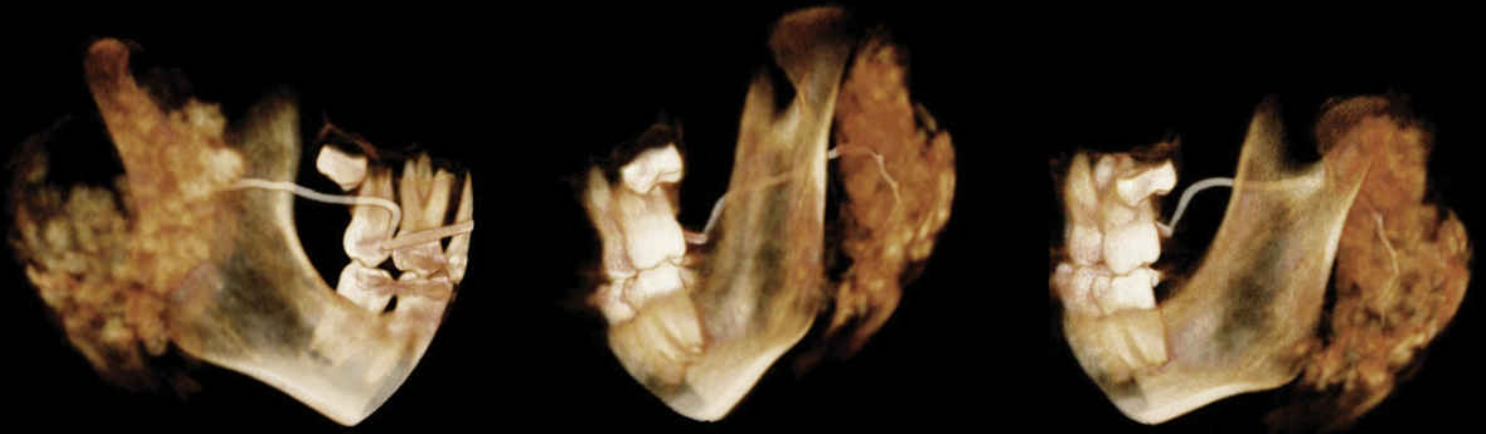


8th
EDITION

White and Pharoah's

ORAL RADIOLOGY

Principles and Interpretation



Sanjay M. **Mallya**
Ernest W.N. **Lam**

ELSEVIER

White and Pharoah's Oral Radiology

Principles and Interpretation

8TH EDITION

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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)

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PART I

Foundations

OUTLINE

- 1 Physics
- 2 Biologic Effects of Ionizing Radiation
- 3 Safety and Protection

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).

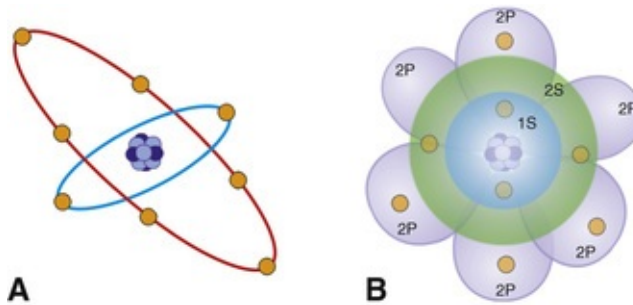


FIG. 1.1 (A) Schematic view of the Bohr model of the oxygen atom showing a nucleus with electrons that travel around the nucleus in circular orbits. (B) Schematic view of the quantum mechanical model of the oxygen atom. The central nucleus is surrounded by an electron cloud that represents probability 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*; $n = 1, 2, 3 \dots$) 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

^aElementary 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, ¹⁸F-fluorodeoxyglucose (¹⁸F-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.

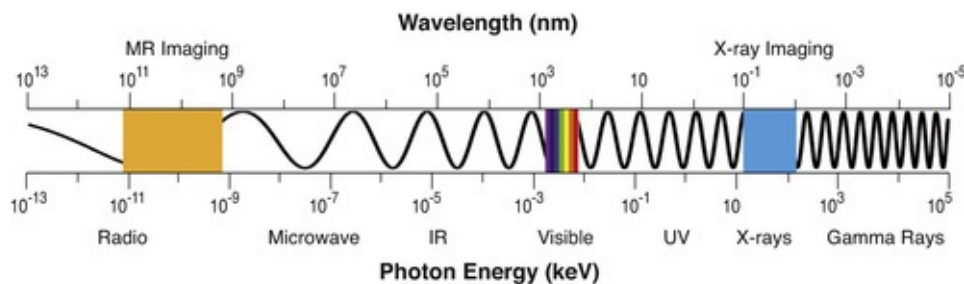


FIG. 1.2 Electromagnetic spectrum showing the relationship between photon wavelength and energy and the physical properties of various portions of the spectrum. Photons with shorter wavelengths have higher energy. Photons used in dental radiography (*blue*) have energies of 10 to 120 keV. Magnetic resonance (*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).

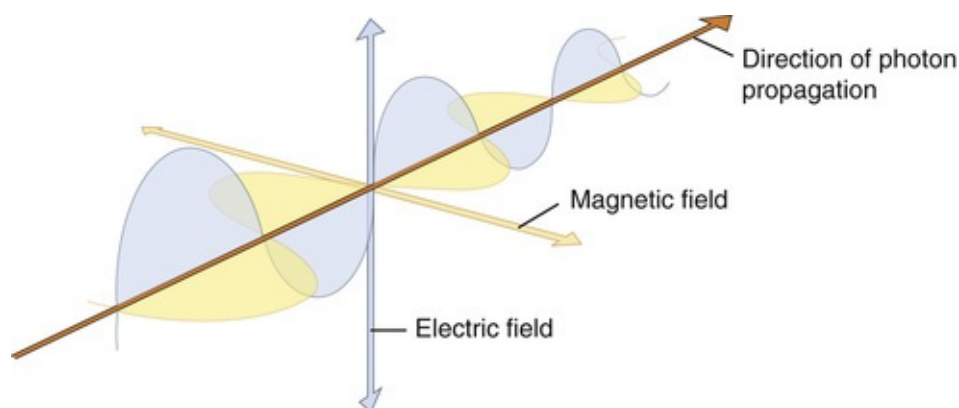


FIG. 1.3 Electric and magnetic fields associated with electromagnetic radiation.

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

$E = h \times \frac{c}{\lambda}$	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)
simplified as $E = \frac{1.24}{\lambda}$ $E \propto \frac{1}{\lambda}$	

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.)

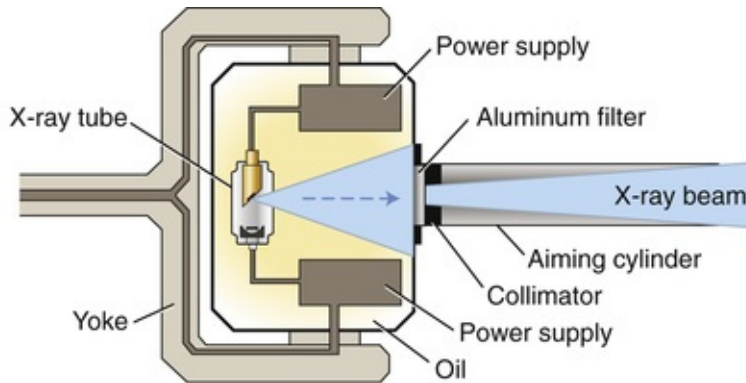


FIG. 1.5 Tube head showing a recessed x-ray tube, components of the power supply, and oil that conducts heat away from the x-ray tube. Path of useful x-ray beam (*blue*) from the anode, through the glass wall of the x-ray tube, oil, and finally an aluminum filter. The beam size is restricted by the metal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

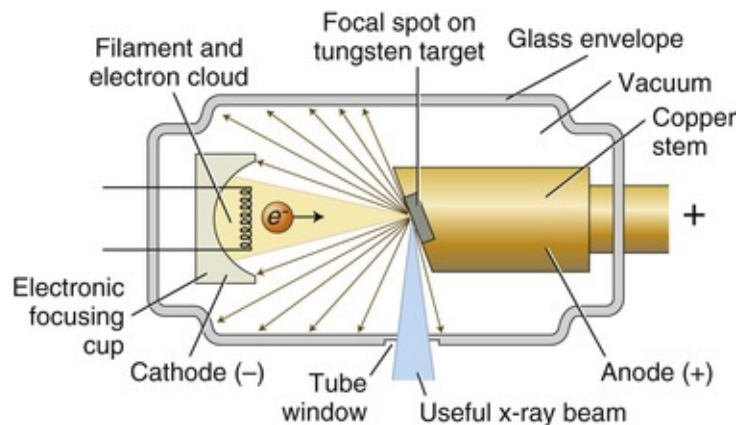


FIG. 1.6 X-ray tube with the major components labeled. The path of the electron beam is shown in *yellow*. X-rays produced at the target travel in all directions. The useful x-ray beam is shown in *blue*.

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.

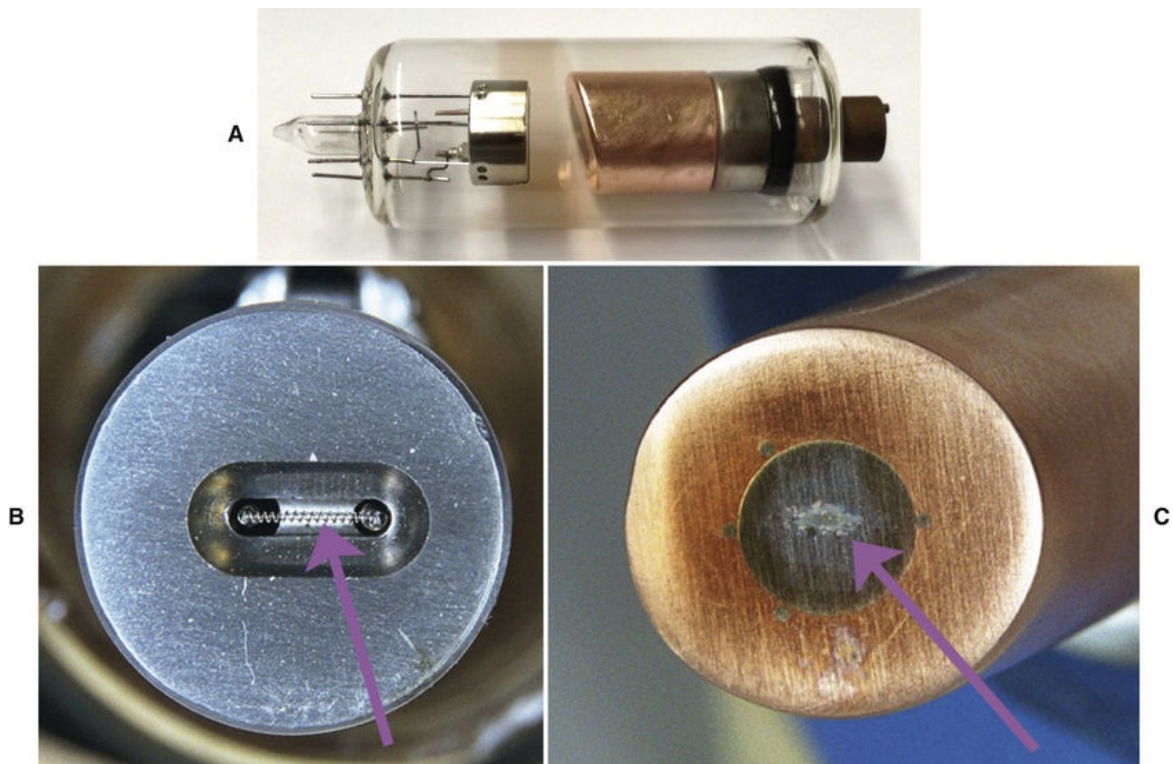


FIG. 1.7 (A) Dental stationary x-ray tube with cathode on left and copper anode on right. (B) Focusing cup containing a filament (*arrow*) in the cathode. (C) Copper anode with tungsten inset. Note the elongated actual focal spot area (*arrow*) on the tungsten target of the anode. ([B] and [C], Courtesy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

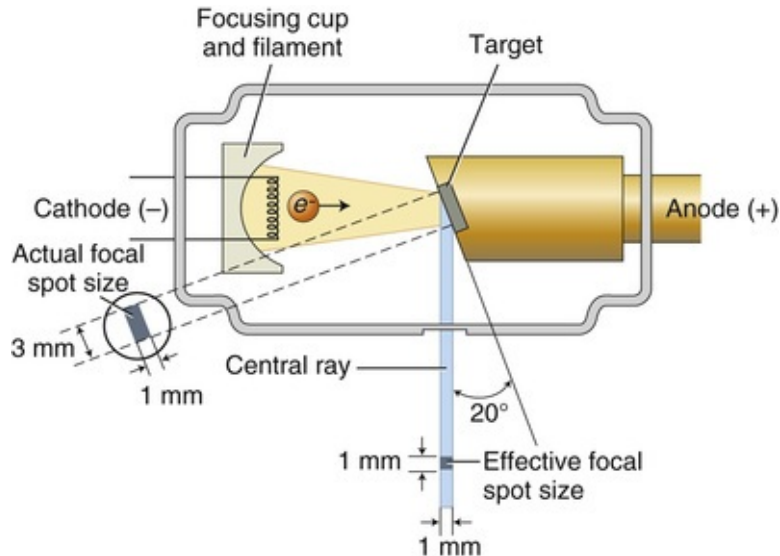


FIG. 1.8 The angle of the target to the central ray of the x-ray beam has a strong influence on the apparent size of the focal spot. The projected effective focal spot (seen below the target) is much smaller than the actual focal spot size (projected to the left). This provides a beam that has a small 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.

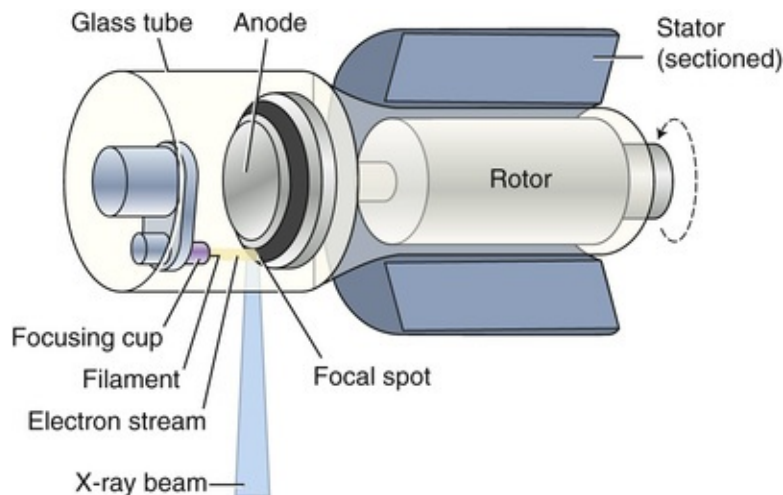


FIG. 1.9 X-ray tube with a rotating anode allows heat at the focal spot to spread out over a large surface area (*dark band*). Current applied to the stator induces rapid rotation of the rotor and the anode. The path of the electron beam is shown in *yellow*, and the useful x-ray beam is shown in *blue*.

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).

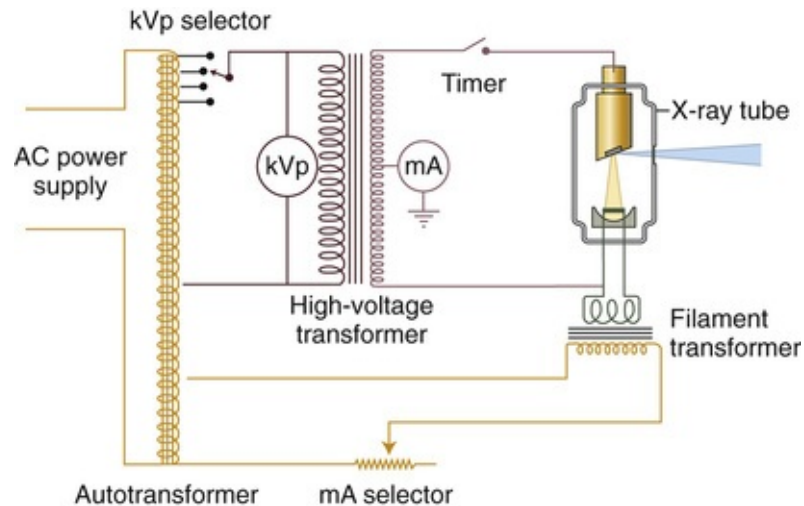


FIG. 1.10 Schematic of dental x-ray machine circuitry and x-ray tube with the major components labeled. The operator selects the desired kVp from the autotransformer. The voltage is greatly increased by the high-voltage step-up transformer and applied to the x-ray tube. The kVp dial measures the voltage 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 milliamperage 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](#)).

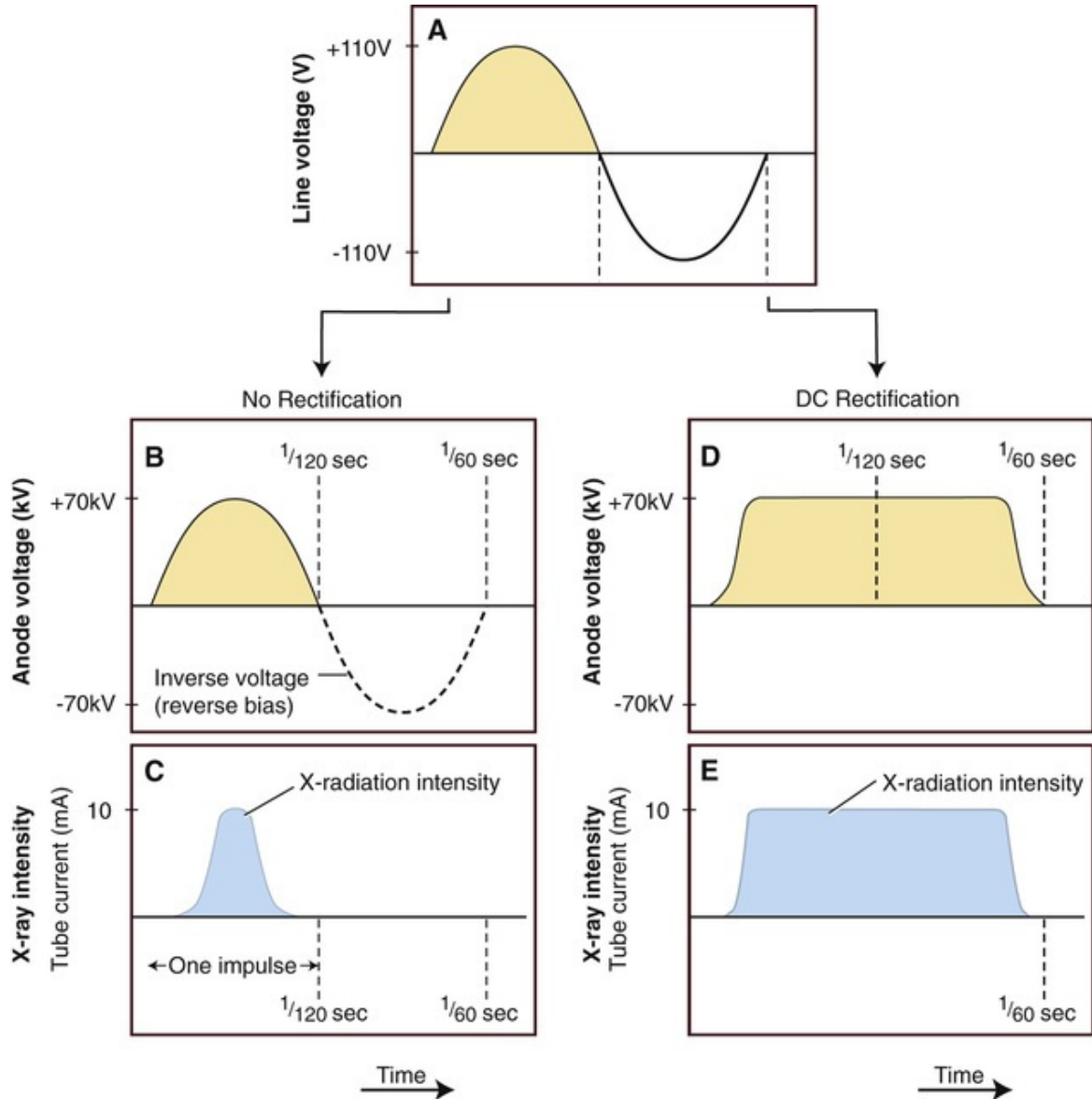


FIG. 1.11 (A) Incoming alternate current line voltage (110 V, 60 cycles per second in this case). (B) Voltage at the anode varies from zero up to the kVp setting (70 kVp in this case). (C) The intensity of radiation produced at the anode (*blue*) is strongly dependent on the anode voltage and is highest 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 $HU = kVp \times mA \times \text{seconds}$. 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.

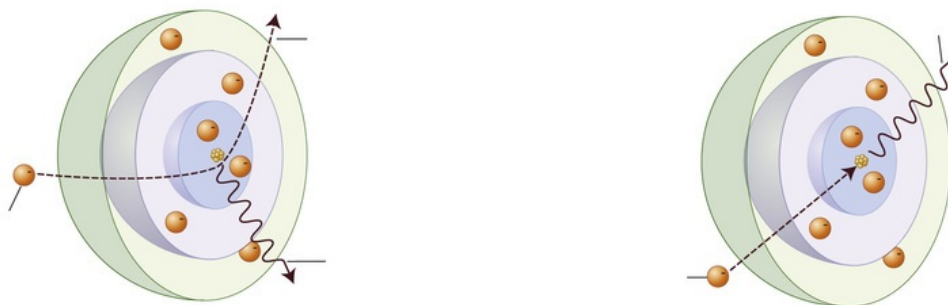


FIG. 1.12 Bremsstrahlung radiation is produced most often by the passage of an electron near a nucleus, which results in electrons being deflected and decelerated (A) or, less frequently, by the direct hit of an electron on a nucleus in the target (B). For the sake of clarity, this diagram

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:

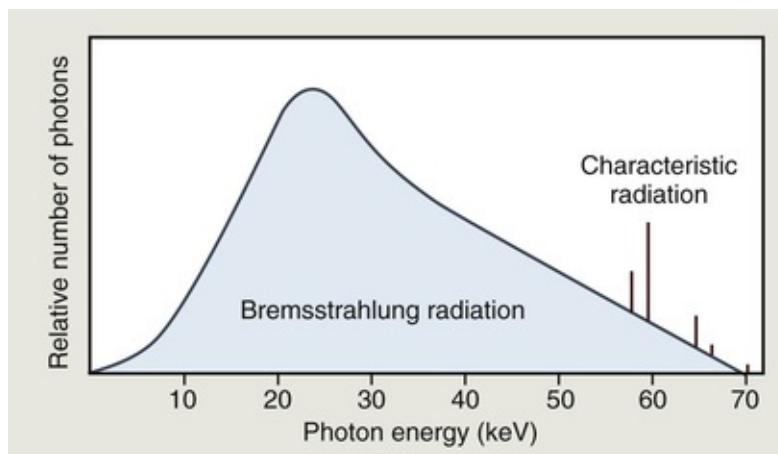


FIG. 1.13 Spectrum of photons emitted from an x-ray machine operating at 70 kVp. The vast preponderance of radiation is bremsstrahlung (*blue*), with a minor addition of characteristic radiation.

- 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.

