10** Ear examination in the nineteenth century

The microscope has up to 40× power magnification, which allows for more detailed viewing of the entire ear canal and eardrum.

The otoscope is a valuable tool beyond its primary role as an examination tool for detecting ear problems. It can also be used for transillumination, dermatologic inspection, examination of the eye, nose, and throat and as an overall handy light source.

13.2.2.1 History of the Otoscope

Early ear examinations were performed by direct observation of the ear canal during daylight. As a consequence, examinations were limited to times of the day and year when there was adequate bright daylight. Furthermore, a device was needed to gain more direct access to the ear canal and to keep it open and provide direct illumination inside. Hence, over the years, the use of a *speculum* (a conical shape device that can be safely inserted into the ear) was adopted. In 1363 Guy de Montpellier in France described the first aural and nasal specula [17]. However, some means or direct illumination was needed in order to perform more effective ear examinations. The next major requirement was for an adequate method of directing concentrated natural daylight into the depths of the ear canal, which was accomplished by using a perforated mirror mounted either on a handle or on the head, which shone light directly into the ear canal. This allowed the doctor to look down the center of the beam of light, thus eliminating shadow effects and parallax (difference in the apparent position of an object viewed along two different lines of sight) (🔲 Figs. 13.10 and 13.11).

Von Troltsch is generally credited with popularizing the use of a mirror in *otoscopy* after he showed it in 1855 at a meeting of the Union of German Physicians in Paris. He ultimately fastened the mirror to his forehead as is still currently practiced by some doctors. The size and focal length of the mirror was not standardized for some time. In an attempt to catch more light, used huge mirrors and only gradually was a diameter of 6–7 cm eventually adopted. A further
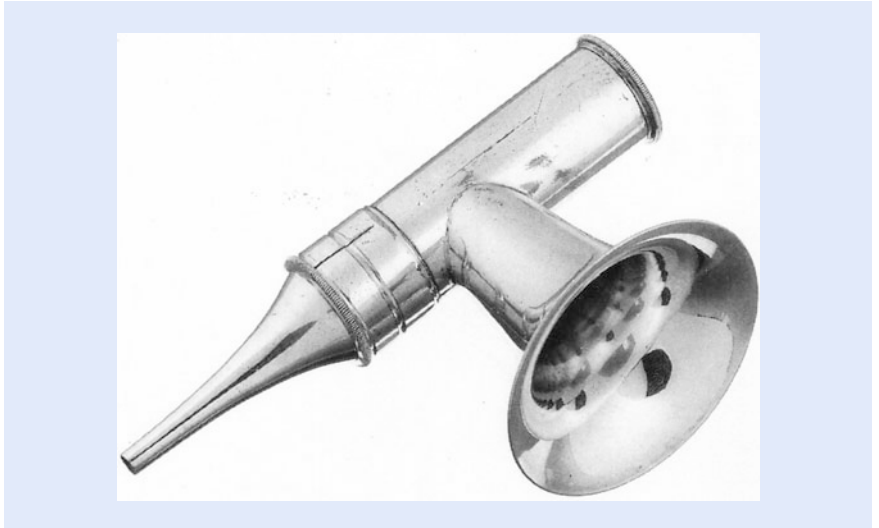
Fig. 13.11 Bunton's Auriscope (c. 1880). It can be observed the metal tip speculum, the rear objective lens for viewing, as well as the middle horn used to direct light from a candle or lamp

improvement to Von Troltsch's early auriscope is Brunton's device which was first described in an 1865 Lancet article. This auriscope combined mirror and speculum into a single instrument and worked on the principle of a periscope: light from a candle or lamp was concentrated by a funnel and then reflected by a plane mirror set at an angle of 45° into the ear canal. The mirror had a central perforation through which the doctor could view the ear. Brunton's auriscope was fitted with a magnifying lens for the observer and could also be sealed with plain glass at the illuminating end. These were the first otoscopes to be electrically illuminated.
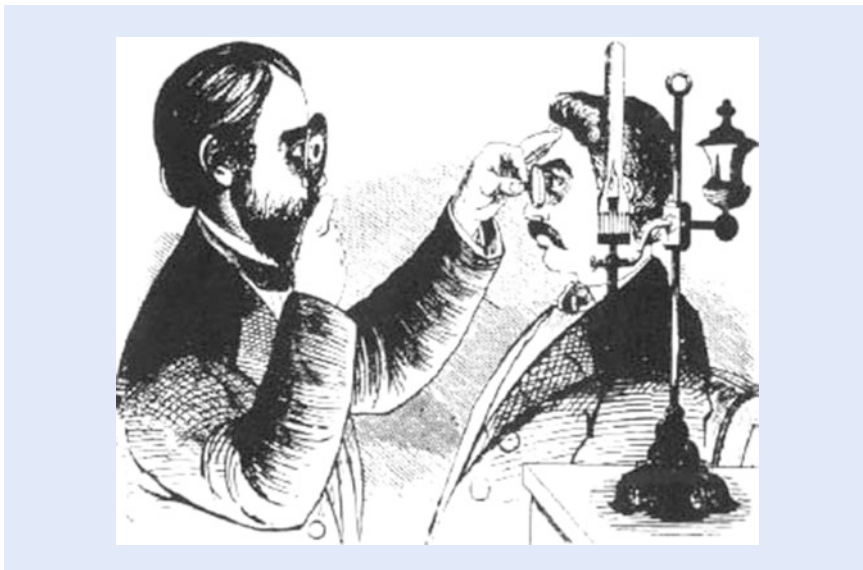
13.2.3 Ophthalmoscope

An *ophthalmoscope* is an optical instrument for examining the interior of the eyeball and its back structures (called the *fundus*) through the pupil by injecting a light beam into the eye and looking at its back-reflection. An ophthalmoscope is also referred to as a *funduscope*. The fundus consists of blood vessels, the optic nerve, and a lining of nerve cells (the retina) which detects images transmitted through the cornea, a clear lens-like layer covering of the eye. Ophthalmoscopes are used by doctors to exam the interior of eyes and help diagnose any possible conditions or detect any problems or diseases of the retina and vitreous humor. For instance, a doctor would look for changes in the color the fundus, the size, and shape of retinal blood vessels, or any abnormalities in the *macula lutea* (the portion of the retina that receives and analyzes light only from the very center of the visual field). Typically, special eyedrops are used to dilate the pupils and allow a wider field of view inside the eyeball.

A modern ophthalmoscope (Fig. 13.12) consists essentially of two systems: one for illumination and another for viewing. The illuminating system is comprised of light source (a halogen or tungsten bulb), a condenser lens system, a reflector (a prism, mirror, or metallic plate) to illuminate the interior of the eye with a central hole through which the eye is examined. The viewing system is made of a sight hole and a focusing system, usually a rotating wheel with lenses of different powers. The lenses are selected to allow clear visualization of the structures of the eye at any depth and compensate for the combined errors of refraction between patient and examiner.
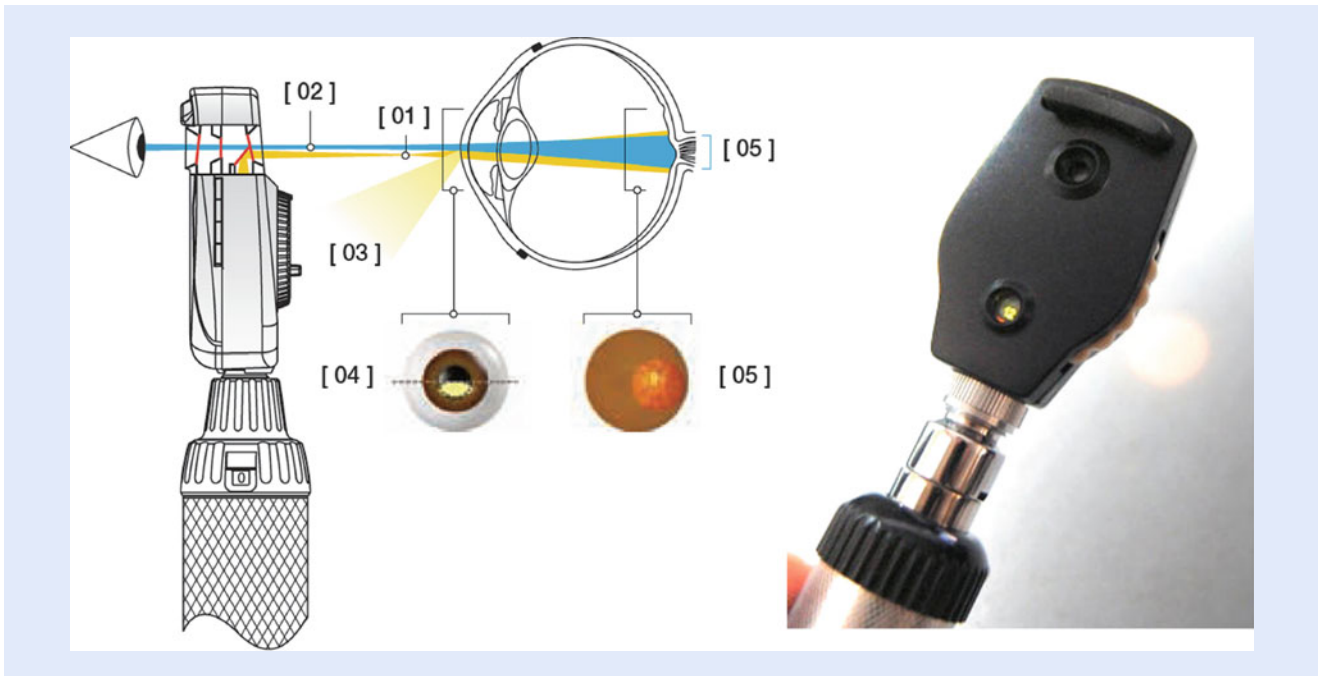
Fig. 13.12 Aspect of a modern ophthalmoscope. A light beam is projected into the eye (*1*). The medical examiner has a direct line of sight into the back of the eye (fundus) (*2*). Path of light reflected of the cornea and iris (*3*). Image observed at the pupil (*4*). Image observed at the back of the eye (*5*) (Images courtesy of Heine)

Fig. 13.13 Nineteenth century illustration of Helmholtz original ophthalmoscope

German physician Hermann von Helmholtz is credited with the invention of the ophthalmoscope back in 1851, which he based on an earlier version developed by Charles Babbage in 1847. Helmholtz original ophthalmoscope (see Fig. 13.13) was very basic (made or cardboard, glue, and microscope glass plates) but it allowed him to place the eye of the observer in the path of the rays of light entering and leaving the patient's eye, thus allowing the patient's retina to be seen. In 1915, Francis A. Welch and William Noah Allyn invented the world's
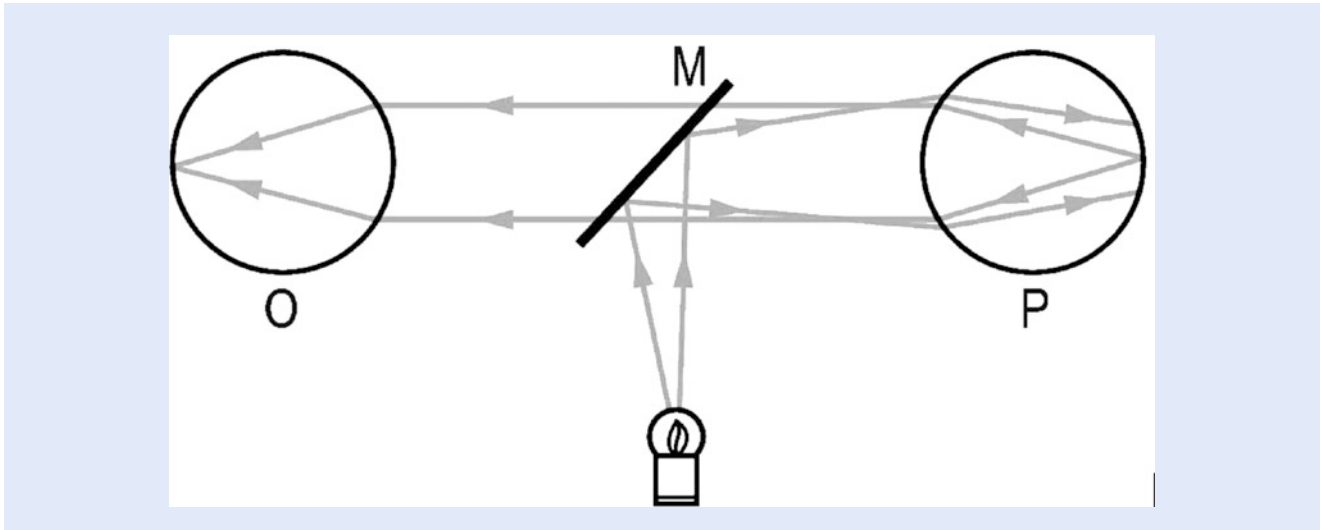
Fig. 13.14 Optical raytracing for a direct ophthalmoscope. Light from the illuminating source is reflected into the eye and then back-reflected by the fundus through a mirror (with either a hole through it or with partial reflectivity). O is the observer's eye, while P is the patient's eye; M, semi-silvered mirror. After [18]

first hand-held direct illuminating ophthalmoscope, and resulted in the formation of the Welch Allyn medical company—still in business today.

There are two types of ophthalmoscope: *direct* and *indirect*. A *direct ophthalmoscope* produces an upright (unreversed) image with 15× magnification. The direct ophthalmoscope is used to inspect the fundus of the eye, which is the back portion of the interior eyeball. Examination is best carried out in a darkened room. Macular degeneration and opacities of the lens can be seen through direct ophthalmoscopy. The instrument is held at close range to the patient's eye and the field of view is small (less than 10°) (■ Fig. 13.14). The magnification M of a direct ophthalmoscope is equal to:

$$M = F_e/4 \tag{13.2}$$

where F_e is the power of the eye.

An *indirect ophthalmoscope* produces an inverted (reversed) image with a 2–5× magnification and formed. A small hand-held lens and either a slit lamp microscope or a light attached to a headband are used to form an image of the back of the eye in space, at approximately arm's length from the doctor. An indirect ophthalmoscope provides a stronger light source, a specially designed objective lens, and opportunity for stereoscopic inspection of the interior of the eyeball. It is invaluable for diagnosis and treatment of retinal tears, holes, and detachments.

This aerial image is usually produced by a strong positive lens ranging in power from +13 diopter to +30 diopter that is held in front of the patient's eye. The practitioner views this aerial image through a sight hole with a focusing lens to compensate for *ametropia* and accommodation. This instrument provides a large field of view (25–40°) and allows easier examination of the periphery of the retina. This instrument has been supplanted by the binocular indirect ophthalmoscope (■ Fig. 13.15). The magnification of an indirect ophthalmoscope M is equal to:

$$M = F_e/F_c \tag{13.3}$$

where F_e and F_c are the powers of the eye and of the condensing lens, respectively.
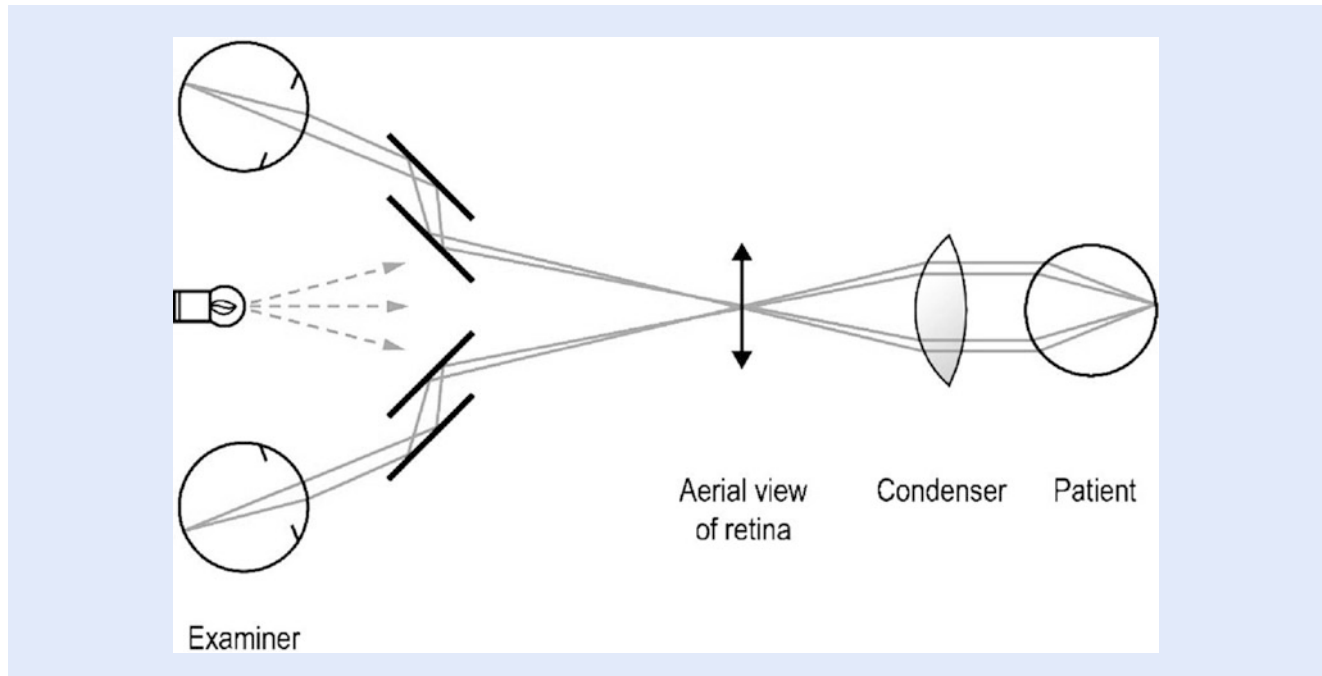
Fig. 13.15 Optical raytracing for a binocular indirect ophthalmoscope. The light source mounted on the doctor's head illuminates a hand-held condenser lens which forms an inverted stereoscopic image of the retina in free space (aerial image). After [18]
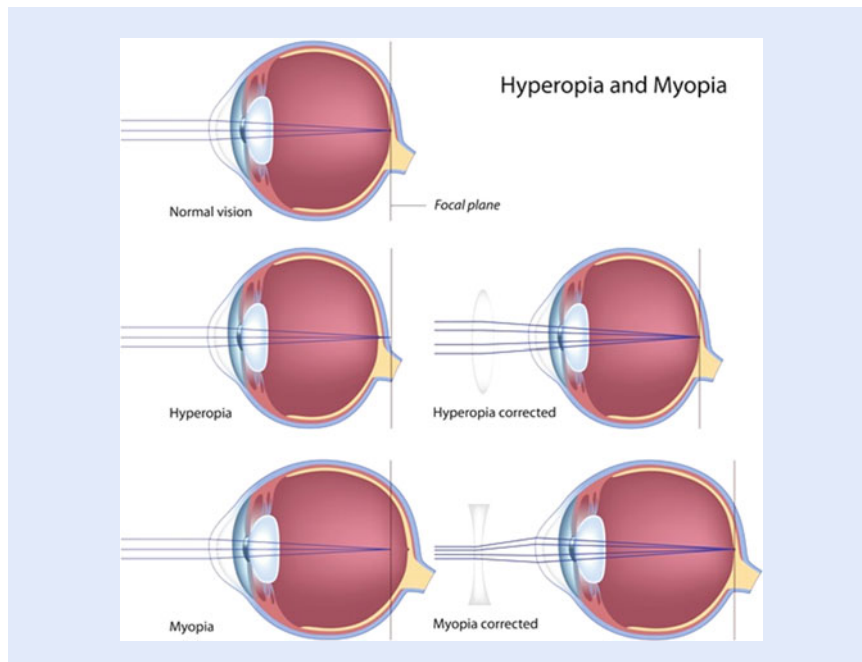
Fig. 13.16 Types of human vision and associated corrective optical lenses

13.2.4 Retinoscope

A *retinoscope* is an optical hand-held device used by optometrists to measure the optical refractive power of the eyes and whether corrective glasses might be needed and the associated prescription value. As shown in Fig. 13.16, a person can have normal vision (*emmetropia*), *myopia* (nearsightedness), *hyperopia*
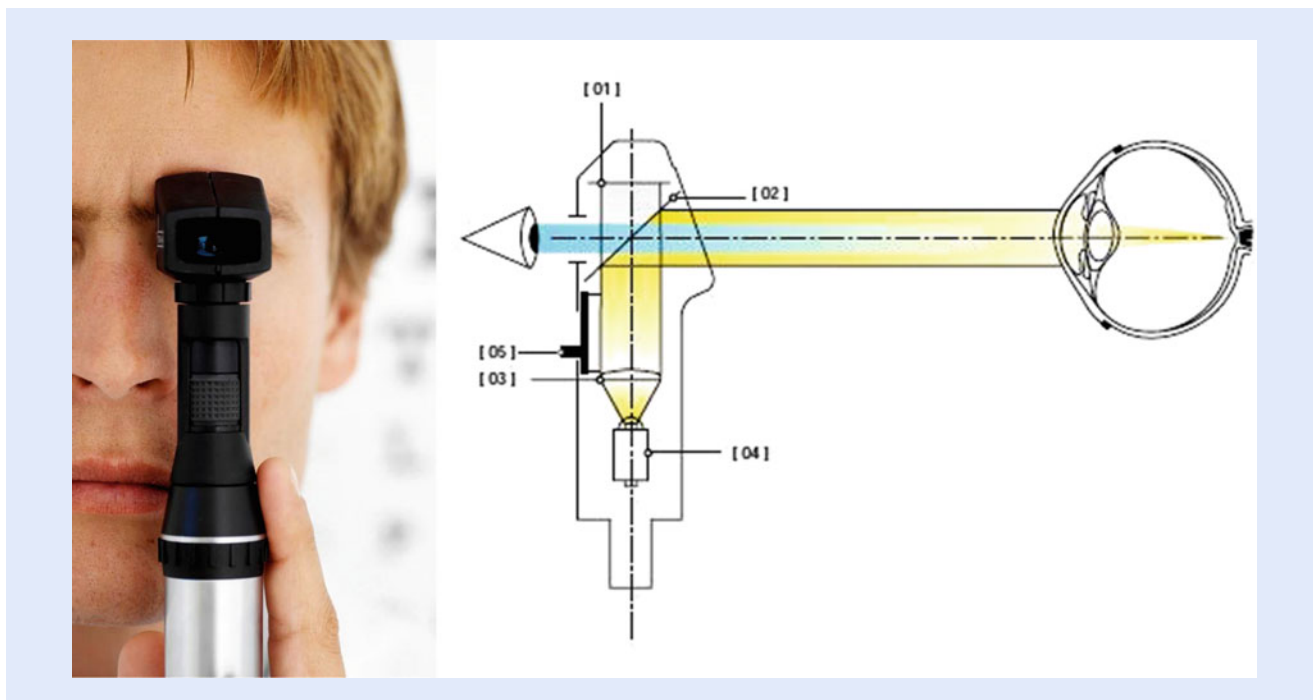
Fig. 13.17 A modern retinoscope. An integrated lamp or LED light source (*4*) shines light through a collimating lens (*3*) onto a partially reflective mirror (*2*), which directs the light to the eye. The back-reflected light from the fundus and the cornea is examined by the doctor through the eyepiece (*1*) and focus adjusted using the lens dial (*5*) (Image courtesy of Heine)

(farsightedness), or astigmatism. The retinoscope is used to illuminate the internal eye (while the patient is looking a far fixed object) and observe how the reflected light rays by the retina (called the *reflex*) align and move with respect to the light reflected directly off the pupil [19]. If the input light beam focuses in front of or behind the retina, there is a "refractive error" of the eye. A high degree of refractive power indicates that the light focus remains in front of the retina, in which case the eye displays myopia. Conversely, if the focal spot happens behind the retina, there is little refractive power and the eye has hyperopia. The error of refraction is then corrected by using a *phoropter*, which introduces a series of lenses of various optical strengths until the retinal reflex focuses at the right position on the retina.

The retinoscope consists of a light, a condensing lens, and a mirror (◘ Fig. 13.17). The mirror is either semi-transparent or has a hole through which the practitioner can view the patient's eye. During the procedure, the retinoscope shines a beam of light through the pupil. Then, the optometrist moves the light vertically and horizontally across the patient's eye and observes how the light reflects off the retina (see pictures in ◘ Fig. 13.18). If the light reflex in the patient's pupil moves "with" or "against" motion. If the reflex moves in same direction, then the correction requires plus power (myopia) and motion against direction of the retinoscope, means negative power correction (hyperopia).

To determine the corrective refractive lens power needed, lenses of increasing refractive power are placed in front of the eye and the change in the direction and pattern of the reflex is observed. The optometrist keeps changing the lenses until reaching a lens power that provides adequate focusing on the retina, which manifests as alignment of the reflex with the streak light image outside of the pupil.
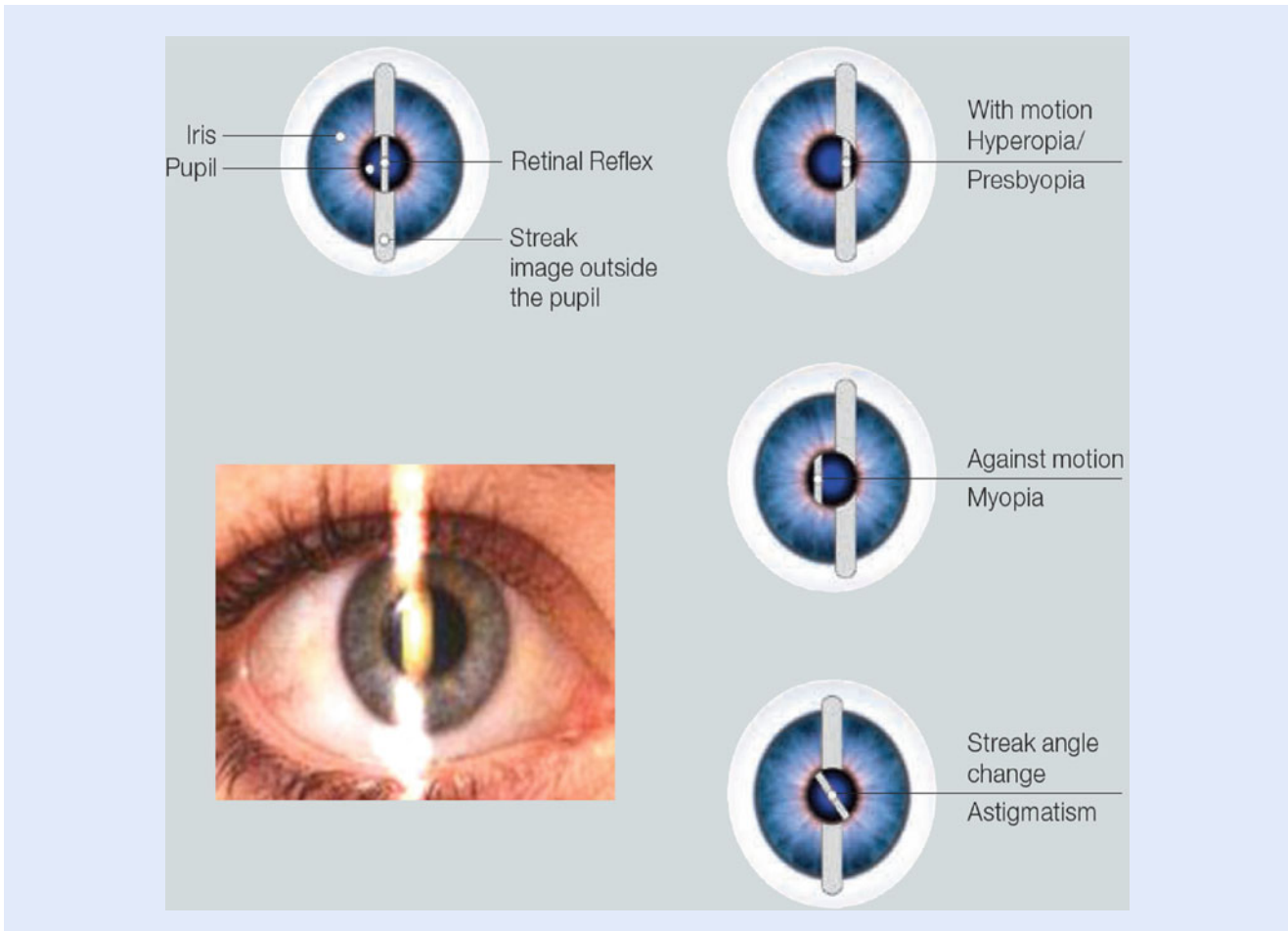
Fig. 13.18 Aspect of reflex images from the human eye seen by a doctor using a retinoscope. If the reflex moves in same direction, then myopia is detected; if reflex is noted on motion against direction of the retinoscope, hyperopia is present. If the reflex line is oblique instead of vertical, then astigmatism is present. An aligned reflex means correct vision (*Source*: Heine)

13.2.5 Phoropter

A *phoropter* is an ophthalmic binocular refracting testing device, also called a *refractor*. It is commonly used by ophthalmologists, optometrists, and eye care professionals during an eye examination to determine the corrective power needed for prescription glasses. It is commonly used in combination with a retinoscope.

Figure 13.19 shows a photograph of phoropter which consists in double sets (one for each eye) of rotating discs containing convex and concave spherical and cylindrical lenses, occluders, pinholes, colored filters, polarizers, prisms, and other optical elements. The patient sits in front of the device and the lenses within a phoropter refract light in order to focus images on the patient's retina at the right spot to compensate for each individual eye refractive errors. The optical power of these lenses is measured in 0.25 diopter increments. By changing these lenses, the examiner is able to determine the spherical and cylindrical power, and cylindrical axis necessary to correct a person's refractive error. These instruments were first devised in the early to mid-1910s.
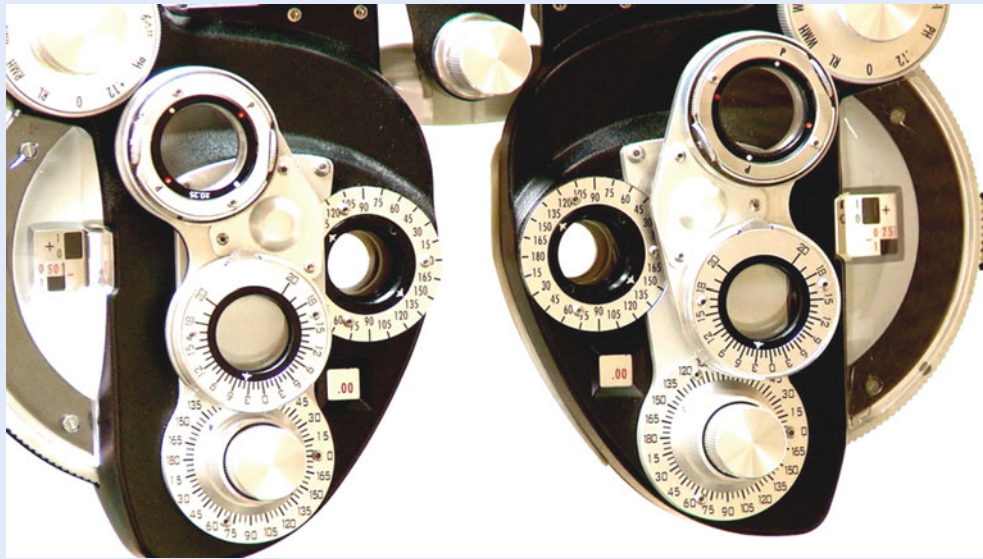
Fig. 13.19 A phoropter is commonly used by optometrists to determine the necessary corrective lens prescription

Fig. 13.20 A direct laryngoscope and its insertion into a human throat for examination

13.2.6 **Laryngoscope**

A *laryngoscope* is an optical instrument used for examining the interior of the larynx and structures around the throat. There are two types of laryngoscopes: *direct* and *indirect* laryngoscopes. A direct laryngoscope (Fig. 13.20) consists of a handle containing batteries, an integrated light source, and a set of interchangeable blades for easy reach and placement into a patient's throat. Besides being used for visualization of the glottis and vocal cords, a direct laryngoscope may also be used during surgical procedures to remove foreign objects in the throat, collect tissue samples (biopsy), remove polyps from the vocal cords, perform laser treatments and, very commonly, as a tool aid to facilitate tracheal intubation during general anesthesia or in cardiopulmonary resuscitation.

The blades in a laryngoscope help provide leverage to open wide the mouth and throat, as well as to keep the tongue in place and avoid a gag reflex. There are two basic styles of laryngoscope blades most commonly used: curved and straight. The Macintosh blade is the most widely used of the curved laryngoscope blades, while the Miller blade is the most popular style of straight blade. Blades come in different sizes, to accommodate different patients.

An *indirect* laryngoscope consists of a combination of a small mirror mounted at an angle on a long stem and a light source. The mirror is usually circular in form and made in various sizes, but is small enough to be placed in the throat behind the back of the tongue. The source of light is either a small bright lamp worn on the forehead of the observer, or a concave mirror, also worn on the forehead, for the purpose of concentrating light from some other source. Light is reflected to the back of the throat by the mirror and directed to illuminate up the interior of the larynx. The mirror also serves to reflect back to the doctor an image of the throat, to appreciate the structure of the glottis and vocal cords.

Some historians credit Benjamin Guy Babington (1794–1866), with the invention of the laryngoscope back in 1829 [20], who called his device the *glottiscope*. However, Manuel Garcia (1805–1906)—a Spanish tenor and singing maestro—experimented back in 1854 with a combination of throat mirror and light to observe the action of his own vocal cords and larynx when producing tones and sounds. His observations were published in the Royal Philosophical Magazine and Journal of Science in 1855 [21], and they constitute the first physiological records of the human voice as based upon observations in the living subject. For this, he is also recognized as the original inventor of the laryngoscope. �’ Figure 13.21 shows a photograph and illustration of his original laryngoscope device.

Mirror-based laryngoscopy for the investigation of laryngeal pathology was pioneered back in 1858 by Johann Czermak, a professor of physiology at the University of Budapest. Czermak applied an external light source and a head-mounted mirror to improve visualization. During this period of time, a laryngoscopic examination was made as depicted in �’ Fig. 13.22. The patient opens his mouth as widely as possible, protruding his tongue. The doctor, with a small napkin takes the protruded tongue between his thumb and forefinger and holds it in place, so as to enlarge opening of the mouth as much as possible. The laryngeal mirror is next inserted and dexterously positioned to the back of the mouth to direct the light from the external light source (mirror or lamp) into the back of the throat. An image of the lower throat is reflected back by the mirror for the doctor to view and assess the condition of the larynx.

All previous observations of the glottis and larynx had been performed under indirect vision (using mirrors) until 1895, when Alfred Kirstein (1863–1922) of Germany performed the first direct laryngoscopy in Berlin, using an *esophagoscope* he had modified for this purpose, calling device an *autoscope*, and the modern, direct laryngoscope was born [22].

13.3 Fiber Optic Medical Devices and Applications

The field of fiber optics has undergone a tremendous growth and advancement over the last 50 years. Initially conceived as a medium to carry light and images for medical endoscopic applications, optical fibers were later proposed in the mid-1960s as an adequate information-carrying medium for telecommunication applications. Ever since, optical fiber technology has been the subject of considerable research and development to the point that today light wave communication systems have become the preferred method to transmit vast amounts of data and information from one point to another.
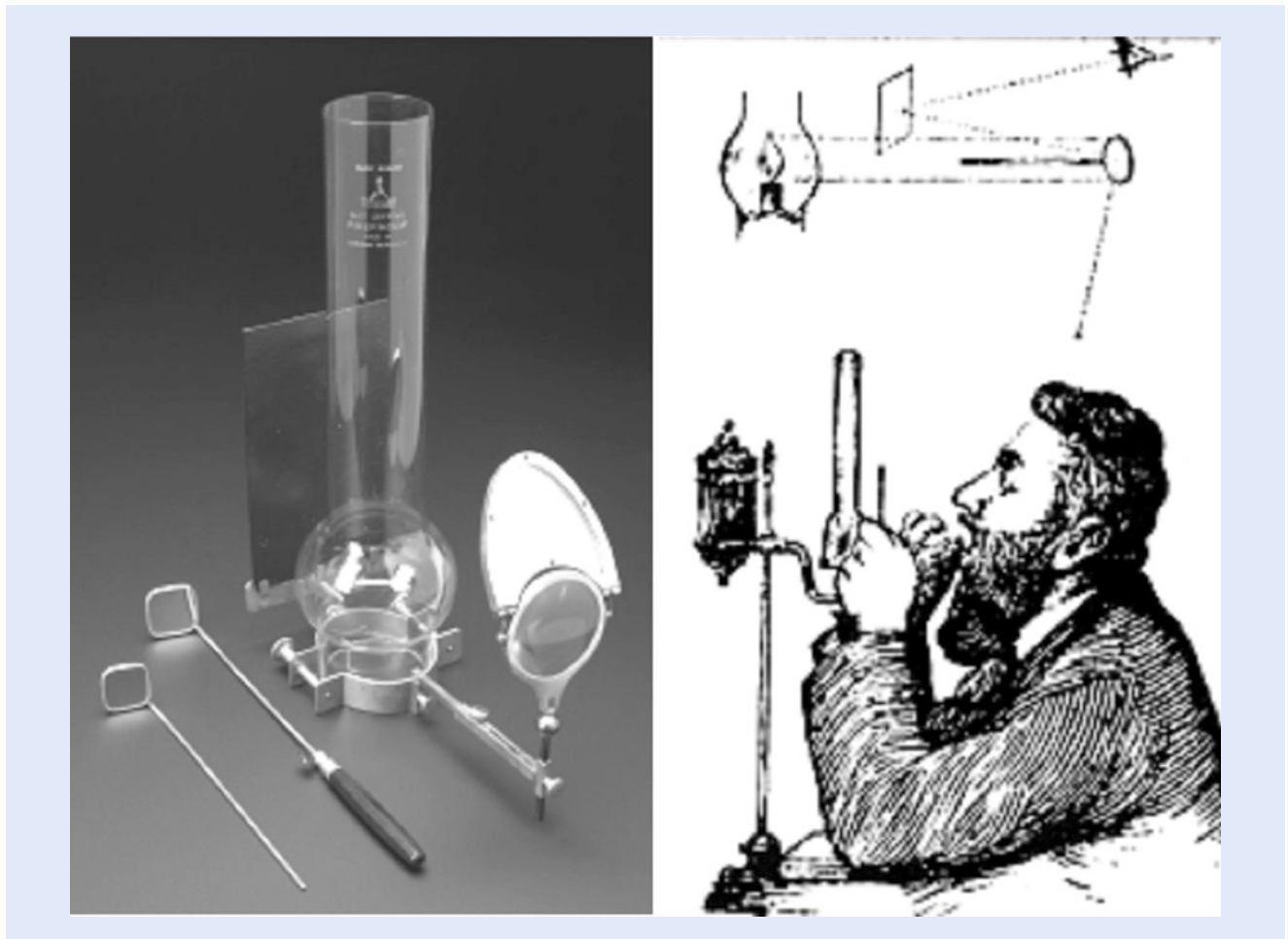
🔷 **Fig. 13.21** Original indirect laryngoscope developed by Manuel Garcia to view the movement of his won vocal chords (c. 1870)

Given their EM immunity, intrinsic safety, small size and weight, autoclave compatibility and capability to perform multi-point and multi-parameter sensing remotely, optical fibers and fiberoptic-based devices are seeing increased acceptance and new uses for a variety of biomedical applications—from diverse endoscopes, to laser-delivery systems, to disposable blood gas sensors, and to intra-aortic probes. This section illustrates—through several application and product examples—some of the benefits and uses of biomedical fiber sensors, and what makes them such an attractive, flexible, reliable, and unique technology.

13.3.1 Optical Fiber Fundamentals

At the heart of this technology is the optical fiber itself. A hair-thin cylindrical filament made of glass (although sometimes are also made of polymers) that is able to guide light through itself by confining it within regions having different optical indices of refraction. A typical fiber structure is depicted in 🔷 Fig. 13.23. The central portion—where most of the light travels—is called the core. Surrounding the core there is a region having a lower index of refraction, called the cladding. From a simple point of view, light trapped inside the core travels along the fiber by bouncing off the interfaces with the cladding, due to the effect of the total internal

● **Fig. 13.22** Nineteenth century illustration of a mirror-based laryngoscope examination of a patient's throat
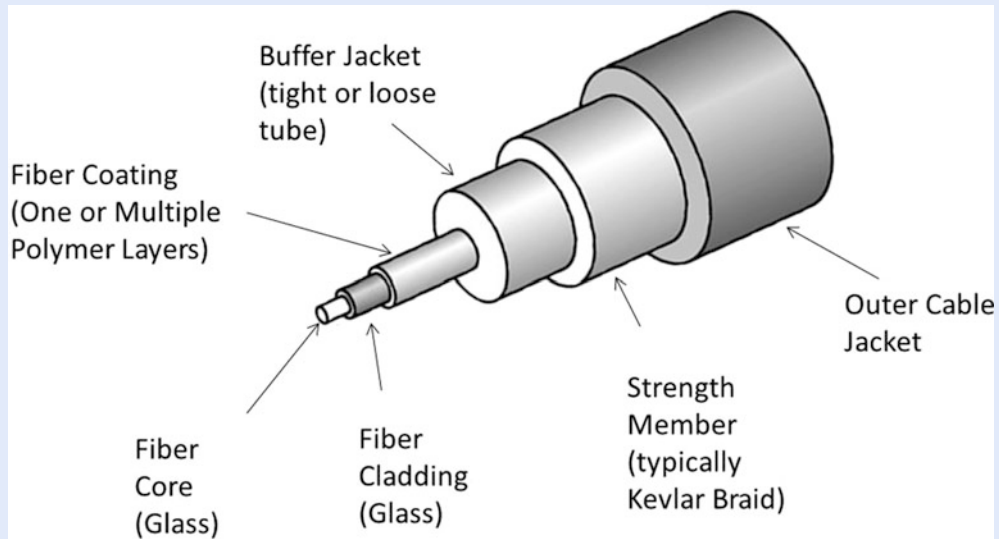
● **Fig. 13.23** Schematic of an optical fiber
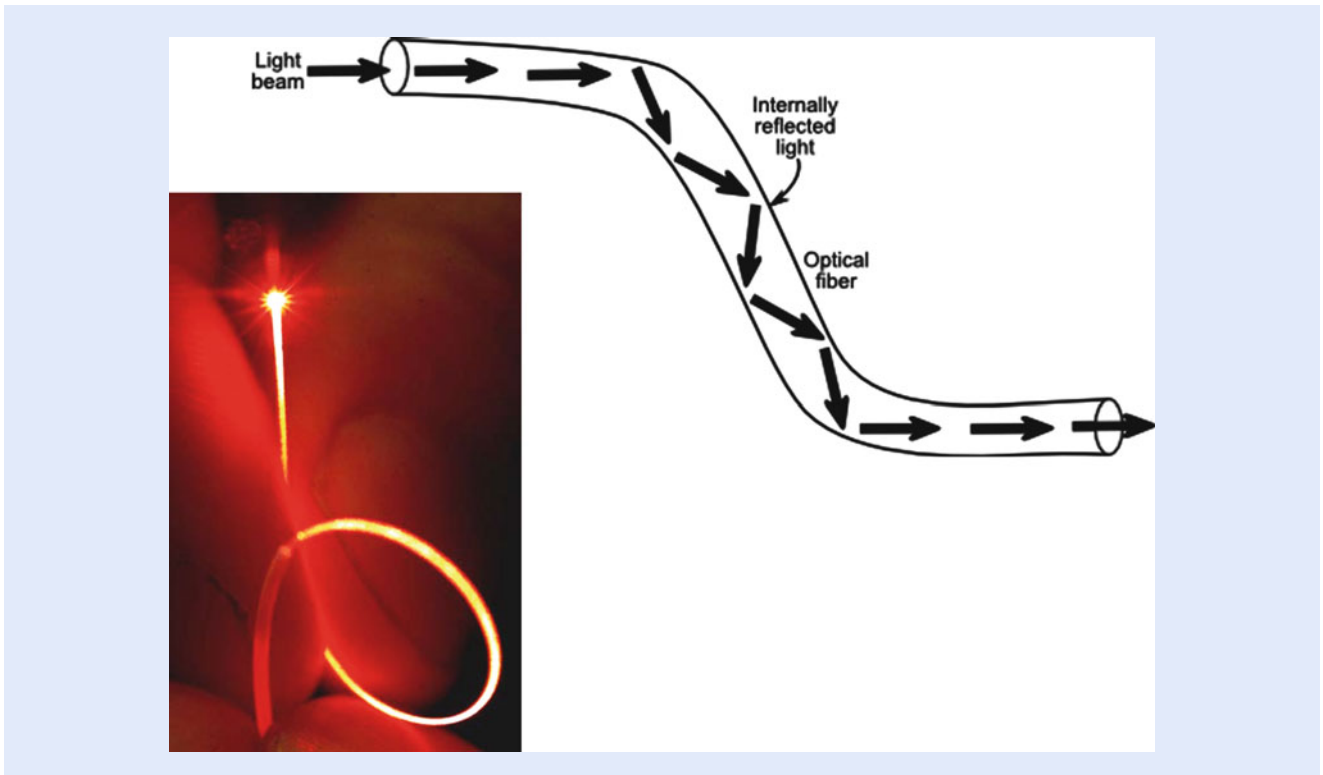
Fig. 13.24 An optical fiber is able to guide light through the principle of total internal reflection. This allows the transmission of light energy (and signals) through any patch or shape taken by the optical fiber

reflection occurring at these boundaries (■ Fig. 13.24). In reality though, the optical energy propagates along the fiber in the form of waveguide modes that satisfy Maxwell's equations as well as the boundary conditions and the external perturbations present at the fiber.

Refraction occurs when light passes from one homogeneous isotropic medium to another; the light ray will be bent at the interface between the two media. The mathematical expression (Eq. (13.4)) that describes the refraction phenomena is known as Snell's law,

$$n_0 \sin \phi_0 = n_1 \sin \phi_1 \tag{13.4}$$

where n_o is the index of refraction of the medium in which the light is initially travelling, n_1 is the index of refraction of the second medium, Φ_o is the angle between the incident ray and the normal to the interface, and Φ_1 is the angle between the refracted ray and the normal to the interface.

■ Figure 13.25a shows the case of light passing from a high-index medium to a lower-index medium. Even though refraction is occurring, a certain portion of the incident ray is reflected. If the incident ray hits the boundary at ever-increasing angles, a value of $\phi_0 = \phi_c$ will be reached, at which no refraction will occur. The angle ϕ_c is called the critical angle. The refracted ray of light propagates along the interface, not penetrating into the lower-index medium, as shown in part ■ Fig. 13.25b. At that point, $\sin \phi_c$ equals to unity. For angles ϕ_0 greater than ϕ_c, the ray is entirely reflected at the interface, and no refraction takes place (see ■ Fig. 13.25c). This phenomenon is known as *total internal reflection*.

In ■ Fig. 13.26, a ray of light incident upon the end of the optical fiber at an angle θ will be refracted as it passes into the core. If the ray travels through the high-index medium at an angle greater than Φ_c it will reflect off of the cylinder
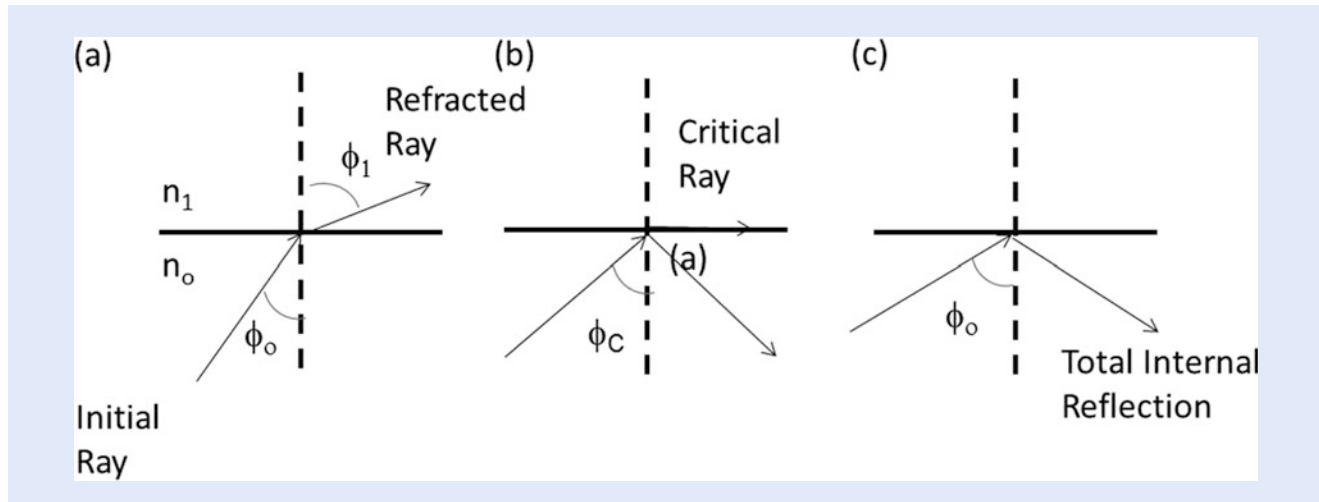
◘ **Fig. 13.25** Light reflection and refraction between two media with different indices of refraction
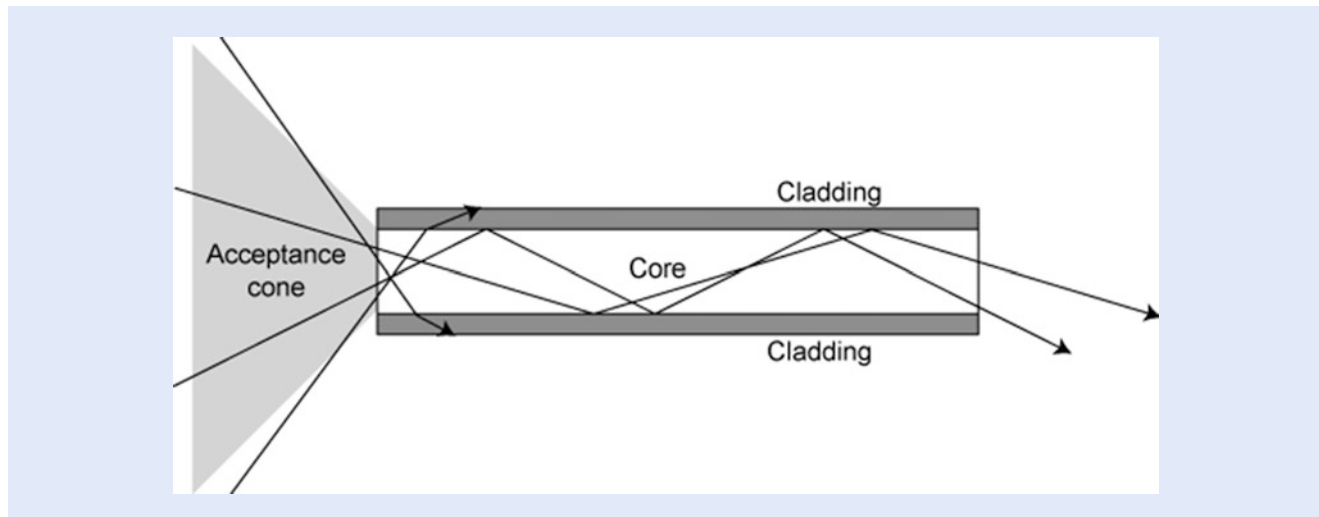

◘ **Fig. 13.26** Light ray propagation along the core of an optical fiber. At the entrance to the fiber, a conical region is defined by the so-called acceptance angle, which is the region in space where light can be effectively collected and coupled into the fiber for guiding

wall, will have multiple reflections, and will emerge at the other end of the optical fiber. For a circular fiber, considering only meridional rays, the entrance and exit angles are equal. Considering Snell's law for the optical fiber, core index n_0, cladding index n_1, and the surrounding media index n,

$$
\begin{aligned}
n \sin \theta &= n_0 \sin \theta_0 \\
&= n_0 \sin \left(\frac{\pi}{2} - \phi_c \right) \\
&= n_0 \left[1 - (n_1 - n_0)^2 \right]^{1/2} \\
&= \left(n_0^2 - n_1^2 \right)^{1/2} = \text{Numerical Aperture.}
\end{aligned}
\tag{13.5}
$$

The term $n \sin\theta$ is defined as the *numerical aperture* or NA for short. The NA is determined by the difference between the refractive index of the core and that of

13
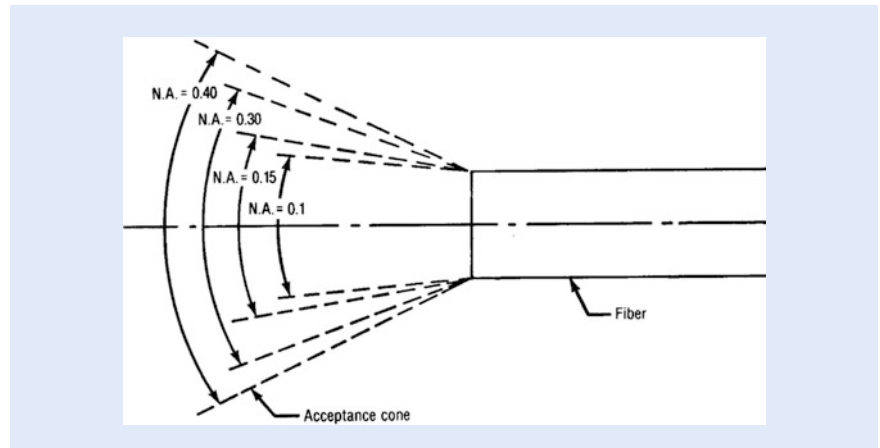
🔲 **Fig. 13.27** The greater the NA of an optical fiber, the bigger the acceptance cone, and broader the angle of capture of light by the fiber

the cladding. It is a measure of the light-acceptance capability of the optical fiber. As the NA increases, so does the ability of the fiber to couple light into the fiber, as shown in 🔲 Fig. 13.27. The larger NA allows the fiber to couple in light from more severe grazing angles. Coupling efficiency also increases as the fiber diameter increases, since the large fiber can capture more light. Therefore, the maximum light-collection efficiency occurs for large-diameter-core fibers and large-NA fibers.

13.3.2 Coherent and Incoherent Optical Fiber Bundles

In medicine, optical fibers have been considered for illuminating and imaging applications since the 1920s. Typically, a single glass optical fiber has a diameter ranging from 1 mm down to ~8 μm. However, a single optical fiber cannot transmit an image—only a bright light spot would be observed at its end. Hence, in order to carry a reasonable amount of light for illumination purposes, or to transmit and image, hundreds to thousands of optical fibers need to be assembled into bundles. Bundles of multiple single optical fibers of small diameter solid glass rods can thus be used to guide light or transmit images around bends and curved trajectories.

Glass optical fiber bundles are of two types: *incoherent* and *coherent*. An *incoherent* bundle consists of a collection of fibers randomly distributed in the bundle and is typically intended for illumination purposes only. In contrast, a *coherent* optical fiber bundle has an ordered array of fibers in which the relative position of each individual fiber at its input and output with respect to the bundle is maintained. That is to say, the position of individual fibers is at same locations over the cross section of both bundle ends as depicted in 🔲 Fig. 13.28. In between the ends, the fibers need not have a fixed orientation and can move flexibly. Coherent bundles are used for conveying an image from one end to the other by the effect created by the grouping of the individual light conducted by each fiber which is perceived in the eye of the observer as a full image. To achieve better image quality and resolution, a large number of small diameter fibers are need for a given bundle diameter. Typically, fibers used in bundles have diameters on the order of 8–12 μm and their count can range from about 2000 up to 40,000 [23]. In the case of imaging bundles, larger diameter fibers are used of 30–50 μm in diameter.
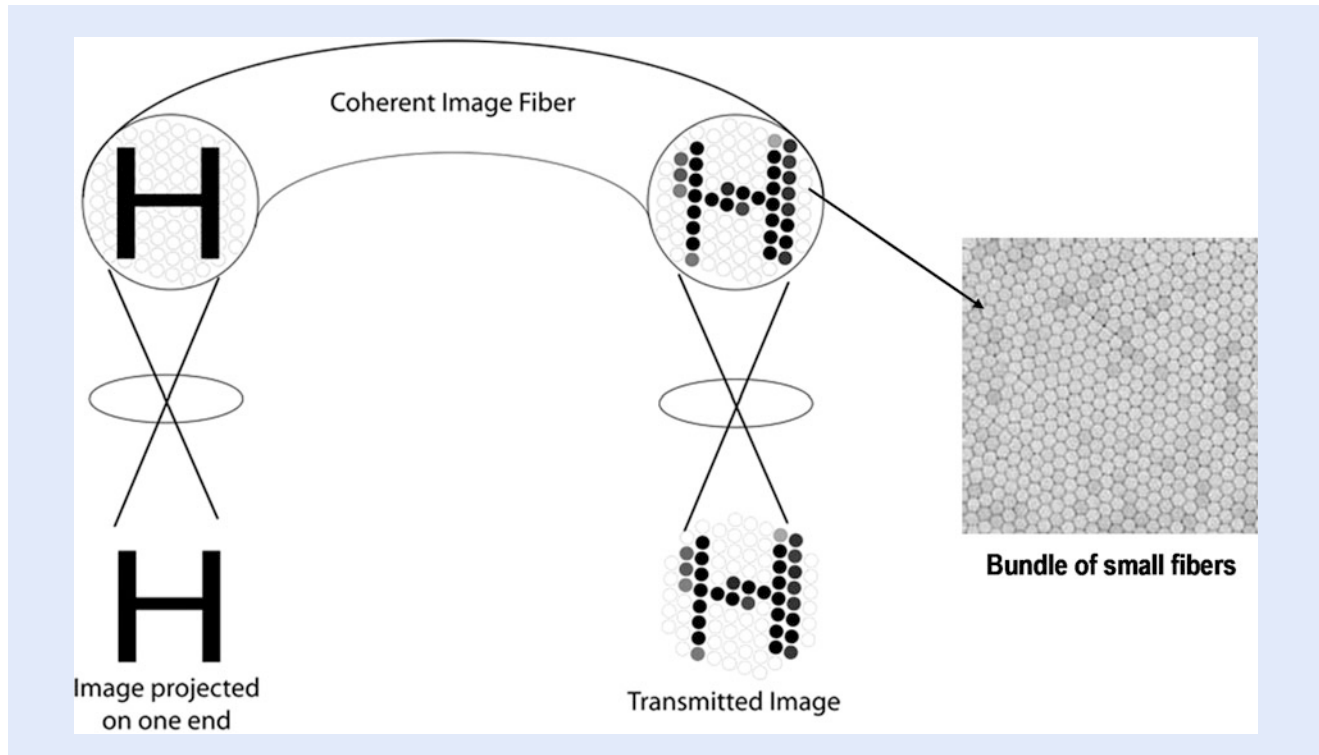
■ **Fig. 13.28** A coherent optical fiber bundle. Images are accurately transmitted by preserving the relative position of the fibers at each end of the bundle. The bundle consists of a multitude of individual glass fibers of a small diameter (~12 μm) that create a lattice effect

Fabrication of illumination (non-coherent) and imaging (coherent) bundles is based on the same processes of drawing optical fibers or glass rods through heating furnaces and doing repeated draws of multi-stack sets, to achieve arrays with the desired quantity of fibers of the appropriate diameter. There are three common to fabrication methods for coherent bundles: fused image bundles, wound image bundles, and leached image bundles. ■ Figure 13.29 illustrates the three steps needed to fabricate fused as well as leached image fiber bundles.

An individual fiber (or rod) is made by starting with a so-called perform made by the tube-in-rod technique where a single glass rod (which will become the fiber's core) is inserted into a tube made of glass with a lower refractive index (cladding). In the case of a leached bundle, an additional glass jacket made of a leachable glass is used. This glass perform is placed in an electric heating furnace that runs at a temperature close to the softening point of the glass. The heat causes the solid glass road to soften. Once soft, the glass is pulled down into a thin filament by a pulling mechanism. The final diameter of the filament is controlled by the ratio of the speeds between the advancing preform and the drawn fiber. Typically, the initial drawn fiber is more of a solid rod with a 2 mm diameter. In the next drawing stage, a multitude of mono fibers are stacked together and drawn in the furnace to produce a multi-fiber rod. The drawn filament from a multi-fiber preform consists of several 100 monofilament fibers. In the third stage, several multi-fiber rods are stacked together to perform the so-called multi-multi drawing process. The multi-multi stack assembly is fed through the furnace and drawn into a filament of rod of the desired diameter. Such filament will be composed of thousands of individual glass fibers. As shown in ■ Fig. 13.30, imaging multi-fiber arrays of square, circular, or hexagonal shape and in different sizes can be fabricated with this process.
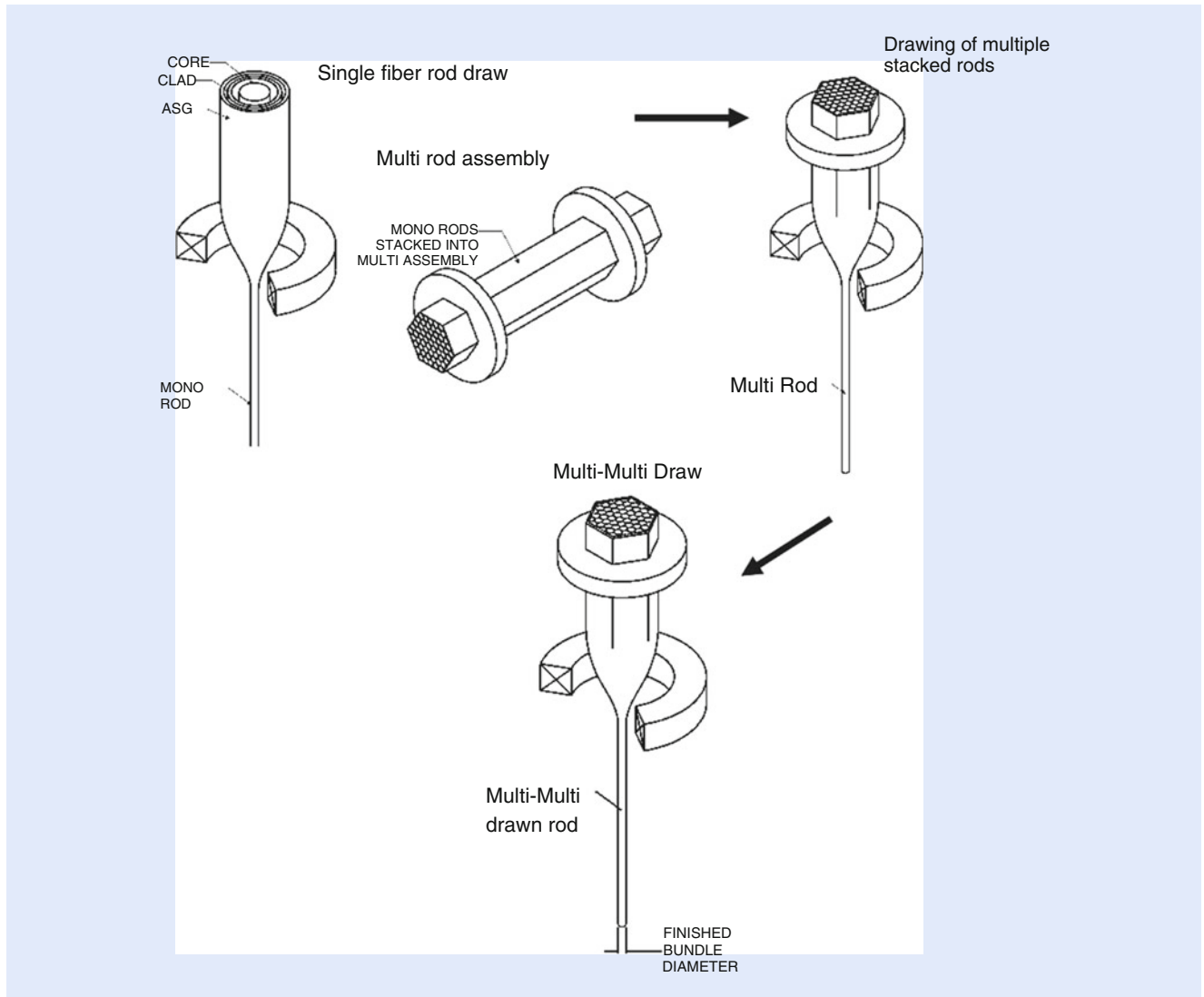
13

● **Fig. 13.29** Fabrication process to make fiber bundles

In the particular case of leached fiber bundles, each bundle end is properly secured and the entire bundle is soaked in an acid solution which will dissolve the leachable glass, allowing the fibers to move freely between the bundle ends.

Wound imaging bundles are made by winding a multi-fiber array as a single layer on a drum, and then stacking the desired number of layers manually in a laminating operation.

13.3.3 Illuminating Guides

Fiber optic illuminating guides are non-coherent and are used primarily to guide light to a desired point to provide illumination and enhance visual clarity. Imaging bundles are typically made of 30–50 μm diameter fibers, with NA values around 0.6. Most commonly, illuminating bundles are used as part of fiberscopes, endoscopes, and personal lights for surgeons. As seen in ● Fig. 13.31, when surgeons are operating on a patient, they need cool, bright light to help then see better tissues and organs—the closer the direct illumination to the operating field,
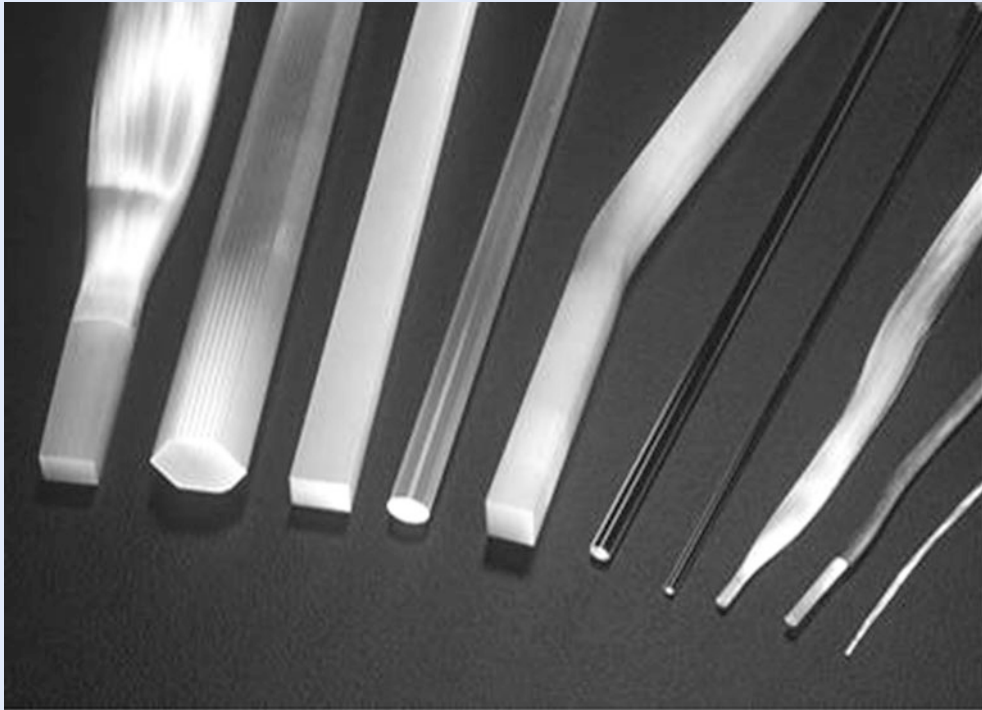
Fig. 13.30 Photograph of different styles and shapes of drawn coherent optical fiber bundles

the better. Rigid, light-guiding rods are also made (from single solid glass rods or from multi-core rods) for applications in dentistry and light therapy (see Fig. 13.32).

13.3.4 Fiberscopes and Endoscopes

An endoscope is an optical instrument used for direct visual inspection of hollow organs or body cavities. Typically, an endoscope is generally introduced through a natural opening in the body (Fig. 13.33), but it may also be inserted through an incision. Instruments for viewing specific areas of the body include the broncho-scope, colonoscope, cystoscope, gastroscope, laparoscope, proctoscope, and several others. Although the design may vary according to the specific use, all endoscopes have similar construction and elements: an objective lens (distal end), illuminating fiber bundle, imaging coherent fiber bundle, fixed or articulating handle, and an eyepiece (proximal end). Accessories that might be used for diagnostic or thera-peutic purposes include irrigation channels, suction tips, tubes, and suction pump; forceps for removal of biopsy tissue or a foreign body; biopsy brushes; an electrode tip for cauterization; as well as a video camera, video monitors, and image recorder. Many modern endoscopes have also articulating ends, that are remotely controlled by the doctor using knobs on the handle that adjust pull wires inside the body of the endoscope. Figure 13.34 shows a modern, flexible, and fiber-optic endoscope.

Endoscopes can be rigid or flexible as depicted in Fig. 13.35. Modern endoscopes (both flexible and rigid) make use of fiber optic imaging bundles to achieve image transmission. However, earlier models relied on miniature flat or rod lenses to guide images from the objective end to the eyepiece as shown in Fig. 13.36. Hippocrates II (460–377 BC) reported using catheters and primitive

**Fig. 13.31** Fiber optic illuminators used in the operating room by surgeons

Fig. 13.32 Solid fiber optic illuminating rods

forms of visualization tubes over two millennia ago. In the nineteenth century, endoscopy was very rudimentary and relied on the insertion of long, rigid metal tubes into body cavities. In 1910 Victor Elner used a gastroscope to view the stomach, while in 1912 the first semi-flexible gastroscope was developed. Then, Heinrich Lamm was the first person to transmit images through a bundle of optical
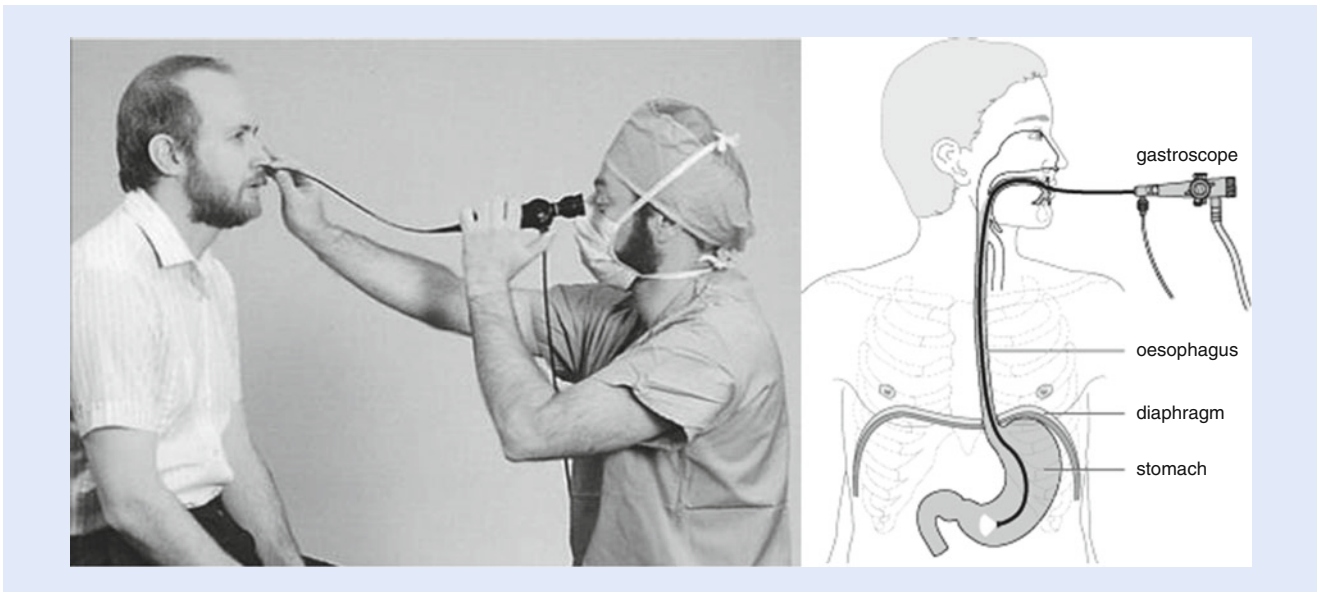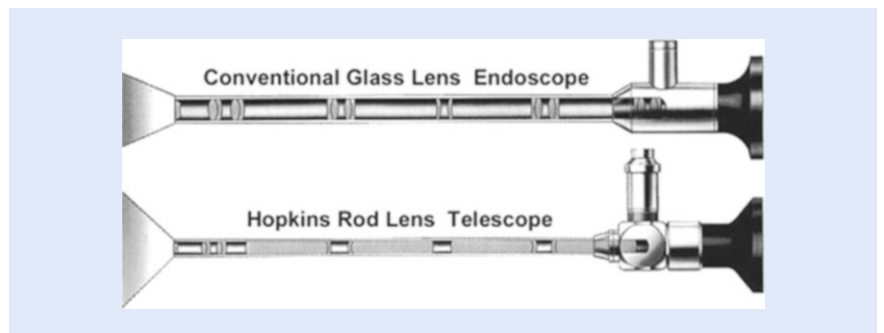
◘ **Fig. 13.33** Endoscopes are commonly used by doctors to inspect patients' internal organs through body cavities, such as the nose or throat
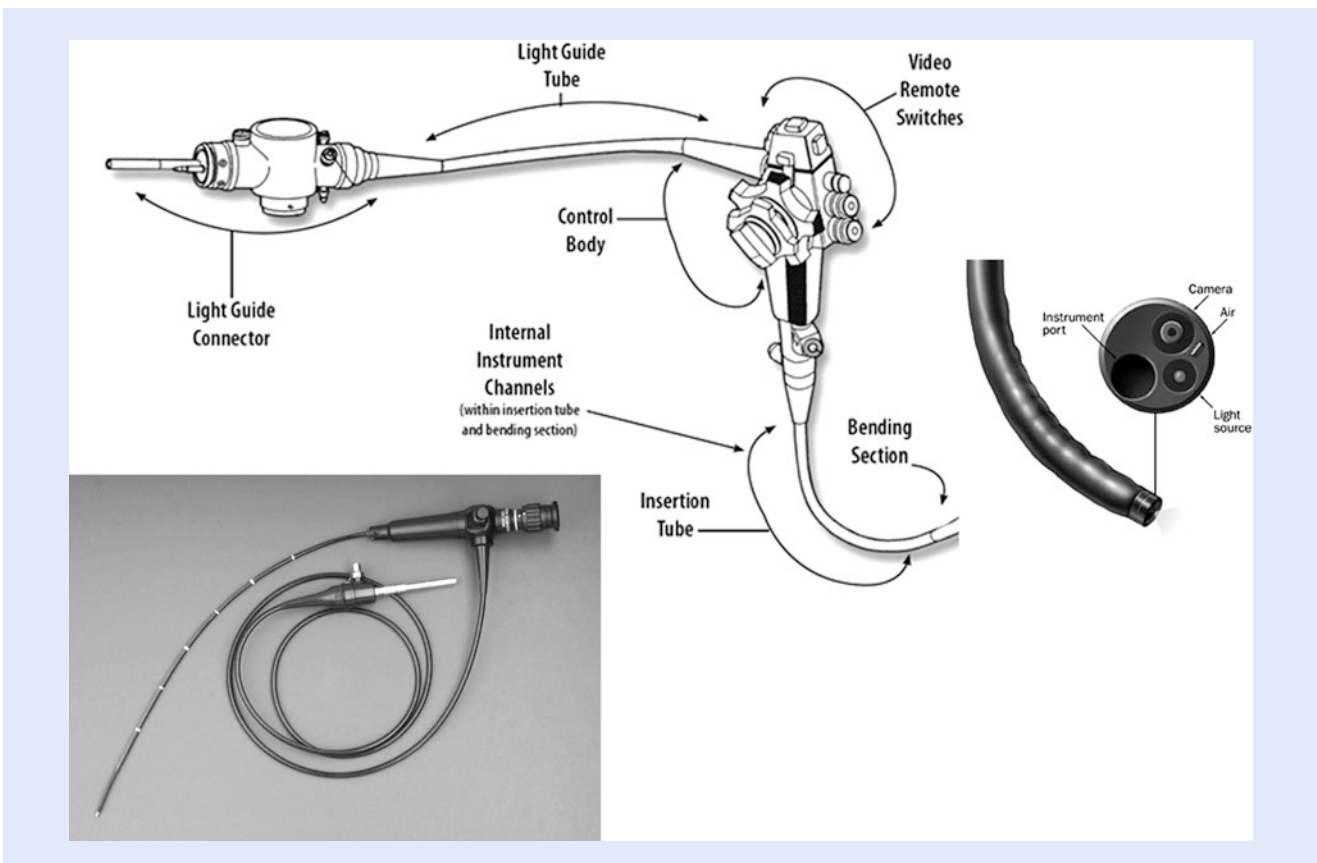
◘ **Fig. 13.34** Aspect of a modern, flexible fiberscope fitted with articulating knobs, camera lens, and instrument port on the distal end

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Fig. 13.35 Examples of a rigid (*upper*) and flexible (*lower*) endoscopes


Fig. 13.36 Early designs of rigid endoscopes using rod or glass lenses for image relaying

fibers in 1930. In 1957, clad optical fibers were first proposed and developed by Lawrence Curtis as a graduate student at the University of Michigan, under the supervision of Dr. Basil Hirschowitz, who in 1957 demonstrated the first fiber optic endoscope [24]. From then on, the devices became known as *fiberscopes*. The fiberoptic endoscope has great flexibility, reaching previously inaccessible areas and has become the norm in medicine.

13.3.5 Fused Fiber Faceplates and Tapers for Digital X-rays

Another type of coherent imaging conduit is the fiber optic fused faceplate (FOFP). FOFPs are made as pre-arranged blocks of multiple pre-drawn multi-fiber glass rods (known as boules), which are then fused together under elevated heat and pressure to form a solid piece (■ Fig. 13.37). Typical individual fiber element sizes range from as small as 4 to 25 μm or larger. Thin plates are then sliced from the fused boule, ground and polished to the desired thickness—ranging from ~100 mm down to a practical limit of 50 μm. Typical shapes are round or rectangular. Depending on the intended application, the FOFP end faces can be coated with a specific spectral filtering, phosphorescent, or anti-reflective coating.
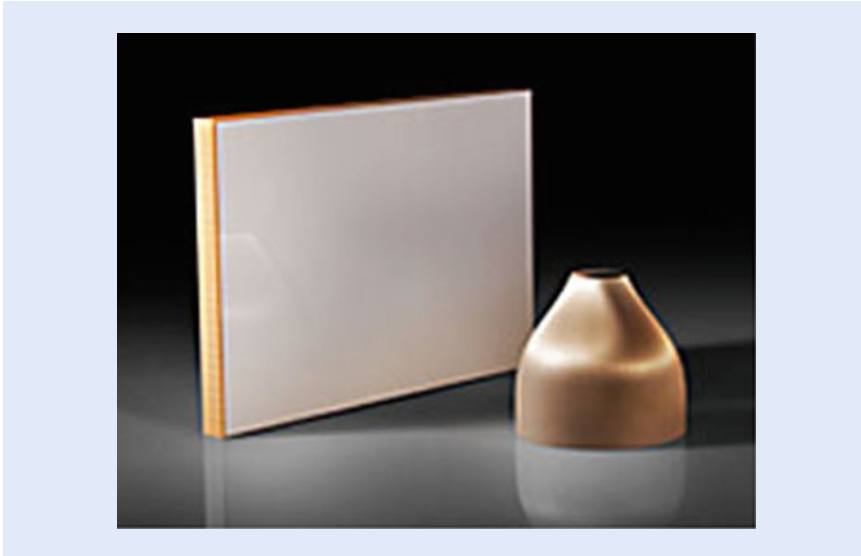
🔹 **Fig. 13.37** A fused fiberoptic faceplate (*back*) and taper (*front*). Photo courtesy of Schott glass

Optically, an FOFP behaves as zero-thickness optical window transferring an image, fiber by fiber, from one face of the plate to the other. Image magnification or reduction can be achieved by tapering the cross section of the bulk plate during the manufacturing process. In this case, the boule is drawn down and a neck region is formed with an hour-glass shape piece. The piece is cut into two pieces, machined and the ends polished resulting in a fused fiber optic taper.

Faceplates and tapers also function as dielectric barrier and mechanical interface and are optically used as a two-dimensional image conduit for energy conversion, field-flattening, distortion correction, and contrast enhancement. They are typically used for imaging applications bonded to cathode ray tubes (CRT) and LCD displays, image intensifiers, charged coupled device (CCD) or complementary metal-oxide semiconductor (CMOS) detectors, image plane transfer devices, X-ray digital detectors, among others.

In the medical area, fiber optic tapers and faceplates have found widespread use for both dental and medical digital radiography (such as mammography, fluoroscopy, intra-oral, panoramic, or cephalometric) where instead of using conventional film to obtain the X-ray images, an electronic photosensitive device such as a CCD or CMOS detector chip is used to convert the X-ray energy into electronic pixel signals via the use of an intermediate faceplate. Digital radiography offers high-resolution images while greatly reducing patient and sensor exposure to harmful X-rays by using low-dose X-ray sources. In addition, digital X-ray imaging speeds the availability of images for diagnostic, while also making the viewing, sharing, transmitting, and storing of X-ray patient data so much easy and compatible with modern electronic record systems. Furthermore, faceplates also provide a critical X-ray absorbing barrier between the X-ray emitter and the semiconductor detector device, prolonging their service life and reducing background noise.

As shown in 🔹 Fig. 13.38, when an X-ray source emits radiation energy (that would pass through the patient) the transmitted energy impinges on a scintillator plate which converts the radiation rays into visible photons. The scintillating coating—e.g., cesium iodide (CSI) or gadolinium oxysulfide (Gadox) doped with Tl or Eu—is deposited directly on the large end of a fused fiber-optic taper. The light is then transferred and reduced through the taper and coupled to a digital CCD chip where a black and white image is formed which can then be viewed on a computer screen or monitor and readily archived as an electronic image file.

Fig. 13.38 In digital X-rays, a fused fiberoptic taper is used to guide the light image from a scintillator an electronic CDD detector array for processing and visualization

13.4   Conclusions

As discussed in this section, optics is a useful, practical, versatile, and powerful technology that, throughout history, has helped human kind perform visual examination, diagnostics, and therapeutics on both the sick and healthy. Optics technology and optical components are at the core in a variety of modern-day optical devices and instruments such as endoscopes, patient monitoring probes, and sensors, as well as in advanced robotic assisted surgery systems.

The harnessing power of light, and its interaction with living matter, is extremely useful and beneficial for a variety of medical purposes and treatments ranging from laser procedures for tattoo removal, to eye surgery to vessel and tissue ablation and coagulation, up to modern photodynamic therapy treatments. We have seen how the field of optics is in itself a subset of a more complex and interdisciplinary area of research known as Biophotonics.

New advancements in optics and photonics are driving the development of a new generation of imaging tools—such as optical coherence and photo-acoustic tomography—that can readily provide two and three-dimensional images of diverse human body tissues and organs.

Optics has, and will continue to be, an *enabling technology* for the advancement of medicine promoting unimaginable new devices, techniques, and applications to happen in the not too distant future.

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