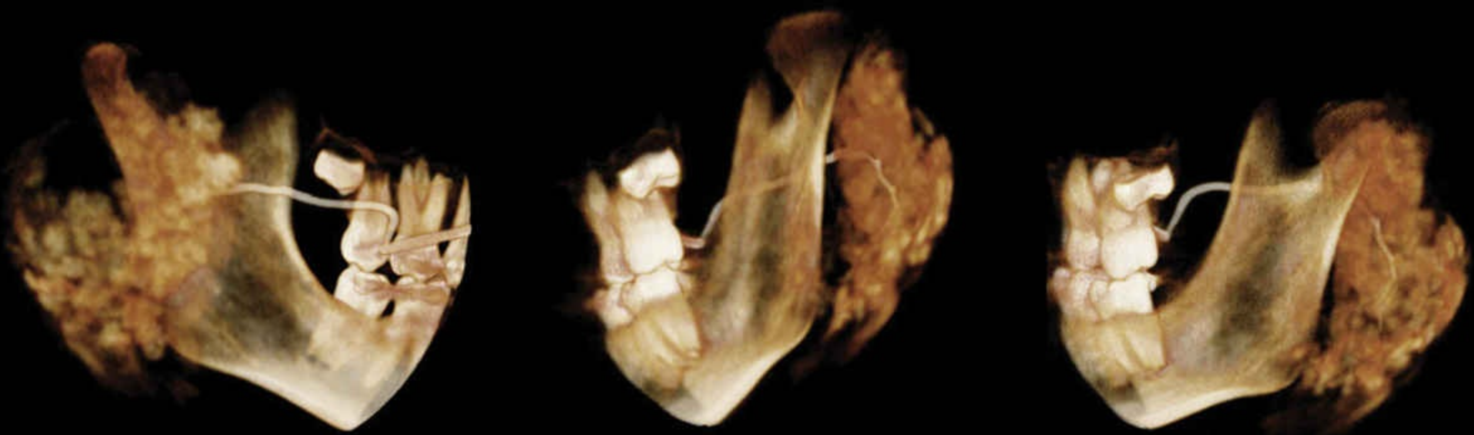
8th EDITION

White and Pharoah's

ORAL RADIOLOGY

Principles and Interpretation



Sanjay M. **Mallya**

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Principles and Interpretation

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Dedication

To our teachers and mentors, and our students, both past and present.

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Preface

We take on our roles as the new editors of this textbook with enthusiasm and energy. The previous seven editions, under the leadership of Professors Paul W. Goaz, Stuart C. White, and Michael J. Pharoah, presented the science of diagnostic oral and maxillofacial radiology to dental students worldwide for over three decades. We hope that our contributions continue this textbook's tradition of excellence and provide our readers with exceptional educational content that is current and scientifically based. The book encompasses the full scope of oral and maxillofacial radiology for the dental student and serves as a comprehensive resource for graduate students and dental practitioners.

Radiologic imaging is an integral component of diagnosis and treatment planning in general and specialty dental practices. Dentists have access to a variety of imaging modalities, either in their offices, or at imaging centers and hospitals. To optimally apply diagnostic imaging in patient care, dentists must understand the basic principles of radiographic image formation and interpretation. To this end, the book provides foundational knowledge, and related guidelines and regulations for the safe and effective use of x-rays, as well as in-depth knowledge on conventional and advanced imaging techniques used to evaluate oral and maxillofacial disease. This new edition also provides us the opportunity to discuss the latest developments in our field. With advances in digital dentistry, information from multiple digital sources is being combined to guide treatment planning or to fabricate appliances and restorations. Oral and maxillofacial radiology often forms the backbone of such integrated data. A new chapter—*Beyond Three-Dimensional Imaging*—introduces advanced applications of 3D imaging, including additive manufacturing. Since the last edition, several professional organizations have published imaging guidelines, technical reports and position statements that impact the practice of oral and maxillofacial radiology. This edition has been updated to incorporate new recommendations for quality assurance and updated guidelines for use of cone beam computed tomography in dentistry.

Dentists must be familiar with the key radiographic features of diseases of the maxillofacial region. This book provides comprehensive coverage of radiographic manifestations and the differential interpretation of diseases affecting the teeth, jaws, paranasal sinuses, salivary glands, and temporomandibular joints. The chapters emphasize the biological foundations of disease as they relate to their radiologic interpretation. To enhance integration of basic and clinical sciences, we include a new chapter that consolidates diseases affecting the structure of bone. Where applicable, radiographic appearances of disease are illustrated using not only conventional, 2-dimensional imaging but also advanced imaging, providing knowledge that is applicable in general and specialty dental practices.

The book also offers supplemental resources to instructors via the companion Evolve website (http://evolve.elsevier.com), including test banks and the image collection.

Our goal is to make the study of oral and maxillofacial radiology stimulating and exciting.

Sanjay M. Mallya BDS, MDS, PhD

Ernest W.N. Lam DMD, MSc, PhD, FRCD(C)

Acknowledgments

We thank our colleagues who have contributed as chapter authors. We appreciate their willingness to share their expertise and knowledge with our readers. This edition welcomes five new authors: Drs. Edwin Chang, Aruna Ramesh, Anitha Potluri, Aditya Tadinada, and Daniel Turgeon.

We thank Mr. John Harvey for creating the new illustrations of the disease processes that may be found in the latter chapters of the book.

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We thank the staff team from Elsevier whose tireless efforts helped keep the book's author and editorial team on track to meet production milestones: Caroline Dorey-Stein, Kathy Falk, Jennifer Flynn-Briggs, Lucia Gunzel, Alexandra Mortimer, Ramkumar Bashyam, and Rachel McMullen.

Finally, we thank Drs. Stuart White and Michael Pharoah for generously sharing their vast experience as the former editors of this book. Their feedback and advice have been invaluable.

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PART I
Foundations

OUTLINE

Physics

Sanjay M. Mallya

Abstract

This chapter provides basic knowledge on the nature of radiation, the operation of an x-ray machine, and the interactions of x-radiation with matter, with an emphasis on diagnostic x-radiation. This foundational knowledge is important for the safe and effective use of x-rays in dentistry.

Keywords

electromagnetic radiation; x-ray machine; DC x-ray unit; photoelectric absorption; compton scatter; bremsstrahlung radiation; kilovoltage; milliamperage; beam filtration; x-ray attenuation

One atom says to a friend, “I think I lost an electron.” The friend replies, “Are you sure?” “Yes,” says the first atom, “I'm positive.”

Radiologic examination is an integral component of the dentist's diagnostic armamentarium. Dentists often make radiographic images of patients to obtain additional information beyond that available from a clinical examination or their patient's history. Information from these images is combined with the clinical examination and history to make a diagnosis and formulate an appropriate treatment plan. This chapter provides basic knowledge on the nature of radiation, the operation of an x-ray machine, and the interactions of x-radiation with matter, with an emphasis on diagnostic x-radiation. This foundational knowledge is important for the safe and effective use of x-rays in dentistry.

Composition of Matter

Matter is anything that has mass and occupies space. The atom is the basic unit of all matter and consists of a nucleus containing protons and neutrons, and electrons that are bound to the nucleus by electrostatic forces. The classic view of the atom, the **Bohr model,** considers the structure of atoms like a solar system, with negatively charged electrons that travel in discrete orbits around a central, positively charged nucleus (Fig. 1.1A). The contemporary view, the **quantum mechanical model,** assigns electrons into complex three-dimensional orbitals with energy sublevels (see Fig. 1.1B).

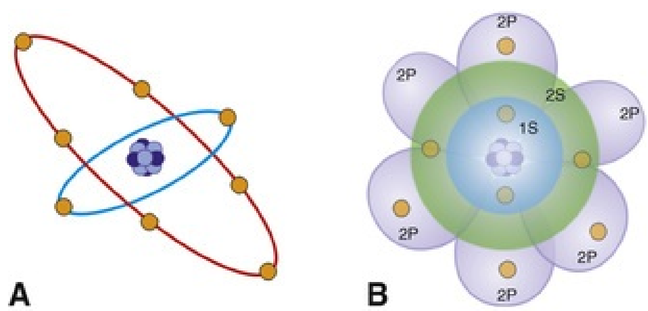

FIG. 1.1 (A) Schematic view of the Bohr model of the oxygen atom showing a nucleus with electrons that travel around the nucleus in circular orbits. (B) Schematic view of the quantum mechanical model of the oxygen atom. The central nucleus is surrounded by an electron cloud that represents probability plots of the location of the electron in a complex arrangement.

Atomic Structure

Nucleus

In all atoms except hydrogen, the nucleus consists of positively charged protons and neutral neutrons. A hydrogen nucleus contains a single proton. The number of protons in the nucleus, its **atomic number** *(Z)*, is unique to each element. Each of 118 known elements has a unique atomic number. The total number of protons and neutrons in the nucleus of an atom is its **atomic mass** *(A)*. The ratio of neutrons to protons determines the stability of the nucleus and is the basis of radioactive decay.

Electron Orbitals

Electrons are negatively charged particles that exist in the extranuclear space and are bound to the nucleus by electrostatic attraction. The *Bohr model* considers that electrons exist in discrete orbits or "shells" denoted as K, L, M, N, O, and P, with the K-shell being closest to the nucleus (see Fig. 1.1A). The shells are also described by a quantum number 1, 2, 3 …, with 1 being the quantum number for the K-shell. Each shell can hold a maximum of $2n^2$ electrons, where n is the quantum number of the shell.

The *quantum mechanical model* describes the electrons within three-dimensional orbitals, or electron clouds (see Fig. 1.1B). The electron orbitals are described based on their distance from the nucleus (*principal quantum number*; $n = 1, 2, 3 \ldots$) and their shape (designated s, p, d, f, g, h, and i). Only two electrons may occupy an orbital. The electron orbitals in order of filling are 1s, 2s, 2p, 3s, 3p, 3d, 4s, 4p, 4d, 4f … and so forth. The Bohr model and the quantum mechanical model both provide an adequate basis to conceptually understand diagnostic x-ray production and interactions.

The energy needed to overcome the electrostatic force that binds an electron to the nucleus is termed the **electron binding energy.** The electron binding energy is related to the atomic number and the orbital type. Elements with a large atomic number (high Z) have more protons in their nucleus and thus bind electrons in any given orbital more tightly than smaller Z elements. Within a given atom, electrons in the inner orbitals are more tightly bound than the more distant outer orbitals. Electron binding energy is the conceptual basis to understand ionization, which occurs when matter is exposed to x-rays.

Ionization

When the number of electrons in an atom is equal to the number of protons in its nucleus, the atom is electrically neutral. If a neutral atom loses an electron, it becomes a positive ion, and the free electron becomes a negative ion. This process of forming an ion pair is termed **ionization.** To ionize an atom, sufficient external energy must be provided to overcome the electrostatic forces, and free the electron from the nucleus. High-energy particles, x-rays, and ultraviolet radiation have sufficient energy to displace electrons from their orbitals and ionize atoms. Such radiations are referred to as **ionizing radiations.** In contrast, visible light, infrared and microwave radiations, and radio waves do not have sufficient energy to remove bound electrons from their orbitals and are

nonionizing radiations.

Nature of Radiation

Radiation is the transmission of energy through space and matter. It may occur in two forms: (1) electromagnetic and (2) particulate (Table 1.1). Practical applications of these radiations in healthcare are listed.

TABLE 1.1

Particulate Radiation

Particle	Symbol	Elementary Charge[a]	Rest Mass (amu)
Alpha	α	+2	4.00154
$Beta^+$ (positron)	β^+	+1	0.000549
$Beta^-$ (electron)	β^-	−1	0.000549
Electron	e^-	−1	0.000549
Neutron	n^0	0	1.008665
Proton	p	+1	1.007276

[a]Elementary charge of 1 equals that the charge of a proton or the opposite of an electron.

amu, Atomic mass units, where 1 amu = $\frac{1}{12}$ the mass of a neutral carbon-12 atom.

- Diagnostic imaging with projection radiography and computed tomography use x-rays, a category of electromagnetic radiation that is ionizing in nature.
- Magnetic resonance imaging (MRI, Chapter 13) uses electromagnetic radiations of significantly lower energies than x-rays and at energies that are nonionizing.
- Some radiopharmaceuticals used in diagnostic nuclear medicine emit particulate radiation. For example, ^{18}F-fluorodeoxyglucose (^{18}F-FDG) emits positrons, a key step in imaging with **positron emission tomography** (PET; Chapter 13).

• High-energy electromagnetic radiations (gamma rays, γ) and high-energy particulate radiations (electron beams and protons) are used in cancer therapy.

Electromagnetic Radiation

Electromagnetic radiation is the movement of energy through space as a combination of electric and magnetic fields. It is generated when the velocity of an electrically charged particle is altered. γ-Rays, x-rays, ultraviolet rays, visible light, infrared radiation (heat), microwaves, and radio waves all are examples of electromagnetic radiation (Fig. 1.2). γ-Rays originate in the nuclei of radioactive atoms. They typically have greater energy than x-rays. In contrast, x-rays are produced outside the nucleus and result from the interaction of electrons with large atomic nuclei, as in x-ray machines. The higher-energy types of radiation in the electromagnetic spectrum—ultraviolet rays, x-rays, and γ-rays—are capable of ionizing matter. Some properties of electromagnetic radiation are best explained by quantum theory, whereas others are most successfully described by wave theory.

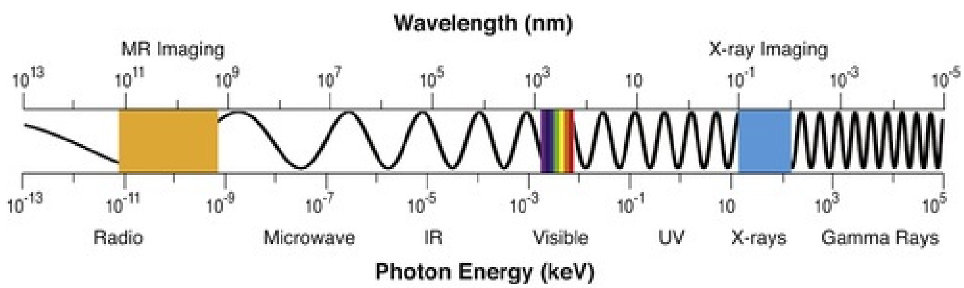

FIG. 1.2 Electromagnetic spectrum showing the relationship between photon wavelength and energy and the physical properties of various portions of the spectrum. Photons with shorter wavelengths have higher energy. Photons used in dental radiography *(blue)* have energies of 10 to 120 keV. Magnetic resonance *(MR)* imaging uses radio waves *(orange)*. *IR,* Infrared radiation; *UV,* ultraviolet radiation.

Quantum theory considers electromagnetic radiation as small discrete bundles of energy called **photons**. Each photon travels at the speed of light and contains a specific amount of energy, expressed with the unit **electron volt** (eV).

The wave theory of electromagnetic radiation maintains that radiation is propagated in the form of waves, similar to the waves resulting from a disturbance in water. Such waves consist of electric and magnetic fields oriented in planes at right angles to one another that oscillate perpendicular to the direction of motion (Fig. 1.3). All electromagnetic waves travel at the velocity of light ($c = 3.0 \times 10^8$ m/s) in a vacuum. Waves are described in terms of their wavelength (λ, meters) and frequency (ν, cycles per second, hertz).

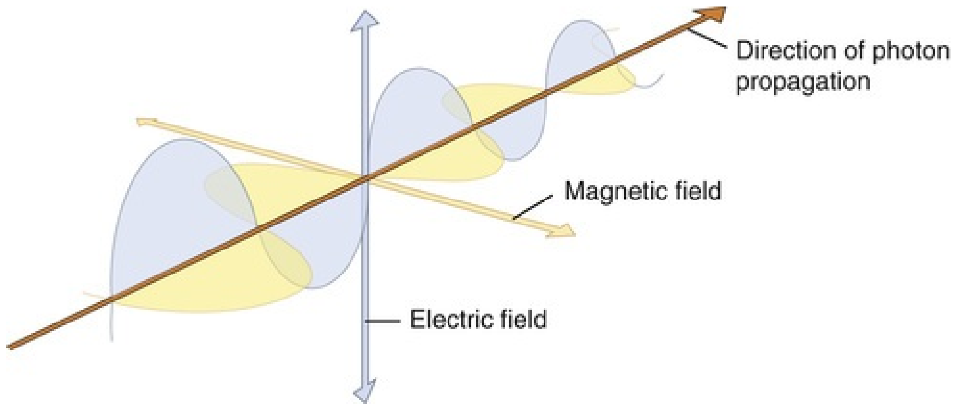

FIG. 1.3 Electric and magnetic fields associated with electromagnetic radiation.

Both theories are used to describe properties of electromagnetic radiation. Quantum theory has been successful in correlating experimental data on the interaction of radiation with atoms, the photoelectric effect, and the production of x-rays. Wave theory is more useful for considering radiation in bulk when millions of quanta are being examined, as in experiments dealing with refraction, reflection, diffraction, interference, and polarization. Considering the value of both theories to understand the properties of electromagnetic radiation energy, wavelength, and frequency are all used to describe these radiations. In practical use, high-energy photons such as x-rays and γ-rays are typically characterized by their energy (eVs), medium-energy photons (e.g., visible light and ultraviolet waves) are typically characterized by their wavelength (nanometers), and low-energy photons (e.g., AM and FM radio waves) are typically characterized by their frequency (KHz and MHz).

Box 1.1 shows the relationships between photon energy, wavelength, and frequency.

Box 1.1

Relationship Between Energy *(E)* and Wavelength (λ) of Electromagnetic Radiation

$E = h \times \frac{c}{\lambda}$ simplified as $E = \frac{1.24}{\lambda}$ $E \propto \frac{1}{\lambda}$	*E* is energy (kiloelectron volts, keV) *h* is the Planck constant (6.626×10^{-34} joule-seconds or 4.13×10^{-15} eV-s) *c* is the velocity of light = 3×10^{8} m/s λ is wavelength (nanometers, nm)

Key point: **Inverse relationship between energy and wavelength of an electromagnetic radiation**

Particulate Radiation

Small atoms have approximately equal numbers of protons and neutrons, whereas larger atoms tend to have more neutrons than protons. Larger atoms are unstable because of the unequal distribution of protons and neutrons, and they may break up, releasing α (alpha) or β (beta) particles or γ (gamma) rays. This process is called **radioactivity**. When a radioactive atom releases an α or a β particle, the atom is transmuted into another element. Another type of radioactivity is γ decay, producing γ-rays. They result as part of a decay chain where a nucleus converts from an excited state to a lower level ground state; this often happens after a nucleus emits an α or β particle or after nuclear fission or fusion.

Examples of radioactive decay that are important in healthcare are listed.

- An unstable atom with an excess of protons may decay by converting a proton into a neutron, a β^{+} particle (positron), and a neutrino. Positrons quickly annihilate with electrons to form two γ-rays. This reaction is the basis for PET imaging (see Chapter

13).

- An unstable atom with an excess of neutrons may decay by converting a neutron into a proton, a β^- particle, and a neutrino. β^- particles are identical to electrons. High-speed β^- particles are able to penetrate up to 1.5 cm in tissue. β^- particles from radioactive iodine-131 are used for treatment of some thyroid cancers.
- α particles are helium nuclei consisting of two protons and two neutrons. They result from the radioactive decay of many large atomic number elements. Because of their double positive charge and heavy mass, α particles densely ionize matter through which they pass and penetrate only a few micrometers of body tissues. This limited range has prompted use of alpha emitters such as radium-223 in targeted radiation therapy for bone metastasis.

The capacity of particulate radiation to ionize atoms depends on its mass, velocity, and charge. The rate of loss of energy from a particle as it moves along its track through matter (tissue) is its **linear energy transfer** (LET). The greater the physical size of the particle, the higher its charge, and the lower its velocity, the greater its LET. For example, α particles, with their high mass compared with an electron, high charge, and low velocity, are densely ionizing, lose their kinetic energy rapidly, and have a high LET. β^- particles are much less densely ionizing because of their lighter mass and lower charge; they have a lower LET. High LET radiations concentrate their ionization along a short path, whereas low LET radiations produce ion pairs much more sparsely over a longer path length.

X-Ray Machine

X-ray machines produce x-rays that pass through a patient's tissues and strike a digital receptor or film to make a radiographic image. The primary components of an x-ray machine are the x-ray tube and its power supply, positioned within the tube head. For intraoral x-ray units, the tube head is typically supported by an arm that is usually mounted on a wall (Fig. 1.4). A control panel allows the operator to adjust the duration of the exposure, and often the energy and exposure rate, of the x-ray beam. An electrical insulating material, usually oil, surrounds the tube and transformers. Often, the tube is recessed within the tube head to increase the source-to-object distance and minimize distortion (Fig. 1.5; also see Chapter 6).

FIG. 1.4 Example of an intraoral wall-mounted x-ray unit, the Planmeca ProX. (Courtesy Planmeca USA, Inc. Roselle, Illinois.)

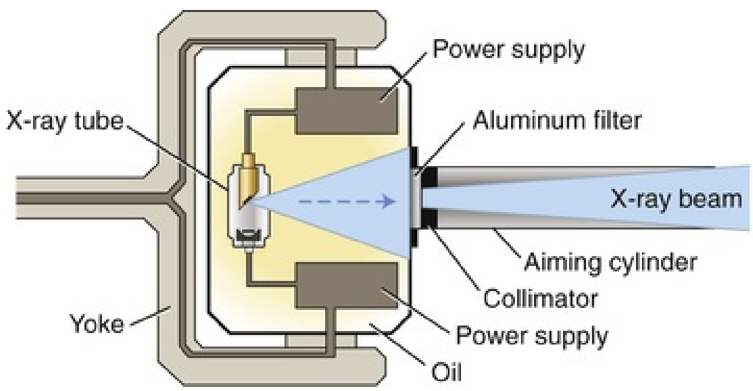

FIG. 1.5 Tube head showing a recessed x-ray tube, components of the power supply, and oil that conducts heat away from the x-ray tube. Path of useful x-ray beam *(blue)* from the anode, through the glass wall of the x-ray tube, oil, and finally an aluminum filter. The beam size is restricted by the metal tube housing and collimator. Low-energy photons are preferentially removed by the aluminum filter.

X-Ray Tube

An x-ray tube is composed of a cathode and an anode situated within an evacuated glass envelope or tube (Fig. 1.6). To produce x-rays, electrons stream from the filament in the cathode to the target in the anode, where the energy from some of the electrons is converted into x-rays.

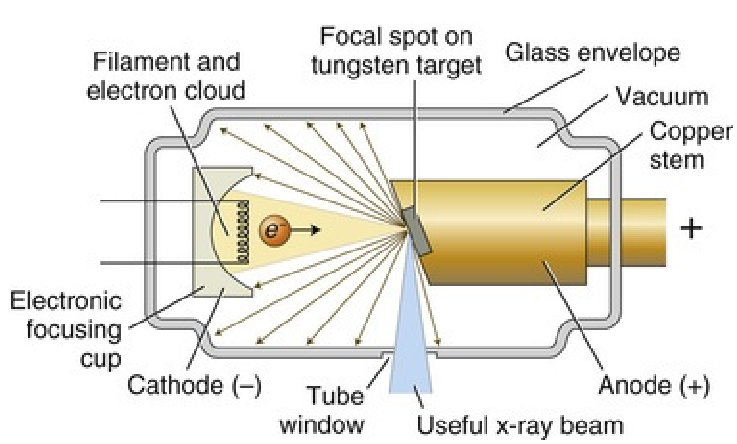

FIG. 1.6 X-ray tube with the major components labeled. The path of the electron beam is shown in *yellow*. X-rays produced at the target travel in all directions. The useful x-ray beam is shown in *blue*.

Cathode

The cathode (Figs. 1.7B and 1.8) in an x-ray tube consists of a filament and a focusing cup. The **filament** is the source of electrons within the x-ray tube. It is a coil of tungsten wire approximately 2 mm in diameter and 1 cm or less in length. Filaments typically contain approximately 1% thorium, which greatly increases the release of electrons from the heated wire. The filament is heated to incandescence with a low-voltage source and emits electrons at a rate proportional to the temperature of the filament.

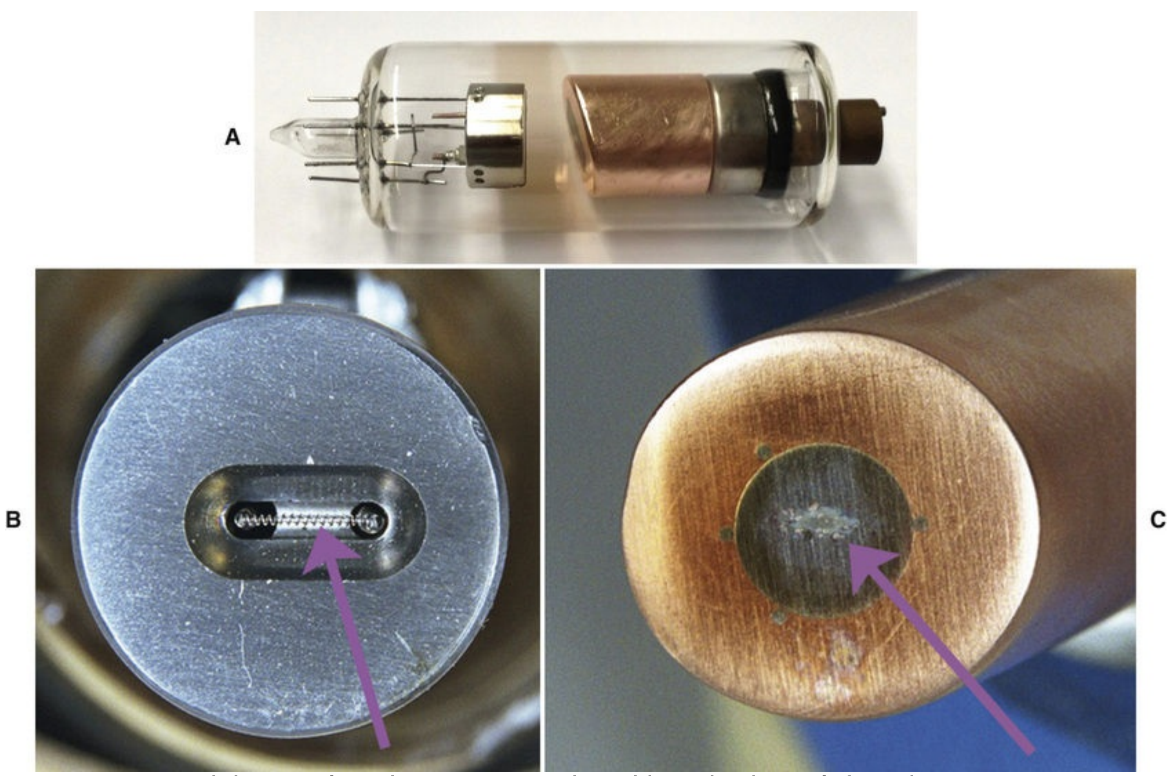

FIG. 1.7 (A) Dental stationary x-ray tube with cathode on left and copper anode on right. (B) Focusing cup containing a filament *(arrow)* in the cathode. (C) Copper anode with tungsten inset. Note the elongated actual focal spot area *(arrow)* on the tungsten target of the anode. ([B] and [C], Courtesy John DeArmond, Tellico Plains, Tennessee.)

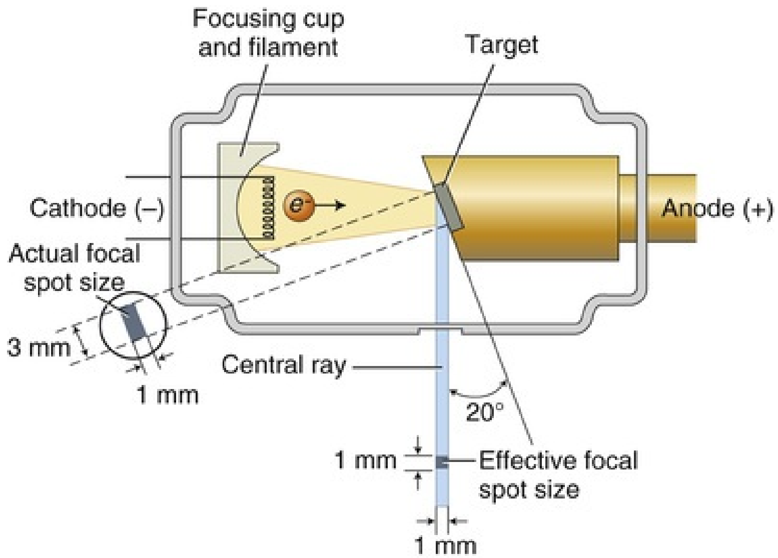

FIG. 1.8 The angle of the target to the central ray of the x-ray beam has a strong influence on the apparent size of the focal spot. The projected effective focal spot (seen below the target) is much smaller than the actual focal spot size (projected to the left). This provides a beam that has a small effective focal spot size to produce images with high resolution, while allowing for heat generated at the anode to be dissipated over the larger area.

The filament lies in a **focusing cup** (see Fig. 1.7B; see also Fig. 1.8), a negatively charged concave molybdenum bowl. The parabolic shape of the focusing cup electrostatically focuses the electrons emitted by the filament into a narrow beam directed at a small rectangular area on the anode called the **focal spot** (see Figs. 1.7C and 1.8). The electrons move to the focal spot because they are both repelled by the negatively charged cathode and attracted to the positively charged anode. The x-ray tube is evacuated to prevent collision of the fast-moving electrons with gas molecules, which would significantly reduce their speed. The vacuum also prevents oxidation, or “burnout,” of the filament.

Anode

The anode in an x-ray tube consists of a tungsten target embedded in a copper stem (see Figs. 1.6 and 1.7C). The purpose of the **target** in an x-ray tube is to convert the kinetic energy of the colliding electrons into x-ray photons. The conversion of the kinetic energy of the electrons into x-ray photons is an inefficient process, with more than 99% of the electron kinetic energy converted to heat.

The target is made of tungsten, an element that has several characteristics of an ideal target material, including the following:

- **High atomic number** (74), allows for efficient x-ray production.
- **High melting point** (3422°C), to withstand heat produced during x-ray production.
- **High thermal conductivity** (173 W m^{-1} K^{-1}), to dissipate the heat produced away from the target.
- **Low vapor pressure** at the working temperatures of an x-ray tube, to help maintain vacuum in the tube at high operating temperatures.

The tungsten target is typically embedded in a large block of copper which functions as a **thermal conductor** to remove heat from the tungsten, reducing the risk of the target melting.

The **focal spot** is the area on the target to which the focusing cup directs the electrons and from which x-rays are produced. The size of the focal spot is an important technical parameter of image quality—a smaller focal spot yields a sharper image (see Chapter 6). A limitation to reducing focal spot size is the heat generated. To overcome this limitation, x-ray tubes use one of the two anode configurations.

Stationary anode: In this configuration, the target is placed at an angle to the electron beam (see Fig. 1.8). Typically, the target is inclined approximately 20 degrees to the central ray of the x-ray beam. When viewed through the aiming ring, the area from which the photons of the useful x-ray beam originate appears smaller, making the **effective focal spot** smaller than the actual focal spot size. This allows production of x-rays from a larger area, allowing better heat distribution while maintaining the image quality benefits of a small focal spot. In the example shown in Fig. 1.8, the effective focal spot is approximately 1 mm × 1 mm, as opposed to the actual focal spot, which is approximately 1 mm × 3 mm. This smaller effective focal spot results in a small apparent source of x-rays and an increase in the sharpness of the image (see Figs. 6.1 and 6.2), with a larger actual focal spot size to improve heat dissipation.

Rotating anode: In this design, the tungsten target is in the form of a beveled disk that rotates during the period of x-ray production (Fig. 1.9). As a result, the electrons strike successive areas of the target disk, distributing the heat over this extended area of the disk. However, at any given time, x-rays are produced from a small spot on the target. X-ray tubes with rotating anode can be used with longer exposures and with higher tube currents of 100 to 500 milliamperes (mA), which is 10 to 50 times that possible with stationary targets. The target and rotor (armature) of the motor lie within the x-ray tube, and the stator coils (which drive the rotor at approximately 3000 revolutions per minute) lie outside the tube. Such rotating anodes are not used in intraoral dental x-ray machines but are occasionally used in cephalometric units; are usually used in cone beam machines; and are always used in multidetector computed tomography x-ray machines, which require high radiation output for longer, sustained exposures.

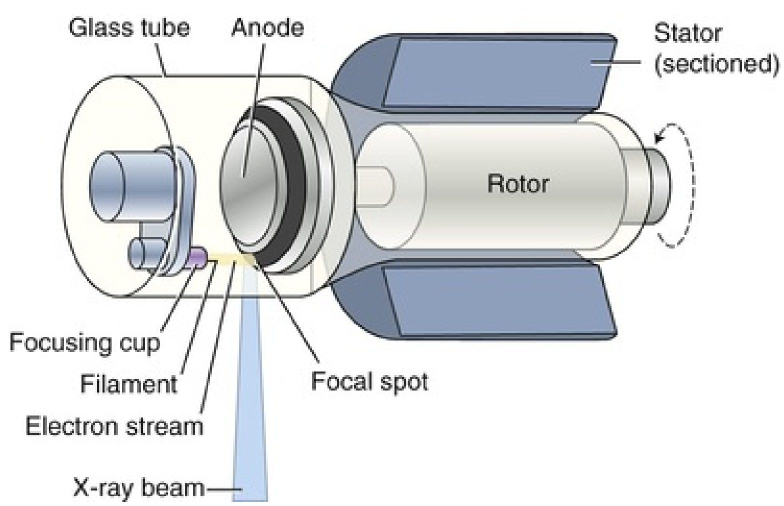

FIG. 1.9 X-ray tube with a rotating anode allows heat at the focal spot to spread out over a large surface area *(dark band).* Current applied to the stator induces rapid rotation of the rotor and the anode. The path of the electron beam is shown in *yellow*, and the useful x-ray beam is shown in *blue*.

Power Supply

The x-ray tube and two transformers lie within an electrically grounded metal housing called the **head** of the x-ray machine. The primary functions of the power supply transformers of an x-ray machine are to:

• Provide a low-voltage current to heat the x-ray tube filament (Fig. 1.10, filament transformer).

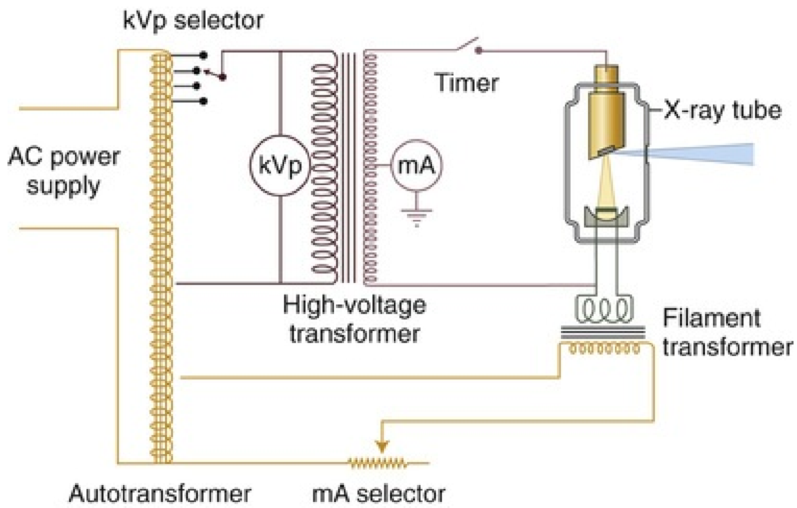

FIG. 1.10 Schematic of dental x-ray machine circuitry and x-ray tube with the major components labeled. The operator selects the desired kVp from the autotransformer. The voltage is greatly increased by the high-voltage step-up transformer and applied to the x-ray tube. The kVp dial measures the voltage on the low-voltage side of the transformer but is scaled to display the corresponding voltage in the tube circuit. The timer closes the tube circuit for the desired exposure time interval. The mA dial measures the current flowing through the tube circuit. The filament circuit heats the cathode filament and is regulated by the mA selector. *AC*, Alternate current.

• Generate a high potential difference to accelerate electrons from the cathode to the focal spot on the anode (see Fig. 1.10, high-voltage transformer).

X-Ray Tube Controls

Tube Current (Milliamperes, mA)

During x-ray production, electrons produced at the filament are attracted to the anode. This flow of electrons from the cathode to the anode generates a current across the x-ray tube and is called the tube current. The magnitude of this current is regulated by the milliampere control (see Fig. 1.10, mA selector), which adjusts the resistance and the current flow through the filament, thereby regulating the number of electrons produced. For many intraoral dental x-ray units, the mA setting is fixed, typically at 7 to 10 mA. Some units offer the flexibility of a selection of mA settings, ranging from 2 to 10 mA.

Tube Voltage (Kilovoltage, kV)

A high voltage is required between the anode and cathode to give electrons sufficient energy to generate x-rays. The kilovolt peak (kVp) selector adjusts the high-voltage transformer to boost the peak voltage of the incoming line current (110 or 220 V). Typically, intraoral, panoramic, and cephalometric machines operate between 50 and 90 kVp (50,000 to 90,000 V), whereas computed tomographic machines operate at 90 to 120 kVp, and higher.

Alternating Current X-ray Generators: For an incoming line with alternating current (AC), the polarity of the line current alternates (60 cycles per second in North America; Fig. 1.11A), and the polarity of the x-ray tube alternates at the same frequency (see Fig. 1.11B). When the polarity of the voltage applied across the tube causes the target anode to be positive and the filament to be negative, the electrons around the filament accelerate toward the positive target, and x-rays are produced (see Fig. 1.11C). When the voltage across the cathode and anode is highest, the efficiency of x-ray production is highest, and thus the intensity of x-ray pulses peaks at the center of each cycle (see Fig. 1.11C).

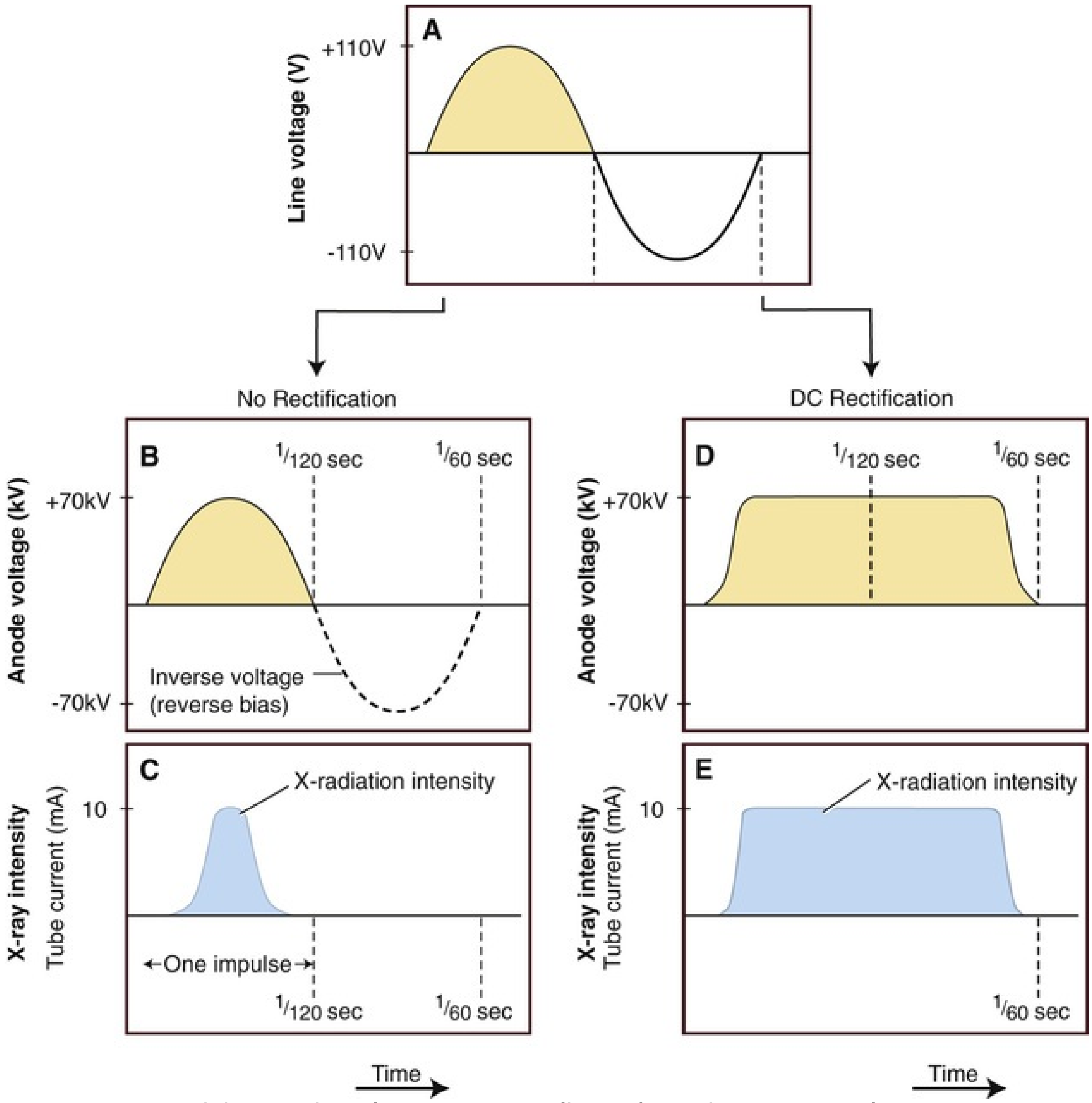

FIG. 1.11 (A) Incoming alternate current line voltage (110 V, 60 cycles per second in this case). (B) Voltage at the anode varies from zero up to the kVp setting (70 kVp in this case). (C) The intensity of radiation produced at the anode *(blue)* is strongly dependent on the anode voltage and is highest when the tube voltage is at its peak. (D) Incoming constant potential (110 V in this case) that is maintained through the operation cycle. (E) Voltage at the anode varies from zero up to the kVp setting (70 kVp in this case). Note that the increase and decrease of the potential difference at the start and end of the cycle is rapid. The intensity of radiation produced at the anode *(blue)* is higher with considerably less heterogeneity of photon energy. (Modified from Johns HE, Cunningham JR. *The Physics of Radiology.* 3rd ed. Springfield, IL: Charles C Thomas; 1974.)

During the following half (or negative half) of each cycle, the filament becomes positive, and the target becomes negative (see Fig. 1.11B). At these times, the electrons do not flow across the gap between the two elements of the tube, and no x-rays are generated. When an x-ray tube is powered with 60-cycle AC, 60 pulses of x-rays are generated each second, each having a duration of $\frac{1}{120}$ second. Thus, when using a power supply with AC, x-ray production is limited to half the AC cycle. Such x-ray units are referred to as **self-rectified** or **half-wave rectified.** Many conventional dental x-ray machines are self-rectified.

Constant Potential (Direct Current) X-ray Generators: Some dental x-ray manufacturers produce machines that replace the conventional 60-cycle AC, half-wave rectified power supply with a high-frequency power supply that provides an almost direct current (see Fig. 1.11D). This results in an essentially constant potential between the anode and cathode (see Fig. 1.11E), and x-rays are produced through the entire cycle. This almost constant voltage yields x-rays with a narrow spectrum of energies, and the mean energy of the x-ray beam produced by these x-ray machines is higher than the mean energy from a conventional half-wave rectified machine operated at the same voltage.

Practical implications with the use of constant potential intraoral x-ray units are as follows:

- Because x-ray production occurs during the entire voltage cycle, constant potential units require shorter exposure times to produce the same number of x-ray photons, minimizing patient motion.
- The intensity of x-ray photons produced is more consistent and reliable, especially with short exposure times. This is of practical importance when using digital receptors that require less radiation.
- When operated at the same kVp, the x-ray beam produced by constant potential units has a higher mean energy, which decreases radiographic image contrast. To offset this effect, constant potential x-ray units are typically operated at a slightly lower kVp,

typically 60 to 65 kVp.

• The narrower spectrum of energies, with fewer lower-energy photons, lowers the patient radiation dose by 35% to 40%, compared with conventional AC x-ray generators.

Timer

A timer is built into the high-voltage circuit to control the duration of the x-ray exposure (see Fig. 1.10). The electronic timer controls the length of time that high voltage is applied to the tube and thus the time during which x-rays are produced. However, before the high voltage is applied across the tube, the filament must be brought to operating temperature to ensure an adequate rate of electron emission. Subjecting the filament to continuous heating at normal operating current shortens its life. To minimize filament damage, the timing circuit first sends a current through the filament for approximately half a second to bring it to the proper operating temperature and then applies power to the high-voltage circuit. In some circuit designs, a continuous low-level current passing through the filament maintains it at a safe low temperature, further shortening the delay to preheat the filament. For these reasons, an x-ray machine may be left on continuously during working hours.

Some x-ray machine timers display the exposure time in fractions of a second. In some intraoral units, the exposure times are preset for different anatomic areas of the jaws. In some units, the exposure time is expressed as number of pulses in an exposure (e.g., 3, 6, 9, 15). The number of pulses divided by 60 (the frequency of the power source) gives the exposure time in seconds. A setting of 30 pulses means that there will be 30 pulses of radiation, equivalent to a 0.5-second exposure (Box 1.2).

Box 1.2

Practical Applications of Exposure Controls

In many intraoral x-ray units, the mA setting,

kVp setting, or both is fixed. If the mA setting is variable, the operator should select the highest mA value available and operate the machine at this setting; this allows the shortest exposure time and minimizes the chance of patient movement.

If tube voltage can be adjusted on an intraoral radiographic unit, the operator may choose to operate at a fixed voltage, typically 65–70 kVp. This protocol simplifies selecting the proper patient exposure settings by using just exposure time as the means to adjust for anatomic location within the mouth and patient size.

The kVp setting is often used to compensate for patient tissue thickness, particularly for panoramic and cephalometric radiography. A rule of thumb is to vary the setting by 2 kVp/cm of tissue thickness.

Tube Rating and Duty Cycle

X-ray tubes produce heat at the target while in operation. The heat buildup at the anode is measured in heat units (HU), where HU = kVp × mA × seconds. The heat storage capacity for anodes of dental diagnostic tubes is approximately 20 kHU. Heat is removed from the target by conduction to the copper anode and then to the surrounding oil and tube housing and by convection to the atmosphere.

Each x-ray machine comes with a **tube rating** chart that describes the longest exposure time the tube can be energized for a range of voltages (kVp) and tube current (mA) values without risk of damage to the target from overheating. These tube ratings generally do not restrict tube use for intraoral radiography. **Duty cycle** relates to the frequency with which successive exposures can be made without overheating the anode. The interval between successive exposures must be long enough for heat dissipation. This characteristic is a function of the size of the anode, the exposure kVp and mA, and the method used to cool the tube. A duty cycle of 1 : 60 indicates that one could make a 1-second exposure every 60 seconds.

Production of X Rays

Most high-speed electrons traveling from the filament to the target interact with target electrons and release their energy as heat. Occasionally, the electron's kinetic energy is converted into x-ray photons by the formation of **bremsstrahlung radiation** and **characteristic radiation.**

Bremsstrahlung Radiation

Bremsstrahlung photons are the primary source of radiation from an x-ray tube. *Bremsstrahlung* means “braking radiation” in German, and these photons are produced by the sudden stopping or slowing of high-speed electrons by tungsten nuclei in the target as follows: Most high-speed electrons pass by tungsten nuclei with near or wide misses (Fig. 1.12A). In these interactions, the electron is attracted toward the positively charged nuclei, its path is altered toward the nucleus, and it loses some of its velocity. This deceleration causes the electron to lose kinetic energy that is given off in the form of x-ray photons. The closer the high-speed electron approaches the nuclei, the greater the electrostatic attraction between the nucleus and the electron, and the resulting bremsstrahlung photons have higher energy. The efficiency of this process is proportional to the square of the atomic number of the target; high *Z* metals are more effective in deflecting the path of the incident electrons, and this is the basis for selection of tungsten (*Z* = 74) as a target material.

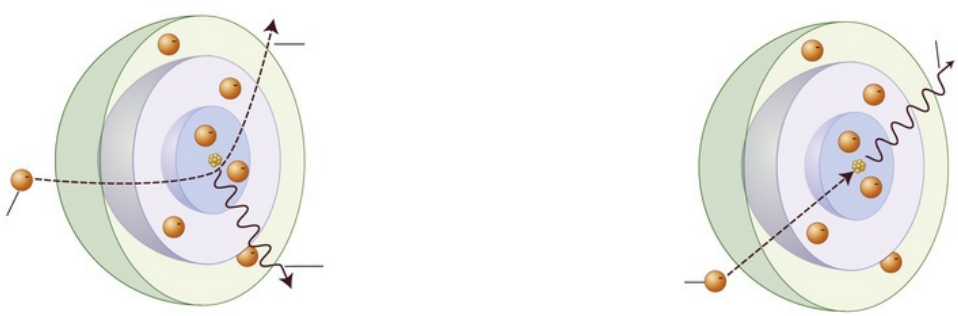

FIG. 1.12 Bremsstrahlung radiation is produced most often by the passage of an electron near a nucleus, which results in electrons being deflected and decelerated (A) or, less frequently, by the direct hit of an electron on a nucleus in the target (B). For the sake of clarity, this diagram

and other similar figures in this chapter show only the 1s, 2s, or 3s orbitals.

Occasionally, electrons from the filament directly hit the nucleus of a target atom. When this happens, all the kinetic energy of the electron is transformed into a single x-ray photon (see Fig. 1.12B). The energy of the resultant photon (in keV) is numerically equal to the energy of the electron (i.e., the voltage applied across the x-ray tube at that instant).

Bremsstrahlung interactions generate x-ray photons with a continuous spectrum of energy. The energy of an x-ray beam is usually described by identifying the peak operating voltage (in kVp). For example, a dental x-ray machine operating at a peak voltage of 70 kVp applies a voltage of up to 70 kVp across the tube. This tube therefore produces a continuous spectrum of x-ray photons with energies ranging to a maximum of 70 keV (Fig. 1.13). The reasons for this continuous spectrum are as follows:

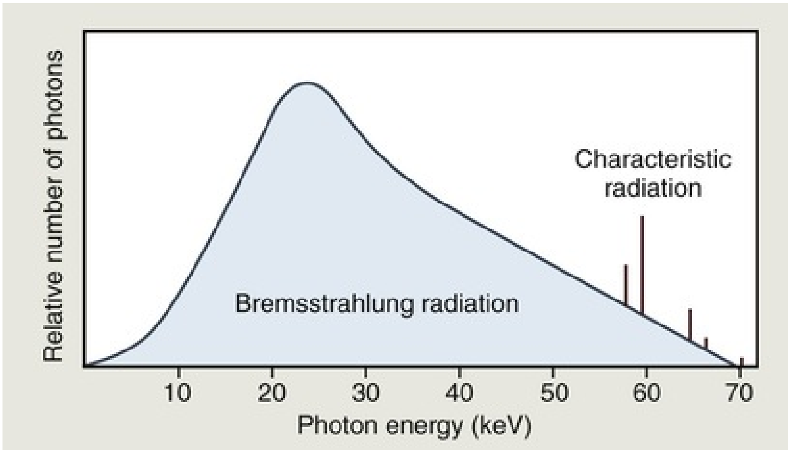

FIG. 1.13 Spectrum of photons emitted from an x-ray machine operating at 70 kVp. The vast preponderance of radiation is bremsstrahlung *(blue)*, with a minor addition of characteristic radiation.

- The continuously varying voltage difference between the target and filament causes the electrons striking the target to have varying levels of kinetic energy.
- The bombarding electrons pass at varying distances

around tungsten nuclei and are thus deflected to varying extents. As a result, they give up varying amounts of energy in the form of bremsstrahlung photons.

• Most electrons participate in multiple bremsstrahlung interactions in the target before losing all their kinetic energy. Consequently, an electron carries differing amounts of energy after successive interactions with tungsten nuclei.

Characteristic Radiation

Characteristic radiation contributes only a small fraction of the photons in an x-ray beam. It is made when an incident electron ejects an inner electron from the tungsten atom. When this happens, an electron from an outer orbital is quickly attracted to the void in the deficient inner orbital (Fig. 1.14). When the outer orbital electron replaces the displaced electron, a photon is emitted with energy equivalent to the difference in the binding energies of the two orbitals. The energies of characteristic photons are discrete because they represent the difference of the energy levels of specific electron orbitals and are characteristic of the target atoms. The production of characteristic radiation has no practical implications for dentomaxillofacial radiography.

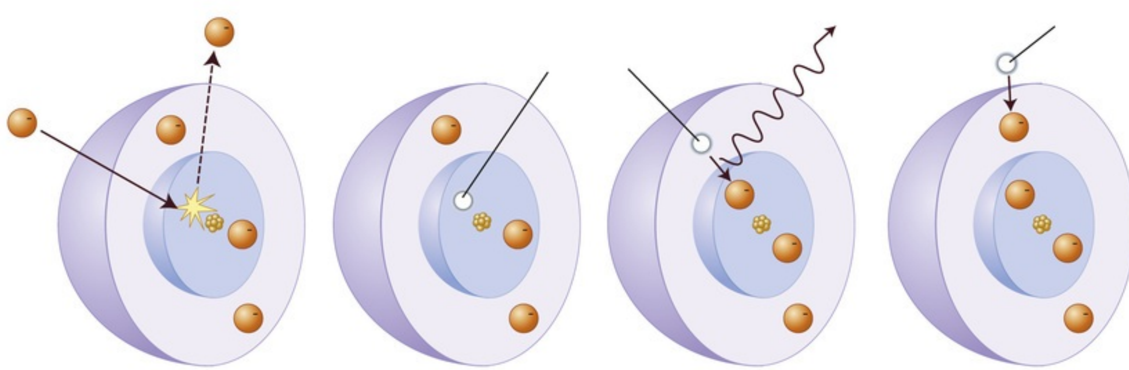

FIG. 1.14 Production of characteristic radiation. An incident electron (A) ejects an electron from an inner orbital creating an electron vacancy (B). (C) An electron from an outer orbital fills this vacancy, and a photon is emitted with energy equal to the difference in energy levels between the two orbitals. (D) Electrons from various orbitals may be involved, giving rise to other characteristic photons. The energies of the photons released are characteristic of the energy transitions for the target atom.

Factors Controlling the X-Ray Beam

An x-ray beam may be modified by altering the beam exposure duration (timer), exposure rate (mA), energy (kVp and filtration), shape (collimation), or intensity (target-patient distance).

Exposure Time (s)

Changing the exposure time—typically measured in fractions of a second—modifies the duration of the exposure and thus the number of photons generated (Fig. 1.15). When the exposure time is doubled, the number of photons generated at all energies in the x-ray emission spectrum is doubled. The range of photon energies is unchanged. Practically, it is desirable to keep the exposure time as short as possible to minimize blurring from patient motion.

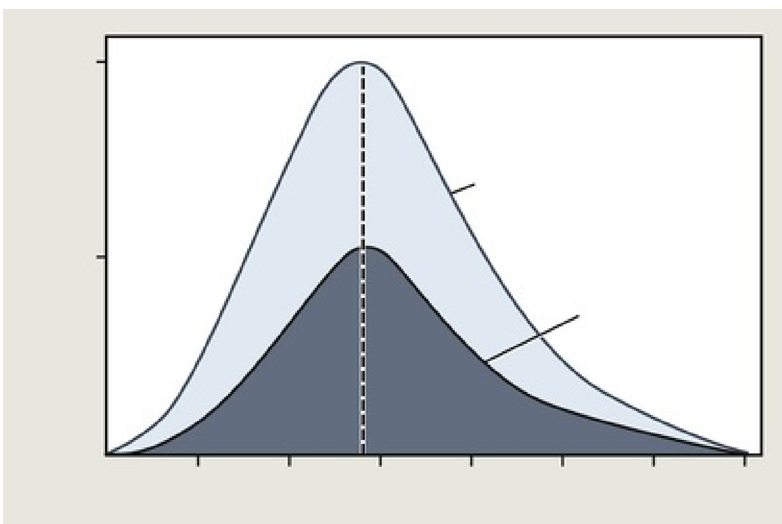

FIG. 1.15 Spectrum of photon energies generated in an x-ray machine showing that as exposure time increases (kVp and mA settings held constant), so does the total number of photons. The mean energies (*dotted line*, approximately 29 keV in this example) and maximal energies (70 keV in this example) of the beams are unchanged.

Milliamperage Setting (mA, Tube Current)

Like the effects of exposure time, the quantity of radiation produced by an x-ray tube (i.e., the number of photons that reach the patient) is directly proportional to the milliamperage setting (mA setting; Fig. 1.16). As the mA setting is increased, more power is applied to the filament, which heats up and releases more electrons that collide with the target to produce radiation. Thus, as with exposure time, doubling the mA setting will double the number of photons produced. The product of mA setting and exposure time (**mA × s**, or **mAs**) is often used as a single parameter to denote the total number of photons produced. For instance, a machine operating at 10 mA for 1 second (**10 × 1 = 10 mAs**) produces the same number of photons when operated at 20 mA for 0.5 second (**20 × 0.5 = 10 mAs**). The term **beam quantity** refers to the number of photons in an x-ray beam. Linearity and reproducibility of the mA and s settings are often included in the quality assurance programs for x-ray units, including those used in dental and maxillofacial imaging.

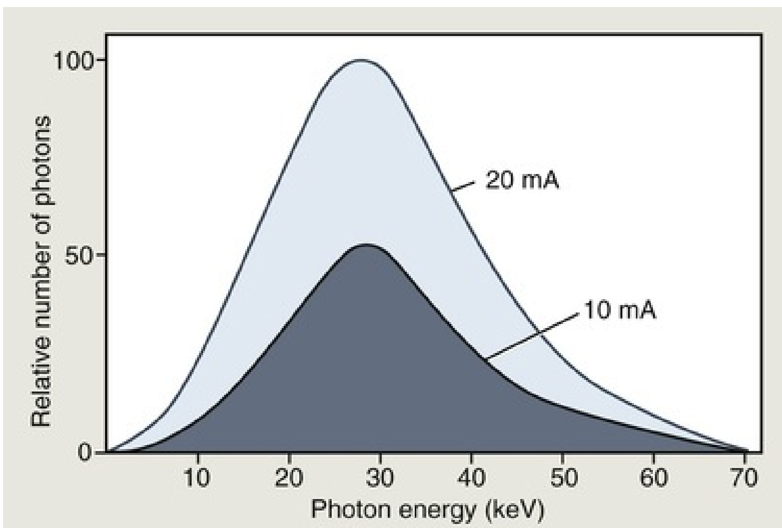

FIG. 1.16 Spectrum of photon energies generated in an x-ray machine showing that as the mA setting increases (kVp and exposure time held constant), so does the total number of photons. The mean energies and maximal energies of the beams are unchanged. Note similarity to the effect of exposure time; see Fig. 1.15.

Tube Voltage Peak (kVp)

Increasing the kVp increases the potential difference between the cathode and the anode, increasing the kinetic energy of the electrons as they move toward the target. The greater the energy of an electron, the greater the probability it will be converted into x-ray photons at the target. Increasing the kVp of an x-ray machine increases:

- The number of photons generated.
- The mean energy of the photons.
- The maximal energy of the photons (Fig. 1.17).

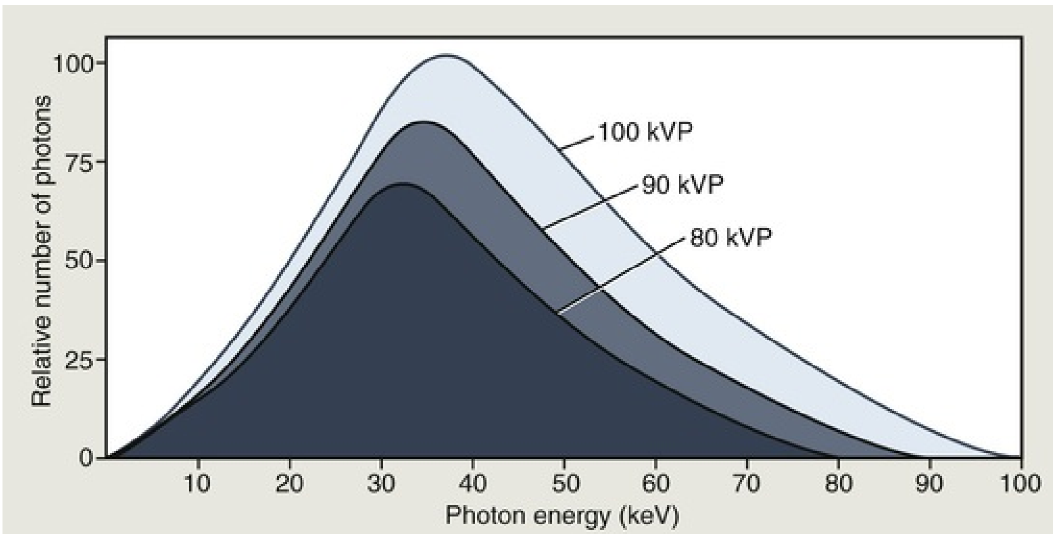

FIG. 1.17 Spectrum of photon energies generated in an x-ray machine showing that as the kVp is increased (mA and s held constant), there is a corresponding increase in the mean energy of the beam, the total number of photons emitted, and the maximal energy of the photons. Compare with Figs. 1.15 and 1.16.

The term **beam quality** refers to the mean energy of an x-ray beam.

Filtration

Although an x-ray beam consists of a continuous spectrum of x-ray photon energies, only photons with sufficient energy to penetrate through anatomic structures and reach the image receptor (digital or film) are useful for diagnostic radiology. Low-energy photons that cannot reach the receptor contribute to patient risk but do not offer any benefit. Consequently, it is desirable to remove these low-energy photons from the beam. This removal can be accomplished in part by placing a metallic disk (filter) in the beam path. A filter preferentially removes low-energy photons from the beam but allows high-energy photons that contribute to making an image to pass through (Fig. 1.18).

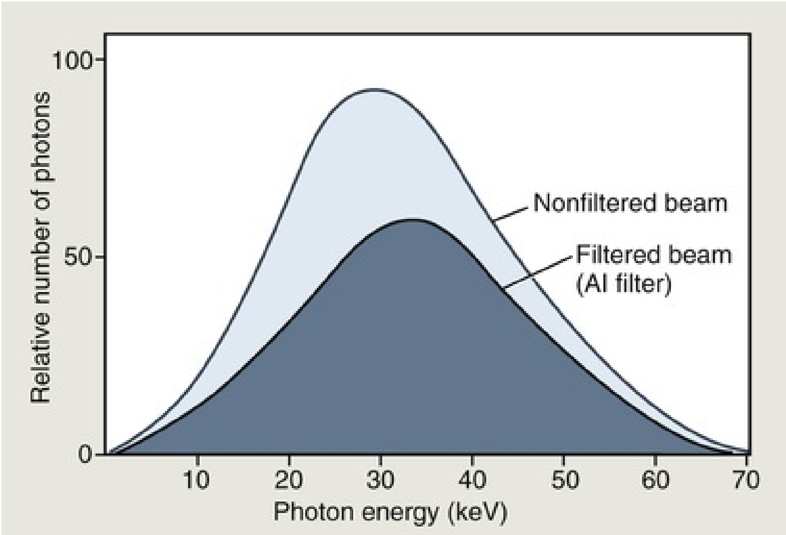

FIG. 1.18 Spectrum of filtered x-ray beam generated in an x-ray machine showing that an aluminum filter preferentially removes low-energy photons, reducing the beam intensity, while increasing the mean energy of the residual beam. Compare with Figs. 1.15–1.17.

Inherent filtration consists of the materials that x-ray photons encounter as they travel from the focal spot on the target to form the usable beam outside the tube enclosure. These materials include the glass wall of the x-ray tube, the insulating oil that surrounds many dental tubes, and the barrier material that prevents the oil from escaping through the x-ray port. The inherent filtration of most x-ray machines ranges from the equivalent of 0.5 to 2 mm of aluminum.

Added filtration may be supplied in the form of aluminum disks placed over the port in the head of the x-ray machine. **Total filtration** is the sum of the inherent and added filtration. Federal regulations in the United States require the total filtration in the path of a dental x-ray beam to be equal to the equivalent of 1.5 mm of aluminum for a machine operating at up to 70 kVp and 2.5 mm of aluminum for machines operating at higher voltages (see Chapter 3).

Collimation

A collimator is a metallic barrier with an aperture in the middle used to shape and restrict the size of the x-ray beam and the volume of tissue irradiated (Fig. 1.19). Round and rectangular collimators are most frequently used in intraoral radiography. Dental x-ray beams are usually collimated to a circle 2.75 inches (7 cm) in diameter at the patient's face. A round collimator (see Fig. 1.19A) is a thick plate of metal with a circular opening centered over the port in the x-ray head through which the x-ray beam emerges. Typically, round collimators are built into open-ended aiming cylinders. Rectangular collimators (see Fig. 1.19B) further limit the size of the beam to just larger than the intraoral receptor, further reducing patient exposure. Some types of receptor-holding instruments also provide rectangular collimation of the x-ray beam (see Chapters 3 and 7).

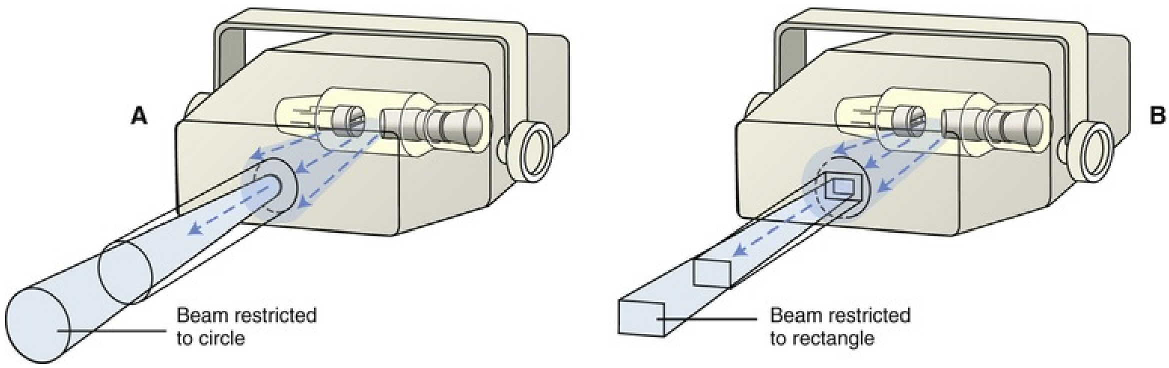

FIG. 1.19 Collimation of an x-ray beam *(blue)* is achieved by restricting its useful size. (A) Circular collimator. (B) Rectangular collimator restricts area of exposure to just larger than the detector size and thereby reduces unnecessary patient exposure.

Collimators also improve image quality. When an x-ray beam is directed at a patient, the hard and soft tissues absorb approximately 90% of the photons and approximately 10% pass through the patient to reach the image receptor (film, or digital receptor). Many of the absorbed photons generate scattered radiation within the exposed tissues by a process called **Compton scattering** (see later in chapter). These scattered photons travel in all directions, and some reach the receptor and degrade image quality. Collimating the x-ray beam thus reduces the exposed volume and thereby the number of scattered photons reaching the image receptor, resulting in reduced patient exposure and improved images.

Inverse Square Law

The intensity of an x-ray beam (the number of photons per cross-sectional area per unit of exposure time) varies with distance from the focal spot. For a given beam, the intensity is inversely proportional to the square of the distance from the source (Fig. 1.20). The reason for this decrease in intensity is that an x-ray beam spreads out as it moves from its source. The relationship is as follows:

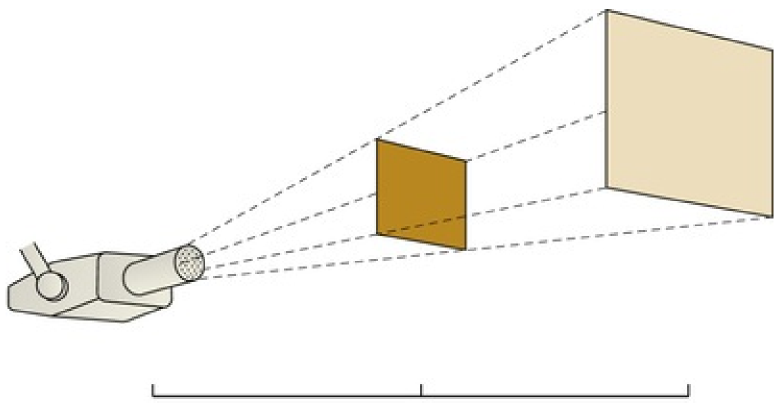

FIG. 1.20 Intensity of an x-ray beam is inversely proportional to the square of the distance between the source and the point of measure. When the distance from the focal spot is doubled, the intensity of the beam decreases to one quarter.

$$\frac{I_1}{I_2} = \frac{(D_2)^2}{(D_1)^2}$$

where I is intensity and D is distance. If a dose of 4 Gy is measured at 1 m, a dose of 1 Gy would be found at 2 m and a dose of 0.25 Gy would be found at 4 m.

Practical Applications

- Changing the distance between the x-ray tube and the patient, such as by switching from a machine with a short aiming tube to one with a long aiming tube, has a marked effect on beam intensity. Such a change requires a corresponding modification of the kVp or mA to maintain the same intensity at the image receptor.

- Increasing operator distance from the x-ray source is an effective method to minimize operator dose (see Chapter 3).

Interactions of X Rays With Matter

In dental and maxillofacial imaging, the x-ray beam enters the face of a patient, interacts with hard and soft tissues, and strikes a digital sensor or film. The incident beam contains photons of many energies but is spatially homogeneous. That is, the intensity of the beam is essentially uniform from the center of the beam outward. As the beam goes through the patient, it is reduced in intensity (attenuated). This **attenuation** results from absorption of individual photons in the beam by atoms in the tissues or by photons being scattered out of the beam. In **absorption interactions,** photons interact with tissue atoms and cease to exist. In **scattering interactions,** photons also interact with tissue atoms but then move off in another direction. The frequency of these interactions depends on the type of tissue exposed (e.g., bone vs. soft tissue). Bone is more likely to absorb x-ray photons, whereas soft tissues are more likely to let them pass through. Although the incident beam striking the patient is spatially homogeneous, the remnant beam—the attenuated beam that exits the patient—is spatially heterogeneous because of differential absorption by the anatomic structures through which it has passed. This differential exposure of the film or digital sensor forms a radiographic image.

There are three means of beam attenuation in a diagnostic x-ray beam (Table 1.2):

TABLE 1.2

Interactions of Photons From a Diagnostic X-Ray Beam

Interaction	Ionization	Scatter	Practical Implications
Photoelectric absorption	Yes	No	Basis of radiographic image formation
Compton scatter	Yes	Yes	Scatter radiation can degrade image, expose personnel and patient
Coherent scatter	No	No	Minimal contribution to scatter

- Photoelectric absorption
- Compton scattering
- Coherent scattering

In addition, approximately 9% of the primary photons pass through the patient's tissues without interaction and strike the sensor to form an image (Fig. 1.21 and Table 1.3).

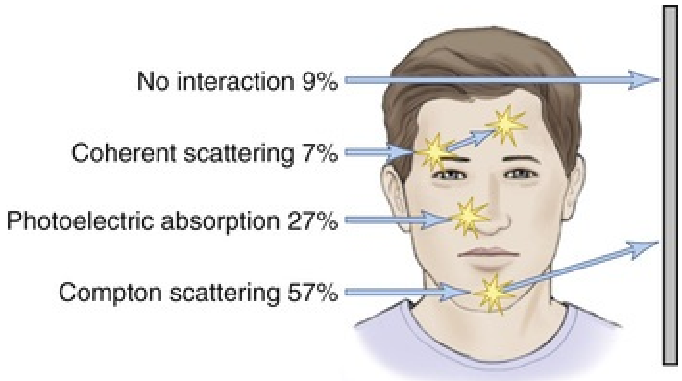

FIG. 1.21 Photons in an x-ray beam interact with the object primarily by Compton scattering (57% of primary interactions), in which case the scattered photon may strike the film and degrade the radiographic image by causing fog. The next most frequent interaction is photoelectric absorption (27%), in which the photons cease to exist. A radiographic image is produced by photons passing through low atomic number structures (soft tissue) and preferentially undergoing photoelectric absorption by high atomic number structures (bone, teeth, and metallic restorations). Relatively few photons undergo coherent scattering (7%) within the object or pass through the object without interaction (9%) and expose the image receptor.

TABLE 1.3

Fate of 1 Million Incident Photons in Bite-wing Projection

Interaction	Fate of Incident Photon	Primary Photons	Scattered Photons[a]	Total[b]
Coherent scattering	Scatters from atom	74,453	78,117	152,570
Photoelectric absorption	Ejects inner electron and ceases to exist; releases characteristic photon	268,104	261,041	529,145
Compton scattering	Ejects outer electron, both scatter	565,939	549,360	1,115,300
No interaction	Passes through patient	91,504	379,350	470,855
Total		*1,000,000*	*1,267,868*	*2,267,869*

[a]The fate of scattered photons resulting from primary Compton, photoelectric, and coherent interactions.

[b]The sum of the total number of photoelectric interactions and photons that exit the patient equals the total number of incident photons.

From SJ Gibbs, personal communication, 1986.

Photoelectric Absorption

Photoelectric absorption is critical in diagnostic imaging because it is the basis of image radiographic formation. This process occurs when an incident photon interacts with an electron in an inner orbital of an atom in the patient. The incident photon loses all its energy to the electron and ceases to exist. The energy absorbed by the electron is expended to overcome the binding energy, and the remainder energy remains as the kinetic energy of the electron as it escapes the confines of its orbital (Fig. 1.22). The kinetic energy imparted to the electron (termed **recoil electron** or **photoelectron**) is equal to the energy of the incident photon minus the binding energy of the electron. In the case of atoms with low atomic numbers (e.g., atoms in most biologic molecules), the binding energy is small and the photoelectron acquires most of the energy of the incident photon. Photoelectrons ejected during photoelectric absorption travel only short distances in the absorber before they give up their energy through secondary ionizations.

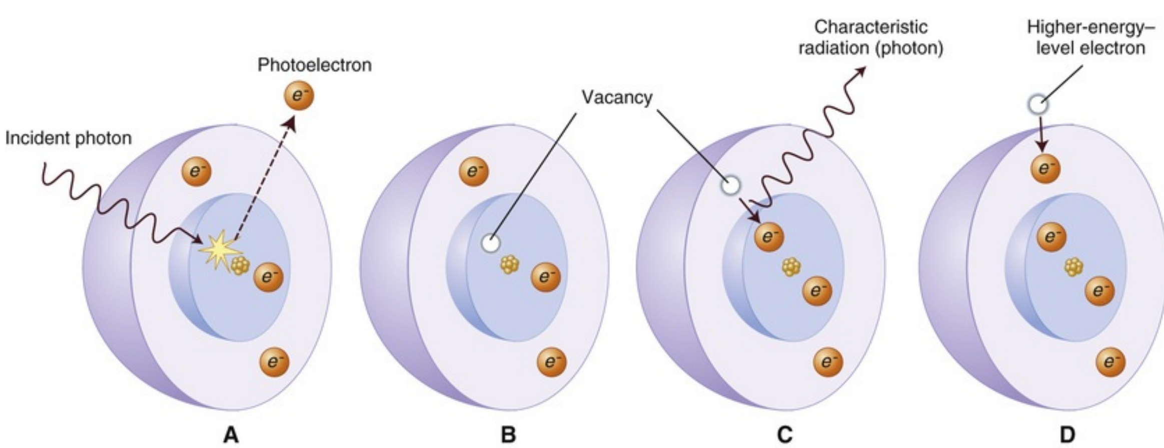

FIG. 1.22 Photoelectric absorption. (A) Photoelectric absorption occurs when an incident photon gives up all of its energy to an inner electron, which is ejected from the atom (a photoelectron). The incident electron ceases to exist at this point. (B) The ionized atom now has an electron vacancy in the inner orbital. (C) An electron from a higher energy level fills the vacancy and emits characteristic radiation. (D) All orbitals are subsequently filled, completing the energy exchange.

Most photoelectric interactions occur in the 1s orbital because the density of the electron cloud is greatest in this region, and there is a higher probability of interaction. Approximately 27% of interactions in a dental x-ray beam exposure involve photoelectric absorption.

The photoelectric interaction causes ionization of the atom because of the loss of an electron. This electron deficiency (usually in the 1s orbital) is instantly filled, usually by a 2s or 2p electron, with the release of characteristic radiation (see Fig. 1.14). Whatever the orbital of the replacement electron, the characteristic photons generated are of such low energy that they are absorbed within the patient and do not fog the receptor.

The probability of photoelectric interaction is directly proportional to the **third power of the atomic number** *(Z)* of the absorber, and inversely proportional to the **third power of the energy of the incident photon** *(E)*.

$$\text{Probablity of photoelectric interaction} \propto \frac{Z^3}{E^3}$$

The practical implications of photoelectric interaction are listed in Box 1.3.

Box 1.3

Practical Implications of Photoelectric Effect

Differential absorption in various tissues and objects (restorations for example) provides radiographic contrast. Because the effective atomic number of compact bone (*Z* = 13.8) is greater than that of soft tissue (*Z* = 7.4), the probability of photoelectric interaction of x-ray photons in bone is approximately 6.5 times greater than in an equal thickness of soft tissue ($13.8^3/7.4^3 = 6.5$). This marked difference in the absorption of x-ray photons

by the soft and hard tissues makes the production of a radiographic image possible. This differential photoelectric absorption of x-ray photons in enamel, dentin, pulp, bone, and soft tissue is what we observe as different degrees of radiopacity on the radiographic image.

Causes ionization and potential for biological damage.

Compton Scatter

Compton scatter occurs when a photon interacts with an outer orbital electron (Fig. 1.23). Approximately 57% of interactions in a dental x-ray beam exposure involve Compton scatter. In this interaction, the incident photon collides with an outer orbital electron, which receives kinetic energy and recoils from the point of impact. The path of the incident photon is deflected by this interaction and is scattered in a new direction. The energy of this scattered photon equals the energy of the incident photon minus the sum of the kinetic energy gained by the recoil electron and its binding energy. In the diagnostic energy range, most of the energy is retained by the scattered photon which can then cause additional ionizations, often at tissue sites outside the circumference of the incident beam. When these scattered photons reach the image receptor, they cause degradation of the image.

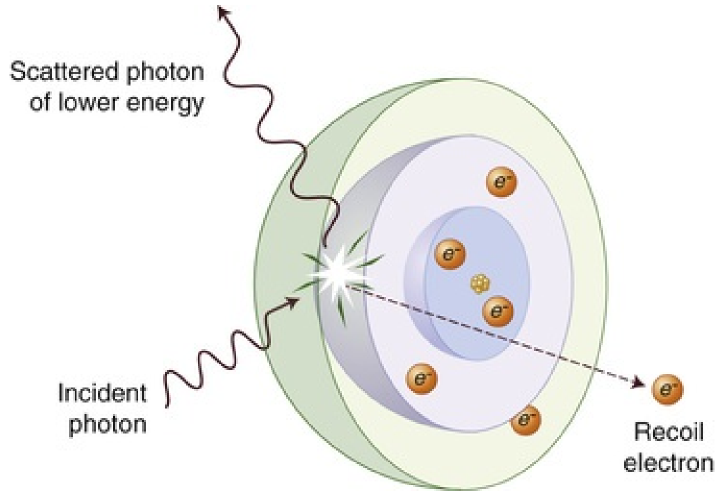

FIG. 1.23 Compton scattering occurs when an incident photon interacts with an outer electron, producing a scattered photon of lower energy than the incident photon and a recoil electron ejected from the target atom. The new scattered photon travels in a different direction from the incident photon.

As with photoelectric absorption, Compton scatter results in the loss of an electron and ionization of the absorbing atom. Additional ionizations are caused by the scattered photons and the recoil electrons as they course through the patient's tissues. The probability of a Compton interaction is inversely proportional to the photon energy and is independent of atomic number. The probability of Compton scatter is dependent on the **electron density** of the absorber, which is relatively constant in tissue.

The practical implications of Compton scatter are listed in Box 1.4.

Box 1.4

Practical Implications of Compton Scatter

Scattered photons travel in all directions and may exit the patient and strike the image receptor. These photons carry no useful information and degrade the image by reducing contrast.

Scattered photons that exit the patient can expose the operator.

Scattered photons travel varying distances within the patient's tissues and cause ionizations. This internal scatter increases patient radiation dose and often exposes organs and tissues outside of and distant from the path of the primary beam.

Coherent Scatter

Coherent scatter (also known as **Rayleigh, classical,** or **elastic scatter**) may occur when a low-energy incident photon (<10 keV) interacts with a whole atom. The incident photon causes it to become momentarily excited (Fig. 1.24). The incident photon then ceases to exist. The excited atom quickly returns to the ground state and generates another x-ray photon with the same energy as the incident photon. Usually the secondary photon is emitted in a different direction than the path of the incident photon. The net effect is that the direction of the incident x-ray photon is altered (scattered). Coherent scattering accounts for only approximately 7% of the total number of interactions in a dental exposure (see Table 1.3). Because no energy is transferred to the biologic atom and no ionizations are caused, the biologic effects of coherent scatter are insignificant. Because coherent scatter occurs primarily in the lower energy range, the scattered photon has insufficient energy to reach the image receptor, and thus coherent scatter has minimal impact on image degradation.

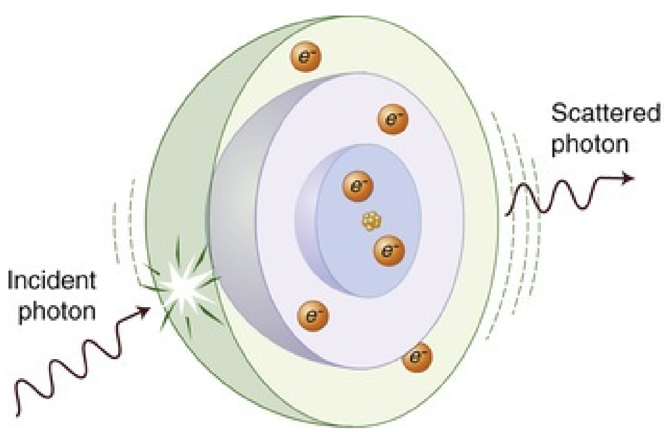

FIG. 1.24 Coherent scattering results from the interaction of a low-energy incident photon with a whole atom, causing it to be momentarily excited. After this interaction, the atom quickly returns to the ground state and emits a scattered photon of the same energy but at a different angle from the path of the incident photon.

Beam Attenuation

As an x-ray beam travels through matter, its intensity is reduced primarily through photoelectric absorption and Compton scattering. The extent of beam attenuation depends primarily on the energy of the beam and the thickness and density of the attenuating material. High-energy x-ray photons have a greater probability of penetrating matter, whereas lower-energy photons have a greater probability of being attenuated. The higher the kVp setting, the greater the penetrability of the resulting beam through matter. A useful way to characterize the penetrating quality of an x-ray beam is by its half-value layer (HVL). The HVL is the thickness of an absorber, such as aluminum, that reduces the number of x-ray photons by 50%. As the mean energy of an x-ray beam increases, so does the amount of material required to reduce the beam intensity by half (its HVL). The HVLs of several materials have been established for a wide range of photon energies. This allows medical physicists to calculate the thickness of material required and design appropriate shielding in diagnostic radiology facilities.

The reduction of beam intensity also depends on physical characteristics of the absorber. Higher-density materials attenuate more because of more photoelectric absorption and more Compton scattering with increasing density. In addition, increasing the thickness of an absorber increases the number of interactions. A monochromatic beam of photons, a beam in which all the photons have the same energy, provides a useful example. When only the primary (not scattered) photons are considered, a constant fraction of the beam is attenuated as the beam moves through each unit thickness of an absorber. For example, if 1.5 cm of water reduces a beam intensity by 50%, the next 1.5 cm reduces the beam intensity by another 50% (to 25% of the original intensity), and so on. This is an exponential pattern of absorption (Fig. 1.25). The HVL described earlier is a measure of beam energy describing the amount of an absorber that reduces the beam intensity by half; in the preceding example, the HVL is 1.5 cm of water.

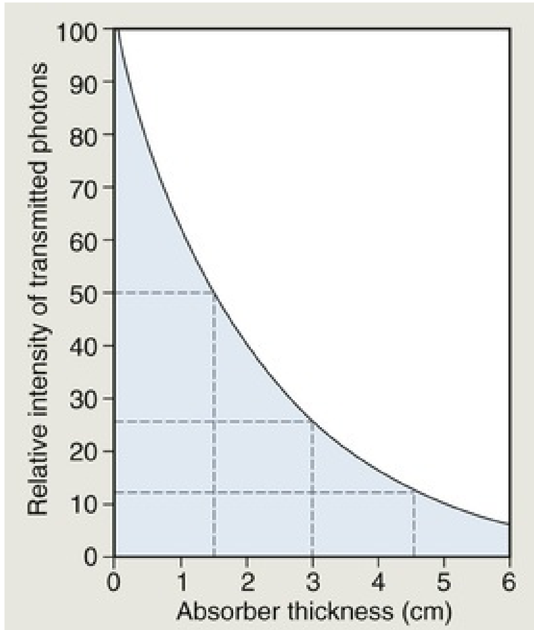

FIG. 1.25 Intensity of an energetically homogeneous x-ray beam declines exponentially as it travels through an absorber. In this instance, the half-value layer of the beam is 1.5 cm of absorber (i.e., every 1.5 cm of the absorber reduces the intensity of the beam by half). The curve for a heterogeneous x-ray beam (e.g., a dental x-ray beam) does not drop quite as precipitously because of the preferential removal of low-energy photons by the absorber and the increased mean energy of the resulting beam.

In contrast to the previous example using a monoenergetic x-ray beam, there is a wide range of photon energies in an x-ray beam. Low-energy photons are much more likely than high-energy photons to be attenuated. Thus the superficial layers of an absorber remove the low-energy photons but transmit many of the higher-energy photons. As an x-ray beam passes through this material, the intensity of the beam decreases from preferential removal of low-energy photons. Because the transmitted photons are predominantly higher energy, the mean energy of the residual beam increases. The term **beam hardening** is used to describe this increase in the mean energy of the beam by preferential removal of lower-energy photons.

As the energy of an x-ray beam increases, so does the transmission of the beam through an absorber. However, when the energy of the incident photon is increased to match the binding energy of the 1s orbital electrons of the absorber, the probability of photoelectric absorption increases sharply and the number of absorbed photons is greatly increased. This is called **K-edge absorption.** The probability that a photon will interact with an orbital electron is greatest when the energy of the photon equals the binding energy of the electron; it decreases as the photon energy increases. Photons with energy less than the binding energy of 1s orbital electrons interact by photoelectric absorption only with electrons in the 2s or 2p orbitals and in orbitals even farther from the nucleus. Rare earth elements are sometimes used as filters because their 1s orbital binding energies, or K edges (e.g., 50.24 keV for gadolinium), greatly increase the absorption of high-energy photons. This is desirable because these high-energy photons degrade image contrast and are not as likely as mid-energy photons that primarily contribute to a radiographic image.

Dosimetry

Table 1.4 presents some frequently used units of radiation and radiation detriment. Contemporary literature uses radiation measurement units from the **SI system** *(Système International d'Unitès)*, and these will be used in this book. Traditional units and their conversion have been included for reference.

TABLE 1.4

Summary of Radiation Quantities and Units

Quantity	**Description**	**SI Unit**	**Traditional Unit**	**Conversion**
Exposure	Amount of ionization of air by x- or γ-rays	coulomb/kg (C/kg)	roentgen (R)	1 C/kg = 3876 R
Kerma	Kinetic energy transferred to charged particles	gray (Gy)	—	—
Absorbed dose	Total energy absorbed by a mass	gray (Gy)	rad	1 Gy = 100 rad
Equivalent dose	Absorbed dose weighted by biologic effectiveness of radiation type used	sievert (Sv)	rem	1 Sv = 100 rem
Effective dose	Sum of equivalent doses weighted by radiosensitivity of exposed tissue or organ	sievert (Sv)	—	—
Radioactivity	Rate of radioactive decay	becquerel (Bq)	curie (Ci)	1 Bq = 2.7 × 10^{-11} Ci

Exposure

Exposure is a measure of the capacity of x-rays or γ-rays to ionize air. It is measured as the amount of charge per mass of air—**coulombs/kg.** It is a measure of the intensity of the radiation field as opposed to the amount of radiation absorbed, although there is a direct relationship. The roentgen has been largely replaced by the SI equivalent unit of air kerma.

Traditional unit: **roentgen (R)**
1 R = 2.58 × 10^{-4} C/kg
One R will produce 2.08 × 10^{8} ion pairs in 1 cm^3 of air.

Air Kerma

When radiation interacts with matter via photoelectric absorption and Compton scattering, it transfers energy to electrons of the absorber. The **kerma,** an acronym for *k*inetic *e*nergy *r*eleased in *ma*tter, measures the kinetic energy transferred from photons to electrons and is expressed in units of dose (gray [Gy]), where 1 Gy equals 1 J/kg. Kerma is the sum of the initial kinetic energies of all the charged particles liberated by uncharged ionizing radiation (e.g., x-rays) in a sample of matter divided by the mass of the sample. Kerma values made in air are called air kerma. The kerma is rapidly replacing exposure measured in coulombs/kg or R. An exposure of 1 R results in an air kerma of approximately 8.77 mGy.

Absorbed Dose

Absorbed dose is a measure of the total energy absorbed by any type of ionizing radiation per unit of mass of any type of matter. It varies with the type and energy of radiation and the type of matter absorbing the energy.

> SI unit: gray, where 1 Gy = 1 J/kg
> Traditional unit: rad (radiation absorbed dose)
> 1 rad = 100 ergs/g of absorber
> 1 Gy = 100 rad.

Equivalent (Radiation-Weighted) Dose

The equivalent dose (H_T) is used to compare the biologic effects of different types of radiation on a tissue or organ. Particulate radiations have a high LET and are more damaging to tissue than is radiation with low LET, such as x-rays. Thus deposition of 1 Gy of α particles causes much more biologic damage than 1 Gy of x-ray photons. The equivalent dose considers not only the absorbed dose but also this relative biologic effectiveness of the incident radiation using a radiation-weighting factor (W_R). The W_{R} of photons, the reference, is 1. The W_R of 5-keV neutrons and high-energy protons is 5, and the W_R of α particles is 20. The equivalent dose (H_T) is computed as the product of the radiation-weighting factor (W_R) and the absorbed dose averaged over a tissue or organ (D_T).

$$H_T = W_R \times D_T$$

SI Unit: Sievert (Sv)
For x-rays, 1 Sv = 1 Gy
Traditional unit: rem (roentgen equivalent mammal)
1 Sv = 100 rems

Effective Dose

The effective dose *(E)* is used to estimate the risk in humans. It is hard to compare the risk from a dental exposure with, for example, the risk from a radiographic chest examination because different tissues with different radiosensitivities are exposed. To allow such comparisons, the effective dose is a calculation that considers the relative biologic effectiveness of different types of radiation *and* the radiosensitivity of different tissues exposed in terms of the risk for stochastic effects of radiation (cancer induction and heritable effects). Tissue weighting factors *(W_T)* have been developed to factor individual tissue radiosensitivity (Table 1.5). *E* is the sum of the products of the equivalent dose to each organ or tissue *(H_T)* and the tissue weighting factor *(W_T)*

TABLE 1.5

Tissue Weighting Factors[a]

Tissue	**Tissue Weighting Factor *(W_T)***
Bone marrow, colon, lung, stomach, breast, remainder tissues[b]	0.12
Gonads	0.08
Bladder, esophagus, liver, thyroid	0.04
Bone surface, brain, salivary glands, skin	0.01

[a]ICRP Publication 103: The 2007 Recommendations of the International Commission on Radiological Protection.

Adrenals, extrathoracic region, gallbladder, heart, kidneys, lymphatic nodes, muscle, oral mucosa, pancreas, prostate, small intestine, spleen, thymus, uterus/cervix.

$$E = \sum W_T \times H_T$$

SI Unit: Sievert (Sv)

Traditional unit: rem (roentgen equivalent mammal)
1 Sv = 100 rems

Radioactivity

The measurement of radioactivity *(A)* describes the decay rate of a sample of radioactive material. Although not directly applicable to dentomaxillofacial radiography, diagnostic nuclear medicine examinations indicate the amount of radiopharmaceutical delivered to the patient using the following units.

SI Unit: becquerel (Bq)
1 Bq = 1 disintegration per second (dps)
Traditional unit: curie (Ci)
1 Ci = 3.7×10^{10} dps
1 Bq = 2.7×10^{-11} Ci
1 mCi = 37 megaBq

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