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REVIEW ARTICLE

Physics for Surgeons-Part 5: Optics for Surgeons

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ABSTRACT

Optical technologies are integral to modern surgical practice, supporting visualization, diagnosis, guidance, and therapy across a wide range of clinical applications. The performance and limitations of optical surgical instruments are fundamentally governed by the physical principles of light generation, propagation, and interaction with biological tissue. A clear understanding of these principles is essential for correct interpretation of optical information and safe, effective use of optical technologies in surgery. This review presents the fundamental principles of optical physics underlying diagnostic and therapeutic optical instruments used in surgical practice. Emphasis is placed on wavelength-dependent light–tissue interaction, optical properties of biological tissue, illumination sources, optical components, and imaging mechanisms relevant to surgery. Established optical technologies, including surgical microscopes, endoscopes, pulse oximeters, lasers, fluorescence-based systems, and optical fibers, are discussed alongside advanced imaging and therapeutic techniques such as optical coherence tomography, confocal microscopy, spectroscopy, and photodynamic therapy. Emerging optical approaches that provide molecular and functional information are also reviewed. By integrating optical physics with clinical context, this article aims to support surgeons and clinicians in understanding and interpreting optical technologies used in modern surgical care, while highlighting recent developments and future directions in biomedical optics relevant to surgery.

Keywords: Biomedical optics; Light–tissue interaction; Optical technologies in surgery.

1 Introduction

Optical technologies play a central role in contemporary surgical practice. Surgeons routinely rely on light-based instruments for visualization, diagnosis, guidance, and therapy across a wide range of clinical specialties. Optical microscopes, endoscopes, surgical loupes, fiber-optic imaging systems, lasers, and fluorescence-based devices are integral to both open and minimally invasive procedures. The performance, limitations, and safety of these instruments are fundamentally governed by the physical principles of light generation, propagation, and interaction with biological tissue [1]–[3]. An understanding of optical physics is therefore essential for surgeons and clinicians who use these technologies in daily practice. Processes such as absorption, scattering, refraction, and reflection determine how light propagates through tissue, how images are formed, and how optical energy is deposited during diagnostic and therapeutic interventions. Biological tissues are optically heterogeneous in nature and consist of several absorbing chromophores, including blood, water, melanin, fat, and yellow pigments. In addition to these light absorbing components, tissues also contain various scattering structures that exist at the cellular and subcellular levels, which together influence the way light propagates through biological tissue. As a result, optical signals detected during surgery represent a complex combination of tissue composition, structure, and physiological state, and their correct interpretation depends on a clear appreciation of the underlying

physical mechanisms [1], [2]. This article forms part of an ongoing series on foundational physics concepts relevant to surgical practice. Previous articles in this series have addressed key topics including mechanical principles in surgery, electrical and electrosurgical systems, medical imaging physics, and energy-based surgical devices [4]–[7]. The present review focuses specifically on optical physics and optical technologies used in surgery. While conceptually aligned with the earlier publications, this review is intended to be fully self-contained and does not require familiarity with previous parts of the series.

Advances in biomedical optics over recent decades have expanded the clinical use of light beyond conventional illumination and magnification. Imaging modalities such as optical coherence tomography provide depth-resolved visualization of tissue microstructure based on low-coherence interferometry [3] [8,9]. Fluorescence-guided surgical techniques enhance contrast between healthy and pathological tissue through endogenous or exogenous fluorophores, improving intraoperative visualization and margin assessment [10], [11]. Spectroscopic approaches, including Raman and diffuse reflectance spectroscopy, enable real-time assessment of tissue biochemical composition by probing wavelength-dependent absorption and inelastic scattering processes [12], [13]. In parallel, optical therapeutic techniques such as laser surgery and photodynamic therapy exploit controlled light–tissue interactions to achieve precise tissue modification and selective cytotoxic effects [14]–[16]. As optical technologies become increasingly integrated into surgical workflows, the gap between device complexity and user-level understanding continues to widen. Although surgeons are not required to design optical systems, a foundational knowledge of optical physics is critical for effective utilization of optical instruments, correct interpretation of optical information, and recognition of technique-specific limitations. This need is further emphasized by the rapid emergence of new optical imaging and therapeutic methods that are transitioning from research environments into clinical practice.

The purpose of this review is to describe the fundamental principles of optical physics that underlie optical diagnostic and surgical instruments used in clinical practice. Emphasis is placed on light–tissue interaction, optical components, imaging mechanisms, and therapeutic applications that are directly relevant to surgery. Established optical technologies commonly used by surgeons are discussed alongside emerging techniques that are shaping future directions in surgical diagnostics and therapy. By integrating physical principles with clinical context, this review aims to support surgeons and clinicians in understanding and interpreting optical technologies used in modern surgical care. A preliminary version of this work has been made available as a preprint to facilitate early dissemination [17]. The present manuscript represents a substantially revised and expanded version, incorporating additional scientific detail, updated references, and enhanced clinical context.

2 Fundamental Optical Physics Relevant to Surgery

Optical technologies used in surgery are based on established principles that describe the behavior of light and its interaction with biological tissues. In surgical environments, light does not simply travel in straight paths as assumed in ideal optical systems. Instead, it encounters complex biological structures that absorb and scatter radiation, influencing how light propagates within tissue. Biological tissues present a complex optical environment in which light propagation is strongly influenced by both absorption and scattering processes. Variations in tissue composition, structural organization, and molecular constituents alter the way light is transmitted, redirected, or attenuated within the tissue. Consequently, factors such as the wavelength of illumination, the intrinsic optical characteristics of the tissue, and the configuration of the optical system play a critical role in determining how light energy is delivered and how optical signals are detected during surgical procedures. Understanding these underlying optical principles is essential for interpreting optical signals and for appreciating the capabilities and limitations of surgical optical instruments used for imaging, diagnosis, and therapy. Fundamental principles of optical physics determine

how light propagates through biological tissue and how optical signals are generated, detected, and interpreted in surgical applications. Wavelength-dependent absorption, scattering, and refractive index variations define penetration depth, contrast, and resolution, influencing the performance of both diagnostic and therapeutic optical technologies. An understanding of these principles provides the necessary foundation for effective clinical use of optical instruments in surgery.

2.1 Nature of Light and Optical Radiation in Medicine

Light is a form of electromagnetic radiation consisting of oscillating electric and magnetic fields that can propagate through space and through various media. In a vacuum, it travels at a constant speed of approximately 3×10^8 m/s. When light enters a material medium such as air, water, or biological tissue, its speed decreases and its direction may change depending on the optical properties of the medium. Electromagnetic radiation exists over a broad range of wavelengths and frequencies, collectively known as the electromagnetic spectrum (Figure 1). This spectrum extends from high energy, short wavelength radiation such as gamma rays and X rays to longer wavelength radiation including ultraviolet, visible light, infrared, microwaves, and radio waves.

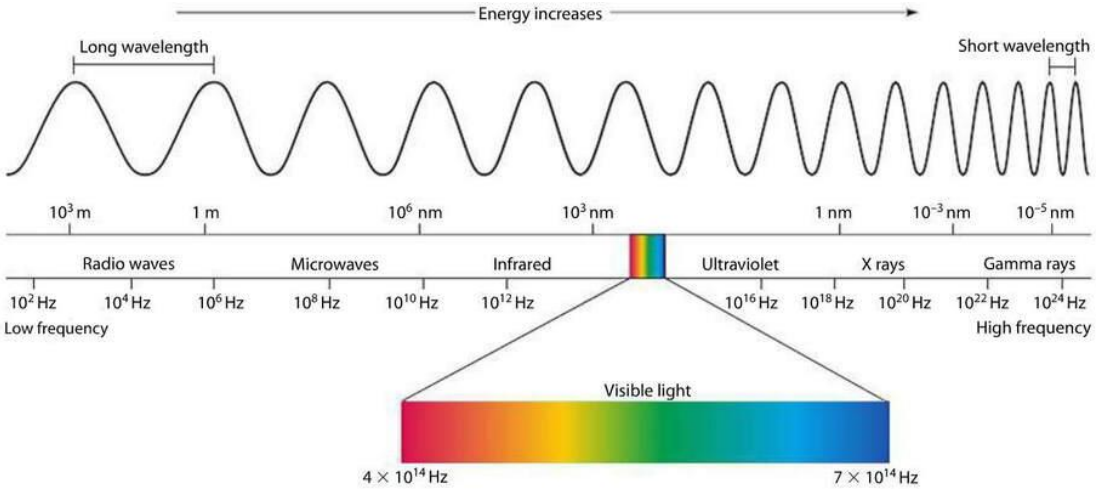

Figure 1: *Electromagnetic spectrum showing the major regions of radiation*

The energy carried by optical radiation is delivered in discrete packets known as photons. The energy of each photon depends on the frequency or wavelength of the radiation and is described by Planck's relation:

$$E = h\nu = \frac{hc}{\lambda}$$

where h is Planck's constant, ν is frequency, c is the speed of light, and λ is wavelength. In biological systems, optical radiation in the ultraviolet, visible, and infrared regions interacts mainly with molecules by inducing electronic and vibrational transitions. These interactions occur without causing ionization, which is typical of higher energy radiation. Because of these properties, visible and near infrared light are widely used in medical diagnostic and therapeutic applications [1], [2]. In surgical practice, optical technologies predominantly operate within the visible and near-infrared spectral ranges. These wavelengths offer a balance between tissue penetration, spatial resolution, and safety. Shorter wavelengths provide higher resolution but are more strongly absorbed and scattered, while longer wavelengths penetrate deeper into tissue with reduced scattering but lower spatial resolution. This trade-off underlies the selection of illumination and imaging wavelengths in surgical systems [3].

2.2 Optical Properties of Biological Tissue

Biological tissue is an optically heterogeneous medium composed of multiple structural and biochemical components. Its interaction with light is characterized by wavelength-dependent optical properties, including absorption coefficient, scattering coefficient, anisotropy factor, and refractive index. These parameters collectively determine how light propagates through tissue and how optical signals are generated and detected [1], [8]. Absorption in tissue arises from endogenous chromophores such as hemoglobin, melanin, water, and lipids. Each chromophore exhibits characteristic absorption spectra that influence optical contrast and penetration depth. Scattering results from refractive index mismatches between cellular and subcellular structures, including cell membranes, organelles, and collagen fibers. In most soft tissues, scattering dominates over absorption in the visible and near-infrared ranges, leading to diffuse light transport rather than ballistic propagation [1], [9]. The refractive index of biological tissues typically lies between 1.35 and 1.55 and depends on tissue composition, hydration, and wavelength as shown in Table 1. Variations in refractive index contribute to both scattering behavior and boundary reflections at tissue interfaces. These optical properties are central to the performance of imaging systems such as microscopes, endoscopes, and optical coherence tomography devices [10], [11].

Table 1: *Typical refractive index of various biological tissues*

Tissue	Refractive Index	Wavelength Range (nm)
Water	1.33	Visible
Blood	1.35-1.41	400-800
Skin (epidermis)	1.37-1.45	400-700
Dermis	1.38-1.42	400-800
Muscle tissue	1.36-1.41	400-900
Fat/Adipose tissue	1.46-1.48	400-800
Cornea (eye)	1.376	589
Lens (eye)	1.386-1.406	589
Retina	1.36-1.38	400-700
Liver Tissue	1.37-1.41	400-800
Brain (gray matter)	1.36-1.38	400-700
Bone (compact)	1.53-1.55	400-800

2.3 Light–Tissue Interaction Mechanisms

When light encounters biological tissue, several interaction processes occur simultaneously. At tissue boundaries, partial reflection and refraction arise due to differences in refractive index between media. Within tissue, photons may be absorbed or scattered multiple times before exiting or being fully attenuated. The relative contribution of each process depends on wavelength, tissue composition, and structural organization [1], [8]. Elastic scattering preserves photon energy and primarily provides structural information. It is responsible for image contrast in many optical imaging modalities and plays a dominant role in determining penetration depth and resolution. Inelastic scattering, such as Raman scattering,

involves energy exchange between photons and molecular vibrational modes, producing spectrally shifted signals that encode biochemical information [12], [13]. Absorption converts optical energy into other forms, most commonly heat or chemical excitation. In diagnostic imaging, absorption contributes to contrast based on chromophore concentration. In therapeutic applications, absorption governs thermal effects during laser surgery and photochemical reactions during photodynamic therapy. The balance between absorption and scattering defines the effective optical penetration depth and determines whether an optical technique is primarily surface-sensitive or capable of probing subsurface structures [14], [15].

2.4 Optical Penetration Depth and the Optical Window of Tissue

The depth to which light can penetrate biological tissue is strongly wavelength dependent as represented in Figure 2. Short-wavelength visible light experiences strong absorption and scattering, resulting in shallow penetration. In contrast, red and near-infrared wavelengths penetrate more deeply due to reduced absorption by hemoglobin and lower scattering coefficients. The spectral region between approximately 600 and 1200 nm is commonly referred to as the optical window of tissue, where optical attenuation is minimized [2], [3]. This optical window underpins the design of many surgical imaging and therapeutic systems, including near-infrared fluorescence imaging, optical coherence tomography, and photodynamic therapy. Selection of wavelengths within this region allows improved signal detection from subsurface structures while maintaining non-ionizing and clinically safe exposure levels.

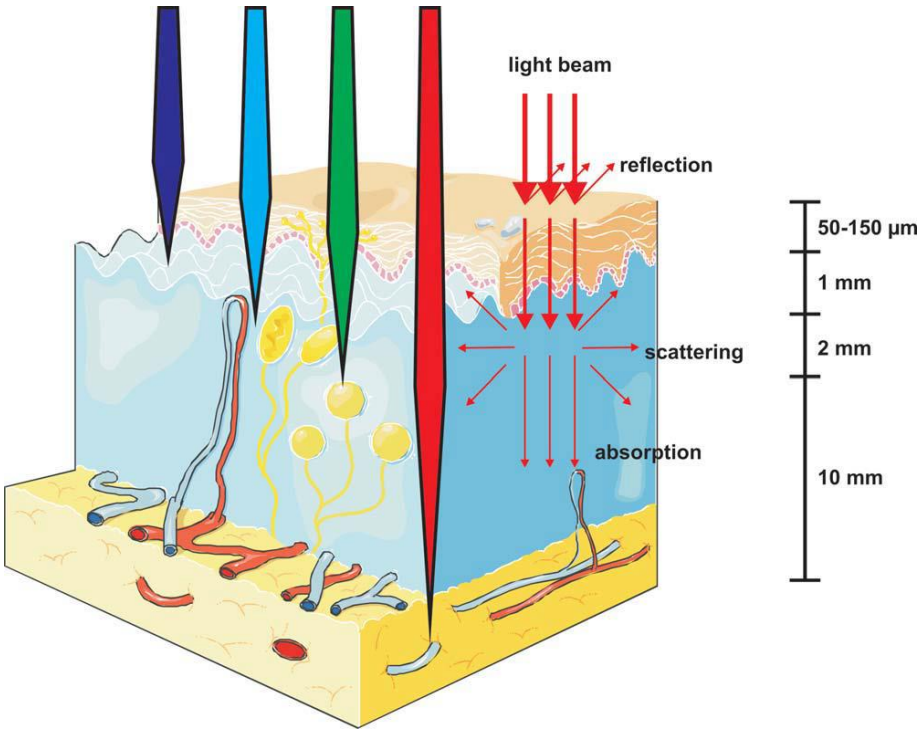

Figure 2: *penetration depth of light in tissue.*

3 Optical Components and Light Sources in Surgery

Optical components and light sources constitute the physical infrastructure of surgical visualization, imaging, and light-based therapy. Their optical characteristics determine illumination quality, image contrast, spatial resolution, penetration depth, and energy deposition in tissue. In surgical environments, optical systems must also satisfy clinical requirements related to sterility, ergonomics, thermal safety, and compatibility with minimally invasive techniques. A clear understanding of these components and their governing physical principles is essential for effective and safe surgical use. Optical components and light sources form the foundation of surgical visualization and optical therapy. Advances

in illumination technology, laser systems, fiber optics, and optical design have improved image quality, precision, and safety in surgical practice. Understanding the physical principles governing these components is essential for effective utilization and interpretation of optical technologies in surgery.

3.1 Illumination Sources Used in Surgical Systems

Surgical illumination sources are designed to provide sufficient brightness, appropriate spectral content, and spatial uniformity while minimizing heat generation and glare. Historically, incandescent and halogen lamps were used in surgical microscopes and headlamps due to their broad spectral output and good color rendering. These sources generate light through thermal emission from a heated filament, resulting in low energy efficiency and significant heat production. Excessive thermal output can contribute to tissue drying and discomfort in the operating room, leading to their gradual replacement in modern systems [18]. Discharge lamps generate light through electrical excitation of ionized gas or vapor. While they offer higher luminous efficiency than incandescent sources, their spectral output is often discontinuous and less controllable. Additionally, their size, warm-up time, and limited spectral flexibility restrict their suitability for advanced optical imaging and fluorescence-based applications in surgery [19].

Light emitting diodes have become the dominant illumination source in contemporary surgical lighting and imaging systems. LEDs produce light through electroluminescence in semiconductor materials, where photon emission occurs during electron–hole recombination across a band gap. The emission spectrum of an LED is determined by the semiconductor composition, allowing controlled generation of specific wavelengths or broadband white light through phosphor conversion. Compared with conventional lamps, LEDs offer high energy efficiency, long operational lifetime, rapid switching capability, and minimal heat generation at the emission site [20].

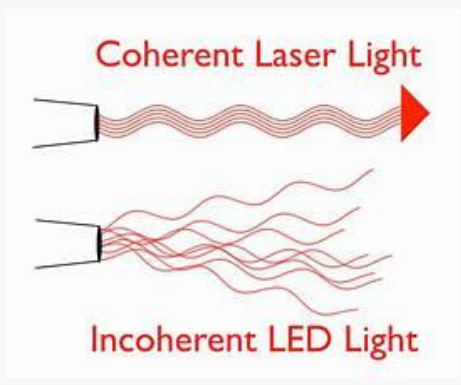

Figure 3: *Difference between coherent and non-coherent light source*

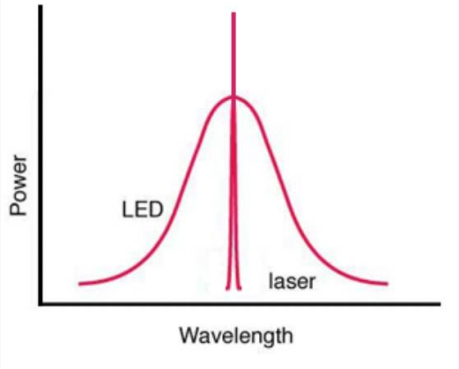

Figure 4: *Difference in intensity of LASER and LED*

In surgical applications, LED based illumination provides stable color rendering, controllable intensity, and compatibility with optical filters used in fluorescence imaging. As shown in Figure 3, LED emission is non-coherent, with light waves emitted in random phase and multiple directions, unlike the coherent and phase aligned waves produced by laser sources. This difference affects the ability to focus and confine light. Figure 4 illustrates the contrast in output intensity between the two sources, where lasers generate a highly concentrated and high intensity beam, while LED emission is broader and less radiant. These characteristics limit the use of LEDs in applications that require high irradiance or precise spatial energy delivery.

3.2 Lasers as Optical Sources for Surgery

Lasers are widely used in surgical applications due to their ability to deliver highly controlled optical energy with defined wavelength, spatial coherence, and temporal characteristics. Laser radiation is generated through stimulated emission, producing monochromatic and coherent light that can be focused to small spot sizes or delivered through optical fibers. The interaction of laser light with tissue is governed primarily by wavelength-dependent absorption and scattering properties. At wavelengths strongly absorbed by water or specific chromophores, laser energy is confined to superficial layers, enabling precise cutting and vaporization. At wavelengths with lower absorption, deeper penetration occurs, resulting in volumetric heating and coagulation. Temporal characteristics of laser delivery further influence tissue response. Continuous-wave lasers produce sustained heating, while pulsed lasers can confine energy deposition to short time scales, reducing thermal diffusion and collateral damage [21], [22]. Laser–tissue interactions are commonly categorized as photothermal, photomechanical, or photochemical. Photothermal effects dominate most surgical laser applications, where controlled heating leads to incision, ablation, or hemostasis. Photomechanical effects arise at high peak powers and short pulse durations, generating stress waves that disrupt tissue structure. Photochemical effects occur when laser light activates molecular processes, as in photodynamic therapy. Selection of laser parameters must balance precision, penetration depth, and safety to avoid unintended tissue injury [22].

3.3 Optical Fibers and Light Delivery Systems

Optical fibers play a central role in modern surgical optics by enabling flexible and controlled delivery of light to internal anatomical sites. An optical fiber consists of a high-refractive-index core surrounded by a lower-refractive-index cladding. Light propagation within the fiber occurs through total internal reflection at the core–cladding interface, provided that the angle of incidence exceeds the critical angle (Figure 5).

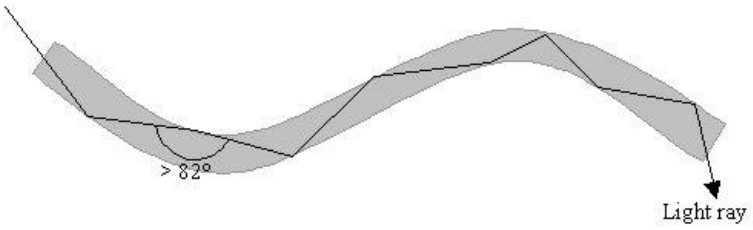

Figure 5: *optical fibre cable indicating total internal reflection with critical angle*

The numerical aperture of a fiber determines its light acceptance and collection efficiency. Fibers with higher numerical aperture collect light over a wider range of angles but may exhibit increased modal dispersion. In medical applications, fibers are typically fabricated from silica glass and coated with protective polymer layers to enhance mechanical strength and biocompatibility [2], [23]. Two main types of fiber bundles are used in surgical systems. Incoherent fiber bundles transmit illumination light without preserving spatial information and are commonly used for endoscopic lighting. Coherent fiber bundles preserve the relative spatial arrangement of individual fibers, enabling image transmission in flexible endoscopes. The spatial resolution of coherent bundles is limited by fiber diameter and packing density, placing a fundamental constraint on image quality [24]. Fiber-based delivery allows the light source to be separated from the surgical field, reducing heat generation and improving sterility. However, bending losses, connector degradation, and fiber breakage can reduce transmission efficiency and image quality over time, emphasizing the importance of proper handling and maintenance.

3.4 Optical Components in Surgical Imaging Systems

Surgical imaging instruments incorporate lenses, mirrors, beam splitters, and optical filters to control light propagation and image formation. Objective lenses collect light reflected or emitted from tissue and form intermediate images that are magnified by eyepieces or projected onto digital detectors. Key optical parameters include magnification, field of view, numerical aperture, and depth of focus. Higher numerical aperture improves resolution and light collection efficiency but reduces depth of field, which can limit usability in uneven surgical fields. Optical filters are used to isolate specific wavelength bands for diagnostic or therapeutic purposes. In fluorescence-guided surgery, excitation and emission filters suppress background illumination and enhance detection of fluorescent signals [25]. Beam splitters allow simultaneous visualization, image capture, and integration of multiple imaging modalities within a single optical pathway [26]. The optical design of surgical instruments must balance resolution, brightness, and ergonomic constraints. Aberrations, misalignment, and contamination of optical surfaces can degrade image quality and must be minimized through careful system design and maintenance.

3.5 Surgical Lighting Systems

Surgical lighting systems are designed to provide uniform, shadow-free illumination of the operative field with minimal glare and heat generation. Modern operating room lights predominantly employ arrays of high-intensity LEDs arranged to reduce shadow formation and provide adjustable illuminance and color temperature. Key performance parameters of surgical lighting include illuminance, typically measured in lux, color rendering index, and depth of illumination. High color rendering is essential for accurate tissue differentiation, while appropriate spectral composition can reduce visual fatigue. Advanced lighting systems allow adjustment of intensity and color temperature to suit specific surgical tasks [27]. Specialized lighting solutions include head-mounted lights and in-cavity illumination systems. Head-mounted lights provide coaxial illumination aligned with the surgeon's line of sight, improving visibility in deep or narrow surgical fields. In-cavity lighting systems, often coupled to optical fibers, provide localized illumination where overhead lighting is insufficient.

3.6 Optical Safety and Energy Considerations

The use of optical energy in surgery requires careful consideration of tissue safety and regulatory standards. Excessive irradiance or prolonged exposure can result in thermal injury, photochemical damage, or unintended stimulation of sensitive structures. Safety guidelines define permissible exposure limits based on wavelength, exposure duration, and tissue type [28]. Thermal effects depend on tissue absorption properties, perfusion, and heat conduction. Pulsed illumination, controlled power delivery, and fiber-based systems are commonly employed to limit heat accumulation. Laser systems require additional safety measures, including protective eyewear, interlocks, and controlled beam paths, to prevent accidental exposure to patients and staff.

4 Optical Diagnostic Instruments Used in Surgery

Optical diagnostic instruments are essential tools for intraoperative visualization and physiological assessment. These devices convert optical signals generated at tissue surfaces or within tissue into visual or quantitative information that guides surgical decision-making. Their diagnostic performance is governed by fundamental principles of geometrical and physical optics, including image formation, magnification, numerical aperture, depth of field, and wavelength-dependent tissue interaction. Understanding these principles is critical for correct interpretation of optical findings and for recognizing device-specific limitations in surgical settings. Optical diagnostic instruments are indispensable in surgical practice, providing magnified visualization and physiological information through controlled light–tissue interaction. Their performance and limitations are governed by principles of geometrical and physical optics. A clear

understanding of these principles enhances accurate interpretation and effective clinical use of optical diagnostic technologies.

4.1 Surgical Microscopes and Magnification Systems

Surgical microscopes provide high-resolution, magnified visualization of anatomical structures and are widely used in neurosurgery, ophthalmology, otolaryngology, plastic surgery, and microsurgical procedures. The optical performance of a surgical microscope is determined by the objective lens, magnification system, numerical aperture, illumination geometry, and imaging pathway [2], [15]. Schematic diagram of microscope is illustrated in Figure 6.

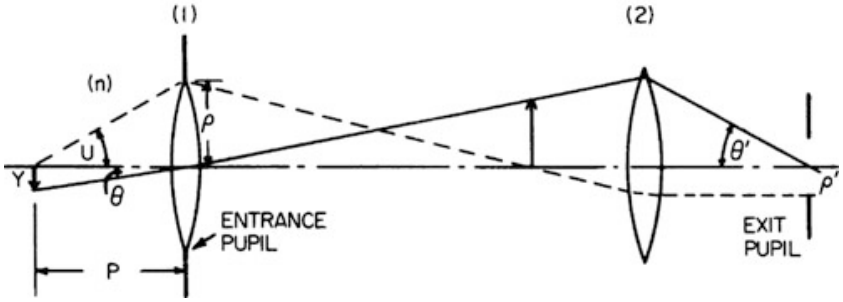

Figure 6: *Schematic diagram of basic optical instruments*

Magnification is achieved through compound optical systems consisting of objective lenses, intermediate magnification changers, and binocular eyepieces or digital imaging sensors. Increasing magnification enhances the visibility of fine structural details but reduces the field of view and depth of focus. Depth of focus is inversely related to numerical aperture and magnification, which can limit the ability to maintain sharp focus across uneven tissue surfaces during surgery [2]. Resolution in optical microscopes is fundamentally limited by diffraction and is described by the Abbe criterion. Lateral resolution improves with shorter illumination wavelengths and higher numerical aperture but at the expense of working distance and depth of field. These trade-offs are clinically relevant, as excessive magnification or high numerical aperture may reduce ergonomic flexibility and increase sensitivity to motion [15], [26]. Modern surgical microscopes incorporate coaxial illumination to minimize shadowing and enhance contrast. Many systems also integrate fluorescence imaging and digital image capture, enabling visualization of functional or molecular contrast in addition to structural detail.

4.2 Endoscopes and Fiber-Optic Imaging Systems

Endoscopes enable visualization of internal anatomical structures through minimally invasive access. Optical endoscopic systems consist of an illumination pathway, an imaging pathway, and a distal optical assembly that focuses light onto tissue and collects reflected or emitted photons. The quality of endoscopic imaging depends on fiber characteristics, lens design, illumination uniformity, and spectral properties of the light source [27]. Flexible endoscopes typically use coherent fiber bundles for image transmission. Each fiber acts as an individual image element, and the overall spatial resolution is limited by fiber diameter and packing density. Typical fiber diameters range from approximately 8 to 15 micrometers, which imposes a fundamental limit on image resolution. Rigid endoscopes often employ relay lens systems, which provide higher resolution and brightness but lack flexibility [25]. Illumination in endoscopy is commonly delivered via incoherent fiber bundles coupled to high-intensity LED or discharge light sources. The spectral composition of illumination influences tissue contrast and color perception. Narrow-band illumination techniques exploit wavelength-dependent absorption by hemoglobin and other chromophores to enhance visualization of vascular structures and mucosal abnormalities [10]. Optical losses in endoscopic systems arise from fiber bending, coupling inefficiencies, and scattering at optical interfaces. These losses

reduce signal-to-noise ratio and may affect image interpretation, particularly in deep or anatomically complex regions.

4.3 Otoscopes, Ophthalmoscopes, and Retinoscopes

Handheld optical diagnostic instruments such as otoscopes, ophthalmoscopes, and retinoscopes rely on simple yet precise optical configurations tailored to specific anatomical regions. Despite their apparent simplicity, their diagnostic performance is governed by fundamental optical principles [23]. An otoscope combines a focused illumination system with a magnifying lens to visualize the ear canal and tympanic membrane. The geometry of the speculum, illumination angle, and lens magnification determine the effective field of view and image clarity. Specular reflections and scattering from moist tissue surfaces can reduce contrast, emphasizing the importance of proper illumination alignment. The design of Otoscope consists of handle and a head (figure 7)[29]. Handle is long for easy grip and contains batteries to power and integrated light. Head consist of magnifying lens on the eyepiece with magnification of 8diaptor, a cone shaped disposable plastic speculum at the distal end, integrated light source (lamp, LED or bulb). Doctors fitted speculum on the Otoscope and instead it into the patient's ear by straightens the ear canal via pulling on the ear. It provide the two dimensional view of ear canal.

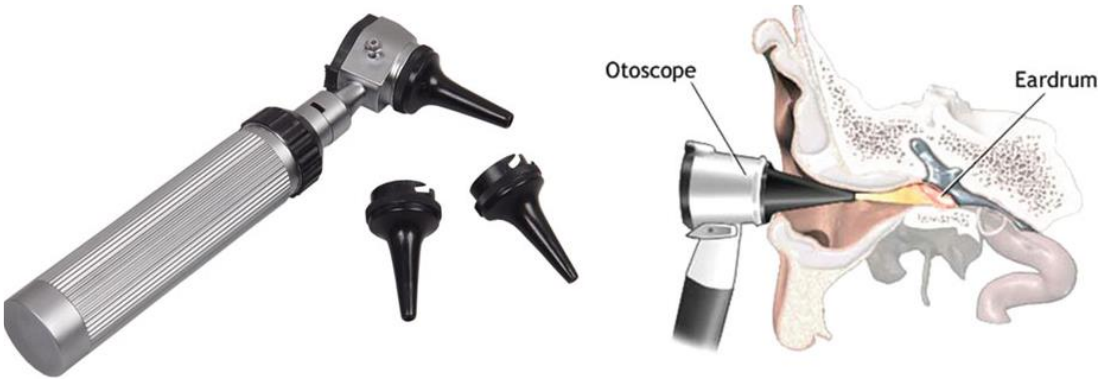

Figure 7: *Otoscope for inspection of ear canal*

Ophthalmoscope is an instrument used to examine the interior of the eye and its back structure (called fundus) through pupil by directing a light beam into the eye and looking its back reflection. Fundus consists of blood vessels, optic nerve and retina which detect the transmitted image through the cornea and a lens. With the helped of ophthalmoscope doctor can detect any problem or disease of retina and vitreous humor. Modern ophthalmoscope consists of two parts: one for illumination and other for viewing purpose. Illumination part has source of light (tungsten or halogen bulb), a condenser lens system, a reflector (mirror, or prism) to illuminate the eye through central hole. Viewing system made of sight hole and a rotating wheel with lens of difference power for focus. Lenses are selected for clear vision of eye structure at any depth and compensate the errors between the patient and examiner. Figure 8 illustrate the schematic diagram of ophthalmoscope which is designed to image the retina through the optical system of the eye. This requires careful alignment of illumination and viewing paths to minimize corneal reflections and compensation for refractive errors. Direct ophthalmoscopy provides high magnification but a limited field of view, while indirect ophthalmoscopy increases field coverage at the expense of image inversion and reduced magnification [15].

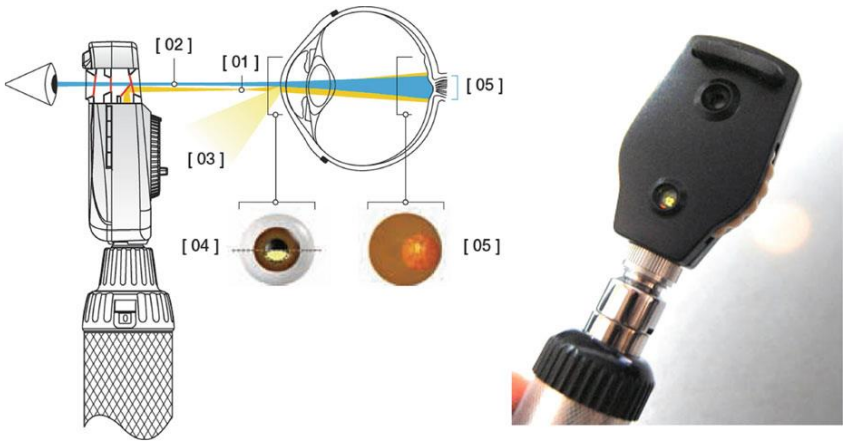

Figure 8: *Schematic diagram of ophthalmoscope (01) is the light from the light source, (02) reflected light from the cornea and iris, (03) image observe at pupil, (04) image observe at back of eye, (05) image of courtesy of heine.*

Retinoscopy is based on observing the movement of light reflected from the retina as illumination is swept across the pupil. In Retinoscope (shown in figure 9), a light source, a condensing lens and a mirror. Mirror either semi-transparent or a hole in the centre of the mirror through which an optometrist view the patient eye. In a procedure, retinoscope shine the light through pupil and optometrist move the light vertically and horizontally across the patient eye and observe the reflex. If the reflex move in the same direction then patient eye requires plus power for correction (i.e; myopia) and if reflex move against direction then need power negative power for correction (hyperopia). To determine the corrective refractive power, lenses of increasing power were placed in front of eye and change in the direction and observe the pattern of reflex. Optometrist keep changing the lenses until he observe the adequate focusing on the retina. The behavior of the retinal reflex depends on the refractive state of the eye and the relative positions of focal planes. Interpretation of retinoscopic findings requires an understanding of geometrical optics and image formation.

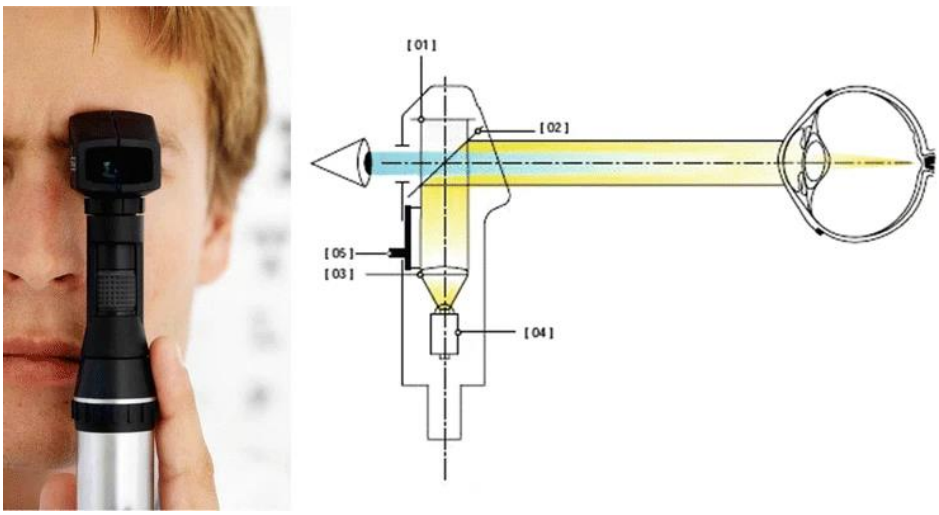

Figure 9: *modern retinoscope. An integrated lamp or LED light source (4) shines light through a collimating lens (3) onto a partially reflective mirror (2), which directs the light to the eye. The back-reflected light from the fundus and the cornea is examined by the doctor through the eyepiece (1) and focus adjusted using the lens dial (5) (Image courtesy of Heine) [30]*

4.4 Pulse Oximetry and Optical Physiological Monitoring

Pulse oximetry is a widely used noninvasive optical technique for monitoring arterial oxygen saturation during surgery and anesthesia. The method is based on differential absorption of red and near-infrared light by oxygenated and deoxygenated hemoglobin [31, 32]. Light emitted by LEDs passes through vascular tissue, where pulsatile changes in arterial blood volume modulate the detected signal (Figure 10). By isolating the alternating component of the signal and applying the Beer–Lambert law, relative concentrations of hemoglobin species can be estimated. Scattering, tissue heterogeneity, and motion artifacts introduce uncertainties that must be considered during interpretation.

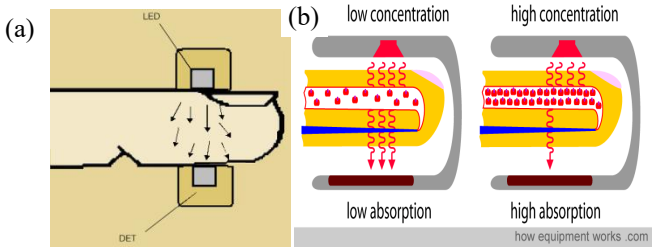

Figure 10: *(a) Pulse oximeter with passage of light, (b) indicate High and low concentration of haemoglobin impact the intensity of light*

The accuracy of pulse oximetry depends on wavelength selection, sensor geometry, and calibration assumptions. Reduced perfusion, pigmentation, and ambient light interference can degrade performance. Despite these limitations, pulse oximetry remains a cornerstone of perioperative monitoring due to its simplicity, rapid response, and clinical utility.

4.5 Interpretation Limitations and Clinical Considerations

While optical diagnostic instruments provide valuable real-time information, their outputs must be interpreted within the context of optical physics and biological variability. Image contrast may arise from multiple interacting factors, including absorption, scattering, illumination geometry, and surface topology. Optical artifacts such as glare, shadowing, saturation, and motion-induced blurring can obscure clinically relevant features. Most optical diagnostic tools provide surface-weighted or shallow-depth information unless combined with depth-resolved techniques. Awareness of these limitations is essential to avoid overinterpretation of optical findings and to integrate optical data appropriately with other clinical and imaging information.

5 Optical Imaging Techniques for Surgical Guidance

Optical imaging techniques extend surgical visualization beyond conventional white-light inspection by exploiting specific light–tissue interaction mechanisms to generate structural, functional, or molecular contrast. These techniques are increasingly integrated into surgical workflows to support intraoperative guidance, tissue differentiation, and margin assessment. Their clinical utility is determined by optical penetration depth, spatial resolution, contrast mechanisms, acquisition speed, and robustness in the operating room environment. Optical imaging techniques provide powerful tools for surgical guidance by enabling real-time visualization of tissue structure, function, and molecular composition. Their capabilities and limitations are determined by fundamental principles of light–tissue interaction. Understanding these principles supports accurate interpretation and effective integration of optical imaging technologies into surgical practice.

5.1 Optical Coherence Tomography

Optical coherence tomography (OCT) is a noncontact imaging modality that provides cross-sectional, depth-resolved images of tissue microstructure. OCT is based on low-coherence interferometry, in which backscattered light from tissue interferes with light from a reference arm only when their optical path lengths match within the coherence length of the source [3], [8]. Axial resolution in OCT is governed by the spectral bandwidth of the light source, with broader bandwidths enabling finer depth resolution, typically on the order of a few micrometers. Lateral resolution is determined by the numerical aperture of the focusing optics and is subject to diffraction limits similar to those of conventional optical microscopy. Imaging depth is limited by multiple scattering and absorption in tissue and typically ranges from 1 to 2 mm in soft tissues [1], [9]. OCT contrast arises primarily from variations in tissue scattering properties, which reflect differences in microstructural organization. As a result, OCT is particularly well suited for imaging layered tissues and identifying boundaries between tissue types. In surgical practice, OCT has been explored for intraoperative assessment of tissue architecture, guidance during microsurgical procedures, and evaluation of surgical margins. Interpretation of OCT images requires awareness that signal intensity reflects optical scattering rather than direct histological staining or cellular composition.

5.2 Fluorescence Imaging and Fluorescence-Guided Surgery

Fluorescence imaging is based on excitation of endogenous or exogenous fluorophores followed by emission of light at longer wavelengths. The spectral separation between excitation and emission allows selective detection of fluorescent signals against a suppressed background, resulting in high sensitivity to molecular contrast [10], [20]. In fluorescence-guided surgery, contrast enhancement is achieved using either intrinsic tissue fluorescence or externally administered fluorescent agents. Endogenous fluorophores include collagen, flavins, and nicotinamide adenine dinucleotide, while exogenous agents may be non-specific dyes or targeted probes designed to accumulate in diseased tissue. Near-infrared fluorophores are often preferred due to reduced tissue absorption and scattering, which improve penetration depth and signal-to-background ratio [10], [11]. Fluorescence signal intensity depends on fluorophore concentration, excitation irradiance, quantum yield, and tissue optical properties. Both excitation and emission light are attenuated by absorption and scattering, leading to depth-dependent signal loss. Consequently, fluorescence imaging is primarily sensitive to superficial tissue layers unless combined with tomographic or quantitative correction approaches. Despite these limitations, fluorescence-guided techniques have demonstrated clinical value in tumor margin visualization, vascular imaging, and sentinel lymph node detection.

5.3 Confocal Microscopy for Intraoperative Imaging

Confocal microscopy provides high-resolution optical sectioning by selectively detecting light from a narrow focal volume while rejecting out-of-focus photons through spatial filtering. This is achieved using pinholes placed at conjugate image planes in the illumination and detection pathways [27]. The spatial resolution of confocal microscopy depends on illumination wavelength and numerical aperture. Higher numerical aperture improves resolution but reduces depth of field and working distance. In biological tissue, imaging depth is limited by scattering and typically does not exceed a few hundred micrometers. Confocal systems can operate in reflectance or fluorescence modes, providing either structural contrast based on refractive index variations or molecular contrast from fluorescent labels. In surgical settings, confocal microscopy has been investigated for real-time assessment of tissue microarchitecture and cellular morphology. However, its limited field of view and shallow penetration depth restrict its application to surface-accessible tissues or probe-based intraoperative implementations.

5.4 Optical Spectroscopic Techniques

Optical spectroscopy provides biochemical and functional information by analyzing wavelength-dependent interactions between light and tissue. These techniques include diffuse reflectance spectroscopy and Raman spectroscopy, each probing different physical mechanisms [12], [13]. Diffuse reflectance spectroscopy measures light reflected from tissue to estimate absorption and scattering coefficients. These parameters can be related to chromophore concentrations, such as hemoglobin, and to microstructural features that influence scattering. Raman spectroscopy relies on inelastic scattering, in which incident photons exchange energy with molecular vibrational modes, producing spectrally shifted signals that serve as biochemical fingerprints of tissue constituents [13], [26]. Spectroscopic signals are inherently weak and sensitive to noise from ambient light, motion, and tissue heterogeneity. Accurate interpretation requires appropriate calibration and statistical analysis. Nevertheless, optical spectroscopy has shown promise for intraoperative tissue differentiation and detection of pathological changes without the need for tissue excision.

5.5 Interpretation and Clinical Integration

Optical imaging techniques provide complementary information that differs fundamentally from conventional visual inspection. Structural, functional, and molecular contrasts are governed by distinct optical mechanisms and must be interpreted accordingly. Depth sensitivity, spatial resolution, and susceptibility to artifacts vary across modalities, and no single optical technique provides comprehensive information for all surgical scenarios. Successful clinical integration of optical imaging requires understanding what each modality measures and how optical signals relate to underlying tissue properties. Awareness of limitations related to penetration depth, scattering, and signal attenuation is essential to avoid misinterpretation and to integrate optical data appropriately with other clinical findings.

6 Optical Therapeutic Technologies in Surgery

Optical therapeutic technologies employ controlled delivery of light to induce localized biological effects for surgical intervention. In contrast to diagnostic optical techniques, therapeutic applications intentionally deposit optical energy within tissue to achieve cutting, coagulation, ablation, or photochemical activation. The resulting tissue response is governed by wavelength-dependent absorption, temporal characteristics of illumination, and tissue thermal and photochemical properties. A clear understanding of these mechanisms is essential for safe and effective clinical use. Optical therapeutic technologies enable precise and localized surgical intervention through controlled light–tissue interaction. Laser surgery, photodynamic therapy, and photobiomodulation represent distinct therapeutic mechanisms governed by specific optical and biological principles. Understanding these principles is essential for optimizing therapeutic outcomes, minimizing risks, and ensuring safe clinical use of optical technologies in surgery.

6.1 Laser–Tissue Interaction in Surgical Procedures

Lasers are the most widely used optical therapeutic tools in surgery due to their ability to deliver spatially confined and spectrally selective energy. Laser radiation interacts with tissue primarily through absorption and scattering, converting photon energy into heat or initiating photochemical reactions depending on wavelength and exposure parameters [3], [22]. The biological response to laser irradiation depends on optical penetration depth and thermal diffusion. At wavelengths strongly absorbed by water or specific chromophores, laser energy is confined to superficial tissue layers, producing precise cutting and vaporization with limited penetration. At wavelengths with lower absorption, deeper penetration occurs, resulting in volumetric heating and coagulation. These effects are further influenced by the temporal delivery of energy. Continuous-wave lasers generate sustained heating, whereas pulsed lasers can confine energy deposition to short time intervals, reducing lateral thermal spread [22], [23]. Laser–tissue interactions

are commonly classified as photothermal, photomechanical, or photochemical. Photothermal effects dominate most surgical laser applications, where controlled heating leads to incision, ablation, or hemostasis. Photomechanical effects arise at high peak powers and short pulse durations, generating stress waves that disrupt tissue structure. Photochemical effects occur when laser light activates molecular processes, as in photodynamic therapy. Appropriate selection of laser wavelength, power, and pulse duration is critical to balance precision and safety and to minimize collateral tissue damage [22].

6.2 Photodynamic Therapy

Photodynamic therapy (PDT) is a light-activated treatment modality that combines a photosensitizing agent, optical irradiation, and molecular oxygen to produce localized cytotoxic effects. Individually, these components are non-toxic; however, their interaction generates reactive oxygen species that cause cellular damage and death [19], [20]. The mechanism of PDT begins with absorption of light by the photosensitizer, promoting it to an excited electronic state. Energy transfer from the excited photosensitizer to molecular oxygen results in the formation of singlet oxygen and other reactive species. These species oxidize cellular membranes, proteins, and nucleic acids, leading to apoptosis or necrosis depending on treatment conditions [20]. The therapeutic outcome of PDT depends on photosensitizer pharmacokinetics, light wavelength and fluence, tissue oxygenation, and timing between drug administration and illumination. Photosensitizers are typically designed to absorb in the red or near-infrared spectral region to improve tissue penetration. Light delivery can be achieved using surface illumination, optical fibers, or interstitial probes for deeper lesions. Although PDT offers high spatial selectivity and minimal systemic toxicity, its clinical applicability is limited by light penetration depth and oxygen availability in tissue [19], [20].

6.3 Photobiomodulation and Low-Level Light Therapy

Photobiomodulation refers to the application of low-intensity optical radiation to modulate cellular activity without inducing thermal damage. This approach typically uses visible or near-infrared light at irradiances well below those required for tissue ablation or coagulation [21]. The primary biological mechanism of photobiomodulation involves absorption of light by mitochondrial chromophores, leading to changes in cellular respiration, adenosine triphosphate production, and intracellular signaling pathways. These effects have been associated with enhanced tissue repair, modulation of inflammation, and reduction of pain responses. Unlike laser surgery, photobiomodulation aims to induce reversible biological responses rather than permanent tissue modification [21]. Therapeutic outcomes depend strongly on wavelength, dose, and treatment timing. Both insufficient and excessive light exposure can reduce effectiveness, highlighting the importance of understanding dose–response relationships. Although photobiomodulation is increasingly used as an adjunct therapy in surgical and rehabilitative contexts, its mechanisms continue to be refined through ongoing experimental and clinical studies.

6.4 Safety Considerations in Optical Therapy

The application of optical energy in surgery requires strict adherence to safety standards to prevent unintended tissue injury and occupational hazards. Excessive irradiance or prolonged exposure can result in thermal damage, photochemical injury, or stimulation of non-target tissues. Safety guidelines define permissible exposure limits based on wavelength, exposure duration, and tissue type [30]. Thermal effects depend not only on optical power but also on tissue perfusion and thermal conductivity, which influence heat dissipation. Pulsed illumination, controlled energy delivery, and fiber-based light transmission are commonly employed to limit heat accumulation. Laser systems require additional safety measures, including protective eyewear, beam containment, interlocks, and controlled operating protocols, to minimize accidental exposure to patients and operating room personnel.

7 Emerging Optical Technologies in Surgery

Continued advances in optical science, materials, and instrumentation are driving the development of new light-based technologies for surgical applications. Emerging optical techniques aim to provide improved contrast, functional and molecular information, and real-time feedback that extend beyond the capabilities of conventional visualization and imaging tools. These developments are enabled by progress in light sources, detectors, optical fibers, contrast agents, and computational methods. Although not all emerging technologies are yet in routine clinical use, many are undergoing active evaluation in intraoperative and translational settings. Emerging optical technologies aim to extend the role of light in surgery by providing molecular, functional, and biochemical information in real time. These approaches build on established principles of light–tissue interaction while incorporating advances in instrumentation and computation. As these technologies mature, understanding their physical basis and limitations will be critical for their effective and safe integration into surgical practice.

7.1 Molecular and Functional Optical Imaging

Molecular optical imaging seeks to visualize specific biological processes rather than relying solely on anatomical structure. This approach employs contrast mechanisms that reflect molecular expression, metabolic activity, or enzymatic function within tissue. In surgical applications, molecular imaging has the potential to improve identification of tumor margins, detect residual disease, and guide targeted resection [10], [20]. From a physical perspective, molecular optical imaging relies on the interaction of light with exogenous agents that possess well-defined absorption or emission characteristics. Targeted fluorescent probes are designed to accumulate selectively in diseased tissue through receptor binding or enzymatic activation. Near-infrared wavelengths are often favored due to reduced tissue absorption and scattering, which improve penetration depth and signal-to-background ratio compared with visible light [9], [12]. Functional optical imaging techniques, such as blood oxygenation and perfusion mapping, exploit wavelength-dependent absorption of hemoglobin to infer physiological status. These approaches provide indirect but clinically relevant information about tissue viability and vascular integrity during surgery. Interpretation of functional signals requires awareness of confounding factors such as scattering, motion, and heterogeneous tissue composition.

7.2 Intraoperative Optical Spectroscopy

Intraoperative optical spectroscopy aims to provide real-time biochemical and compositional information without the need for tissue excision or frozen section analysis. Spectroscopic techniques analyze wavelength-dependent interactions between light and tissue to extract information related to molecular composition and structure [12], [13]. Diffuse reflectance spectroscopy measures light reflected from tissue to estimate absorption and scattering coefficients, which can be related to chromophore concentrations and microstructural features. Raman spectroscopy probes inelastic scattering, producing spectral shifts corresponding to molecular vibrational modes that serve as highly specific biochemical fingerprints [13], [30]. The clinical utility of intraoperative spectroscopy depends on signal strength, acquisition speed, and robustness to motion and ambient light. Advances in detector sensitivity, fiber-optic probe design, and real-time data processing have improved feasibility in surgical environments. However, reliable interpretation still requires appropriate calibration and validation, as spectroscopic signals can be influenced by tissue heterogeneity and measurement geometry.

7.3 Advanced Fiber-Optic and Probe-Based Optical Systems

Emerging optical technologies increasingly rely on miniaturized fiber-optic probes that can be integrated into surgical instruments or deployed through minimally invasive access routes. These probes enable localized illumination, imaging, and signal collection in confined anatomical spaces where

conventional optics are impractical [2], [24]. Advances in fiber design, including multicore fibers, specialty fibers with tailored dispersion properties, and combined illumination-detection architectures, have expanded the range of optical functions that can be delivered through flexible platforms. Probe-based systems are being explored for subsurface imaging, localized spectroscopy, and targeted light delivery during surgery. Physical constraints related to probe diameter, numerical aperture, and bending tolerance impose limitations on resolution and signal collection efficiency. Nonetheless, continued refinement of probe architectures and coupling strategies is improving performance and clinical usability.

7.4 Challenges and Future Directions

Despite significant progress, several challenges limit the widespread adoption of emerging optical technologies in surgery. These include limited penetration depth, sensitivity to tissue heterogeneity, and variability in optical properties across patients. Integration into surgical workflows must also consider sterility, ergonomics, acquisition speed, and ease of interpretation. Future developments are expected to focus on multimodal systems that combine complementary optical techniques, improved contrast agents with higher specificity, and more robust real-time processing. Continued collaboration between optical scientists, engineers, and clinicians will be essential for translating emerging optical technologies into practical surgical tools.

8 Conclusion

Optical technologies play a central and expanding role in contemporary surgical practice. From illumination and magnification to advanced imaging and light-based therapy, surgical optical instruments rely on well-defined physical principles that govern light propagation and interaction with biological tissue. Processes such as absorption, scattering, refraction, and wavelength-dependent penetration directly influence image formation, contrast, resolution, and energy deposition, thereby determining the clinical performance and limitations of optical systems. This review has outlined the fundamental optical physics underlying commonly used diagnostic and therapeutic optical instruments in surgery. By examining optical components, light sources, imaging modalities, and therapeutic technologies within a unified physical framework, the article highlights how basic optical principles translate into practical clinical outcomes. Established tools such as surgical microscopes, endoscopes, pulse oximeters, lasers, and fluorescence-based systems were discussed alongside advanced techniques including optical coherence tomography, confocal imaging, spectroscopy, and photodynamic therapy. Emerging optical technologies extend the capabilities of surgical optics by providing molecular, functional, and biochemical information in real time. While these approaches offer significant potential to improve surgical precision and intraoperative decision-making, their successful integration into clinical practice depends on recognition of inherent optical limitations, including restricted penetration depth, sensitivity to tissue heterogeneity, and signal interpretation challenges. By linking optical physics with surgical applications, this review provides a coherent foundation for understanding how optical technologies function and how their outputs should be interpreted in practice. Such understanding is increasingly important as optical systems continue to evolve and assume a more prominent role in surgical diagnostics, guidance, and therapy.

9 Declarations

9.1 Competing Interests

The authors declare that there is no conflict of interest.

9.2 Publisher's Note

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