ORAL RADIOLOGY
Principles and Interpretation
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Each new edition of this textbook provides the opportunity to include recent progress in the rapidly changing field of diagnostic imaging. We note with particular satisfaction the recent recognition by the American Dental Association of Oral and Maxillofacial Radiology as their ninth specialty. The ADA's definition of oral and maxillofacial radiology is: the specialty of dentistry and discipline of radiology concerned with the production and interpretation of images and data produced by all modalities of radiant energy that are used for the diagnosis and management of diseases, disorders, and conditions of the oral and maxillofacial region. It is the continuing goal of our text to present the underlying science of diagnostic imaging as well as describe the core principles of image production and interpretation for the dental student.

In this edition the chapters on panoramic imaging and extraoral radiographic examinations have been extensively rewritten and expanded beyond radiographic technique and anatomy to provide new emphasis on radiographic interpretation. These chapters now lead the reader through systematic approaches to evaluate the complex anatomic relationships displayed on panoramic and skull radiographs for evidence of abnormalities. In recent years there has been a major move to digital imaging in dentistry. We added a new chapter, Digital Imaging, that explains the various types of digital imaging available in dentistry, the underlying concepts of how the different methods work, and their comparative strengths and weaknesses in clinical usage. Just within the last two years cone-beam tomography has become available in dentistry. The chapter on specialized radiographic techniques now explains how this technology works and how it is best used in dentistry. The chapter on dental caries has also been expanded and updated. The chapters on radiographic manifestations of disease in the orofacial region have been updated to include the latest information on etiology and diagnosis. The chapter on orofacial implants has been expanded and updated to keep abreast of this rapidly changing field. We improved the images throughout the book and added examples of advanced imaging where appropriate.
Acknowledgments

We have drawn upon the special talents of many of our colleagues as authors of chapters, some for the first time and others for return visits. We thank all for sharing their knowledge and skills. In particular we welcome the first timers: Dr. Alan G. Lurie—Panoramic Imaging; Drs. Sotirios Tetradis and Mel L. Kantor—Extraoral Radiographic Examinations; Drs. John B. Ludlow and André Mol—Digital Imaging; Dr. Ann Wenzel—Dental Caries; Dr. Ernest W.N. Lam—Paranasal Sinuses; Dr. Laurie C. Carter—Soft Tissue Calcification and Ossification; and Dr. Carol Anne Murdoch-Kinch—Developmental Disturbances of the Face and Jaws. We also wish to acknowledge the insightful thoughts of Mr. Charlie Brayer and Ms. Ruth K. Arbuckle related to x-ray film and intensifying screens. The immense knowledge and experience of all these individuals adds immeasurably to this text. We are most grateful for the skillful and generous support from the staff at Elsevier for their energy and creativity in the presentation of the content. And finally, we thank our students whose sharp eyes and minds constantly discover new ways for us to improve this book.

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PART ONE

The Physics of Ionizing Radiation
PART ONE

The Physics of Ionizing Radiation
Composition of Matter

Matter is anything that occupies space and has inertia. It has mass and can exert force or be acted on by a force. Matter occurs in three states—solid, liquid, and gas—and may be divided into elements and compounds. Atoms, the fundamental units of elements, cannot be subdivided by ordinary chemical methods but may be broken down into smaller (subatomic) particles by special high-energy techniques. More than 100 subatomic particles have been described; the so-called fundamental particles (electrons, protons, and neutrons) are of greatest interest in radiology because the generation, emission, and absorption of radiation occur at the subatomic level.

ATOMIC STRUCTURE

Because the atom cannot be directly observed, various models are used to describe its structure, each of which is capable of explaining observable actions. The phenomena associated with radiology employ the quantum mechanical model proposed by Niels Bohr in 1913. Bohr conceived the atom as a miniature solar system, at the center of which is the nucleus, analogous to the sun. Electrons revolve around this nucleus at high speeds, analogous to the planets orbiting the sun. In all atoms except hydrogen, the nucleus consists of two primary subatomic particles: protons and neutrons. A single proton constitutes the nucleus of the hydrogen atom.

Electrons orbit the nucleus of all atoms. All electrons are alike, as are all protons and neutrons.

Figure 1-1, A, illustrates Bohr's model, using a stylized rendering of three atoms. The paths of the electrons are drawn as sharply defined orbits to facilitate graphic representation of the generation of x rays and their interaction with matter. In reality the orbit should be represented by broad parameters defining a space in which the electron is most likely to be found. The orbits, or shells, lie at defined distances from the nucleus and are identified by a letter (Fig. 1-1, B). The innermost shell is the K shell, and the next in order are the L, M, N, O, P, and Q shells. The shells also have numbers for identification: 1 for the K shell, 2 for the L shell, and so on. These are the principal quantum numbers, represented by the letter n. No known atom has more than seven shells. Only two electrons may occupy the K shell, with increasingly larger numbers of electrons occupying the outer shells. The maximal number of electrons in a given shell is 2(n^2), where n is the principal quantum number.

Electrons, protons, and neutrons have unique characteristics. The electron carries an electrical charge of −1, the proton a charge of +1, and the neutron no charge at all. The mass of an electron at rest is about 9.1 × 10^-28 g. In contrast, the mass of a proton is 1.67 × 10^-24 g, which is 1836 times the mass of an electron. The mass of a neutron is 1.68 × 10^-24 g, making it 1852 times heavier than an electron and slightly heavier than a proton. Most of the mass of an atom consists of protons and neutrons concentrated in the nucleus. The nucleus contributes only a small fraction (about 1/100,000) of the total size of an atom; most of the size of an atom consists of the cloud of orbiting electrons.

The number of protons contained in the nucleus determines the positive charge. Because any atom in its ground state is electrically neutral, the total number of protons and electrons it carries must be the same. The number of protons in the nucleus also determines the identity of an element. This is its atomic number (Z). Consequently, each of the more than 100 elements has a definitive atomic number, a corresponding number of orbital electrons, and unique chemical and physical properties. Nearly the entire mass of the atom consists of the protons and neutrons in the nucleus. The total number of protons and neutrons in the nucleus of an atom is its atomic mass (A).

The electrostatic attraction between a positively charged nucleus and its negatively charged electrons balances the centrifugal force of the rapidly revolving electrons and maintains them in their orbits. Consequently, the amount of energy required to remove an electron from a given shell must exceed the electrostatic force of attraction between it and the nucleus. This is called the electron binding energy of the electron (or ionization energy) and is specific for each shell of each element. Electrons in the K shell of a given element have the greatest binding energy because they are closest to the nucleus. The binding energy of the electrons in each successive shell decreases. For an electron to move from a specific orbit to another orbit farther from the nucleus, energy must be supplied in an amount equal to the difference in binding energies between the two orbits. In contrast, in moving an electron from an outer orbit to one closer to the nucleus, energy is lost and given up in the form of electromagnetic radiation (see “Characteristic Radiation,” p. 13). The K-shell electrons or any other electrons of large (high-Z) atoms have greater binding energies than those in comparable shells of smaller (low-Z) atoms. This is because large atoms have more protons and thus bind the orbital electrons more tightly to the nucleus than do small atoms.

IONIZATION

When the number of orbiting electrons in an atom is equal to the number of protons in its nucleus, the atom is electrically neutral. If an electrically neutral atom loses an electron, it becomes a positive ion and the free electron is a negative ion. This process of forming an ion pair is termed ionization. Electrons can be lost from an atom by heating or by interactions (collisions) with high-energy x rays or particles such as protons. Such ionization requires sufficient energy to overcome the electrostatic force binding the electrons to the nucleus. The electrons in the inner shells (K, L, and M) are so tightly bound to the nucleus that only x rays, gamma rays, and high-energy particles can remove them. In contrast, the electrons in the outer shells have such low binding energies that they can be easily displaced by photons of lower energy (e.g., ultraviolet or visible light).

Nature of Radiation

Radiation is the transmission of energy through space and matter. It may occur in two forms: particulate and electromagnetic.

PARTICULATE RADIATION

Particulate radiation consists of atomic nuclei or subatomic particles moving at high velocity. Alpha parti-
icles, beta particles, and cathode rays are examples of particulate radiation. Alpha particles are helium nuclei consisting of two protons and two neutrons. They result from the radioactive decay of many elements. Because of their double charge and heavy mass, alpha particles densely ionize matter through which they pass. Accordingly, they quickly give up their energy and penetrate only a few microns of body tissue. (An ordinary sheet of paper absorbs them.) After stopping, alpha particles acquiring two electrons and become neutral helium atoms.

Beta particles are electrons emitted by radioactive nuclei. High-speed beta particles are able to penetrate matter to a greater depth than alpha particles, to a maximum of 1.5 cm in tissue. This deeper penetration occurs because beta particles are smaller and lighter and carry a single negative charge; therefore they have a much lower probability of interacting with matter than do alpha particles. Accordingly, they ionize matter much less densely than alpha particles. Beta particles are used in radiation therapy for treatment of skin lesions. Cathode rays are also high-speed electrons but are produced by manufactured devices (e.g., x-ray tubes).

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). A particle loses kinetic energy each time it ionizes adjacent matter; the greater its physical size and charge and the lower its velocity, the greater is its LET. For example, alpha particles, with their high charge and low velocity, lose kinetic energy rapidly and have short path lengths (are densely ionizing); thus they have a high LET. Beta particles are much less densely ionizing because of their lighter mass and lower charge and thus have a lower LET. They penetrate through tissue more readily than do alpha particles.

**ELECTROMAGNETIC RADIATION**

Electromagnetic radiation is the movement of energy through space as a combination of electric and magnetic fields (Fig. 1-2). It is generated when the velocity of an electrically charged particle is altered. Gamma rays, x rays, ultraviolet rays, visible light, infrared radiation (heat), microwaves, and radio waves are all examples of electromagnetic radiation (Fig. 1-3). Gamma rays are photons that originate in the nuclei of radioactive atoms. They typically have greater energy than x rays. X rays, in contrast, are produced extranuclearly from the interaction of electrons with nuclei in x-ray machines. The types of radiation in this spectrum are ionizing or nonionizing, depending on their energy. If sufficient energy is associated with the radiation to remove orbital electrons from atoms in the irradiated matter, the radiation is ionizing.

Some of the properties of electromagnetic radiation are best expressed by wave theory, whereas others are most successfully described by quantum theory. The wave theory of electromagnetic radiation maintains that radiation is propagated in the form of waves, not unlike 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. 1-4). All electromagnetic waves travel at the velocity of light \( c = 3.0 \times 10^8 \text{ m per sec} \) in a vacuum. Waves of all kinds exhibit the properties of wavelength (\( \lambda \)) and frequency (\( \nu \)). Wavelength and frequency of electromagnetic radiation are related as follows:

\[
\lambda \times \nu = c = 3 \times 10^8 \text{ meters/second}
\]

where \( \lambda \) is in meters and \( \nu \) is in cycles per second (hertz). 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.

Quantum theory considers electromagnetic radiation as small bundles of energy called *photons*. Each photon travels at the speed of light and contains a
specific amount of energy. The unit of photon energy is the electron volt (eV) (Fig. 1-5). The relationship between wavelength and photon energy is as follows:

$$ E = h \times \frac{c}{\lambda} $$

where $E$ is energy in kiloelectron volts (keV), $h$ is Planck's constant ($6.626 \times 10^{-34}$ joule-seconds, or $4.3 \times 10^{-18}$ keV), $c$ is the velocity of light, and $\lambda$ is wavelength in nanometers. This expression may be simplified as follows:

$$ E = 1.24/\lambda $$

The quantum theory of radiation has been successful in correlating experimental data on the interaction of radiation with atoms, the photoelectric effect, and the production of x rays.

Typically, high-energy photons such as x rays and gamma rays are characterized by their energy, whereas lower-energy photons (ultraviolet through radio waves) are characterized by their wavelength.

**The X-Ray Machine**

The heart of an x-ray machine is the x-ray tube and its power supply. The x-ray tube is positioned within the tube head, along with some components of the power supply (Fig. 1-6). Often the tube is recessed within the
FIG. 1-5 An electron volt is the amount of energy acquired by one electron accelerating through a potential difference of 1 volt (1.602 \( \times 10^{-19} \) joules).

FIG. 1-6 Tube head (including the recessed x-ray tube), components of the power supply, and the oil that conducts heat away from the x-ray tube.

FIG. 1-7 X-ray tube with the major components labeled.

FIG. 1-8 Dental x-ray machine circuitry with the major components labeled. A, Filament step-down transformer; B, filament current control (mA switch); C, autotransformer; D, kVp selector dial (switch); E, high-voltage transformer; F, x-ray timer (switch); G, tube voltage indicator (volt-meter); H, tube current indicator (ammeter); I, x-ray tube.

X-RAY TUBE

All dental and medical x-ray tubes are called Coolidge tubes because they follow the original design of W. C. Coolidge introduced in 1913. The basic apparatus for generating x rays, the x-ray tube, is composed of a cathode and an anode (Fig. 1-7). The cathode serves as the source of electrons that flow to the anode. The cathode and anode lie within an evacuated glass envelope or tube. When electrons from the cathode strike the target in the anode, they produce x rays. For the x-ray tube to function, a power supply is necessary to (1) heat the filament to generate electrons, and (2) establish a high-voltage potential between the anode and cathode to accelerate the electrons (Fig. 1-8).

Cathode

The cathode (see Fig. 1-7) 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 about 2 mm in diameter and 1 cm or less in length. It is mounted on two stiff wires that support it and carry the electric current. These two mounting wires lead through the glass envelope and connect to both the high- and low-voltage electrical sources. The filament is heated to incandescence by the flow of current from
FIG. 1-9  A, Focusing cup (arrow) containing a filament in the cathode of the tube from a dental x-ray machine. B, Focal spot area (arrows) on the target of the tube. The size and shape of the focal area approximate those of the focusing cup.

the low-voltage source and emits electrons at a rate proportional to the temperature of the filament.

The filament lies in a focusing cup (Fig. 1-9, A; see also Fig. 1-7), a negatively charged concave reflector made of molybdenum. The focusing cup electrostatically focuses the electrons emitted by the incandescent filament into a narrow beam directed at a small rectangular area on the anode called the focal spot (Fig. 1-9, B; see also Fig. 1-7). The electrons move in this direction because they are repelled by the negatively charged cathode and attracted to the positively charged anode. The x-ray tube is evacuated to prevent collision of the moving electrons with gas molecules, which would significantly reduce their speed. This also prevents oxidation and "burnout" of the filament.

Anode
The anode consists of a tungsten target embedded in a copper stem (see Fig. 1-7). The purpose of the target in an x-ray tube is to convert the kinetic energy of the electrons generated from the filament into x-ray photons. This is an inefficient process with more than 99% of the electron kinetic energy converted to heat. The target is made of tungsten, a material that has several characteristics of an ideal target material. It has a high atomic number (74), high melting point, high thermal conductivity, and low vapor pressure at the working temperatures of an x-ray tube. A target made of a high atomic number material is best because it is most efficient in producing x rays. Because heat is generated at the anode, the requirement for a target with a high melting point is clear. Tungsten also has high thermal conductivity, thus dissipating heat into the copper stem. Finally, the low vapor pressure of tungsten at high temperatures also helps maintain the vacuum in the tube at high operating temperatures.

The tungsten target is typically embedded in a large block of copper to dissipate heat. Copper, a good thermal conductor, dissipates heat from the tungsten, thus reducing the risk of the target melting. In addition, insulating oil between the glass envelope and the housing of the tube head carries heat away from the copper stem. This type of anode is a stationary anode.

The focal spot is the area on the target to which the focusing cup directs the electrons from the filament. The sharpness of the radiographic image increases as the size of the focal spot—the radiation source—decreases (see Chapter 5). The heat generated per unit target area, however, becomes greater as the focal spot decreases in size. To take advantage of a small focal spot while distributing the electrons over a larger area of the target, the target is placed at an angle to the electron beam (Fig. 1-10). The projection of the focal spot perpendicular to the electron beam (the effective focal spot) is smaller than the actual size of the focal spot. Typically, the target is inclined about 20 degrees to the central ray of the x-ray beam. This causes the effective focal spot to be almost $1 \times 1$ mm, as opposed to the actual focal spot, which is about $1 \times 3$ mm. The effect is a small apparent source of x rays and an increase in sharpness of the image (see Fig. 5-2) with a larger actual focal spot for heat dissipation.

Another method of dissipating the heat from a small focal spot is to use a rotating anode. In this case the tungsten target is in the form of a beveled disk that rotates when the tube is in operation (Fig. 1-11). As a result, the
electrons strike successive areas of the target, widening the focal spot by an amount corresponding to the circumference of the beveled disk and distributing the heat over this expanded area. As a consequence, small focal spots can be used with tube currents of 100 to 500 milliamperes (mA), 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 about 3000 revolutions per minute) lie outside the tube. Such rotating anodes are not used in intraoral dental x-ray machines but may be used in tomographic or cephalometric units and in medical x-ray machines requiring higher radiation output.

POWER SUPPLY

A brief review of some aspects of an electric circuit may be useful in understanding the power supply in an x-ray machine. An electric current is the movement of electrons in a conductor, for example, a wire. The rate of the current flow—the number of electrons moving past a point in a second—is measured in amperes. It depends on two factors: the pressure, or voltage of the current, measured in volts, and the resistance of the conductor to the flow of electricity, measured in ohms. Ohm's law relates these units:

\[ V = I \times R \]

where \( V \) is the electric potential in volts, \( I \) is the current flow in amperes, and \( R \) is the resistance of the conductor in ohms. Such an electric circuit is often compared to a simple water supply system in which the rate of water flow through a pipe (amperes) depends both on the water pressure (volts) and the pipe resistance or diameter (ohms).

The primary functions of the power supply of an x-ray machine are to (1) provide a low-voltage current to heat the x-ray tube filament by use of a step-down transformer and (2) generate a high potential difference between the anode and cathode by use of a high-voltage transformer. These transformers and the x-ray tube lie within an electrically grounded metal housing called the head of the x-ray machine. An electrical insulating material, usually oil, surrounds the transformers.

Tube Current

The filament step-down transformer (see Fig. 1-8, A) reduces the voltage of the incoming alternating current (AC) to about 10 volts. Its operation is regulated by the filament current control (mA switch) (see Fig. 1-8, B), which adjusts the resistance and thus the current flow through the low-voltage circuit, including the filament. This in turn regulates the temperature of the filament and thus the number of electrons emitted. The tube current is the flow of electrons through the tube, that is, from the filament to the anode and then back to the filament through the wiring of the power supply. The mA setting on the filament current control refers to the tube current, which is measured by the ammeter (see Fig. 1-8, H).

Tube Voltage

A high voltage is required between the anode and cathode to generate x rays. An autotransformer (see
Fig. 1-8, C) converts the primary voltage from the input source into the secondary voltage. The secondary voltage regulated by the kilovolts peak (kVp) selector dial (see Fig. 1-8, D). The kVp dial selects a voltage from different levels on the autotransformer and applies it across the primary winding of the high-voltage transformer. The kVp dial therefore controls the voltage between the anode and cathode of the x-ray tube. The high-voltage transformer (see Fig. 1-8, E) provides the high voltage required by the x-ray tube to accelerate the electrons from the cathode to the anode and generate x rays. It accomplishes this by boosting the peak voltage of the incoming line current to as high as 60 to 100 kV, thus boosting the peak energy of the electrons passing through the tube to as high as 60 to 100 keV. The kVp selector dial setting thus determines the peak kilovoltage across the tube (see Fig. 1-8, I).

Because the line current is AC (60 cycles per second), the polarity of the x-ray tube alternates at the same frequency (Fig. 1-12, A). 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 current flows through the tube (Fig. 1-12, B). Because the line voltage is variable, the voltage potential between the anode and cathode varies. As the
tube voltage is increased, the speed of the electrons toward the anode increases. When the electrons strike the focal spot of the target, some of their energy converts to x-ray photons. X rays are produced at the target with greatest efficiency when the voltage applied across the tube is high. Therefore the intensity of x-ray pulses tends to be sharply peaked at the center of each cycle (Fig. 1-12, C). During the following half (or negative half) of the cycle, the polarity of the AC reverses, and the filament becomes positive and the target negative (see Fig. 1-12, B). At these times the electrons stay in the vicinity of the filament and do not flow across the gap between the two elements of the tube. This half of the cycle is called inverse voltage or reverse bias (see Fig. 1-12, B). No x rays are generated during this half of the voltage cycle (see Fig. 1-12, C). Therefore 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. This type of power supply circuitry, in which the alternating high voltage is applied directly across the x-ray tube, limits x-ray production to half the AC cycle and is called self-rectified or half-wave rectified. Almost all conventional dental x-ray machines are self-rectified.

A tube energized with a self-rectifying power supply must not be operated for extended periods. With overuse the target may get so hot that it emits electrons, and during the negative half cycle, the inverse voltage may drive electrons from the target to the filament, causing the filament to overheat and melt. The glass envelope also may be damaged if the electrons are driven in the wrong direction by the reverse bias on the tube.

Some dental x-ray manufacturers produce machines that replace the conventional 60-cycle AC high-voltage current of the x-ray tube with a high-frequency power supply. This effect is an essentially constant potential between the anode and cathode. The result is that the mean energy of the x-ray beam produced by these x-ray machines is higher than that from a conventional half-wave rectified machine operated at the same voltage. This is because the number of lower-energy (nondiagnostic) x rays is reduced. These photons are produced as the voltage across the x-ray tube rises from zero to its peak and then decreases back again to zero during the voltage cycle in the half-wave rectified machine. For a given voltage setting and radiographic density, the images resulting from these constant-potential machines have a longer contrast scale and lower patient dose compared with conventional x-ray machines.

**TIMER**

A timer is built into the high-voltage circuit to control the duration of the x-ray exposure (see Fig. 1-8, F). The timer controls the length of time that high voltage is applied to the tube and therefore the time during which tube current flows and x rays are produced. Before the high voltage is applied across the tube, however, 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 is not practical because maintaining the filament at a high temperature for a long period shortens its life. Failure of the filament is a common source of malfunction of x-ray tubes. To minimize filament burnout, the timing circuit first sends a current through the filament for about half a second to bring it to the proper operating temperature. After the filament is heated, the timer 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. In this case the delay to preheat the filament before each exposure is even shorter. Accordingly, the machine should be left on continuously during working hours.

Some x-ray machine timers are calibrated in fractions and whole numbers of seconds. The time intervals on other timers are expressed as number of impulses per exposure (e.g., 3, 6, 9, 15). Such numbers represent the number of impulses of radiation emitted during the exposure; thus the number of impulses divided by 60 (the frequency of the power source) gives the exposure time in seconds. Therefore a setting of 30 impulses means that there will be 30 impulses of radiation and is equivalent to a half-second exposure.

**TUBE RATING AND DUTY CYCLE**

Each x-ray machine comes with tube rating specifications that describe the maximal exposure time the tube can be energized without risk of damage to the target from overheating. These specifications describe in graph form the maximal safe intervals (seconds) that the tube can be used for a range of voltages (kVp) and filament current (mA) values. These tube ratings generally do not impose any restrictions on tube use for daily intraoral radiography. If a dental x-ray unit is to be used for extraoral exposures, however, the tube-rating chart should be mounted by the machine for easy reference.

Duty cycle relates to the frequency with which successive exposures can be made. The heat buildup at the anode is measured in heat units defined by the following equation: heat units (HU) = kVp \times mA \times seconds. The heat storage capacity for anodes of dental diagnostic tubes is approximately 20 kHU. Because of heat generated at the anode, the interval between successive exposures must be long enough for its dissipation.
characteristic is a function of the size of the anode and the method used to cool it. The cooling characteristics of anodes are described by the maximal number of heat units it can store without damage and the heat dissipation rate, which can be determined from the cooling curves provided by the manufacturer of each tube.

Production of X Rays

Electrons traveling from the filament to the target convert some of their kinetic energy into x-ray photons by the formation of bremsstrahlung and characteristic radiation.

BREMSSTRahlUNG

Bremsstrahlung interactions, the primary source of x-ray photons from an x-ray tube, are produced by the sudden stopping or slowing of high-speed electrons at the target. (Bremsstrahlung means "braking radiation" in German.) When electrons from the filament strike the tungsten target, x-ray photons are created if the electrons hit a target nucleus directly or if their path takes them close to a nucleus. If a high-speed electron directly hits the nucleus of a target atom, all its kinetic energy is transformed into a single x-ray photon (Fig. 1-13, A). The energy of the resultant photon (in keV) is numerically equal to the energy of the electron. This in turn is equal to the kilovoltage applied across the x-ray tube at the instant of its passage.

Most high-speed electrons, however, have near or wide misses with atomic nuclei (Fig. 1-13, B). In these interactions, a negatively charged high-speed electron is attracted toward the positively charged nuclei and loses some of its velocity. This deceleration causes the electron to lose some kinetic energy, which is given off in the form of many new photons. The closer the high-speed electron approaches the nuclei, the greater is the electrostatic attraction on the electron, the braking effect, and the energy of the resulting bremsstrahlung photons.

Bremsstrahlung interactions generate x-ray photons with a continuous spectrum of energy. The energy of an x-ray beam may be described by identifying the peak operating voltage (in kVp). A dental x-ray machine operating at a peak voltage of 70,000 volts (70 kVp), for example, applies a fluctuating voltage of as much as 70 kVp across the tube. This tube therefore produces x-ray photons with energies ranging to a maximum of 70,000 eV (70 keV). Fig. 1-14 demonstrates the continuous spectrum of photon energies produced by an x-ray machine operating at 100 kVp. The reasons for this continuous spectrum are as follows:

1. The continuously varying voltage difference between the target and filament, which is characteristic of half-wave rectification, causes the electrons striking the target to have varying levels of kinetic energy.
2. 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.
3. Most electrons participate in many bremsstrahlung interactions in the target before losing all their kinetic energy. As a consequence, an electron carries differing amounts of energy at the time of each interaction with a tungsten nucleus that results in the generation of an x-ray photon.
CHAPTER 1 RADIATION PHYSICS

**Bremsstrahlung radiation**

characteristic of the target atoms. Characteristic radiation is only a minor source of radiation from an x-ray tube.

**Factors Controlling the X-Ray Beam**

The x-ray beam emitted from an x-ray tube may be modified by altering the beam exposure length (timer), exposure rate (mA), beam energy (kVp and filtration), beam shape (collimation), and target-patient distance.

**EXPOSURE TIME**

Figure 1-16 portrays the changes in the x-ray spectrum that result when the exposure time is increased while the tube current (mA) and voltage (kVp) remain constant. When the exposure time is doubled, the number of photons generated at all energies in the x-ray emission spectrum is doubled, but the range of photon energies is unchanged. Therefore changing the time simply controls the *quantity* of the exposure, the number of photons generated.

**TUBE CURRENT (mA)**

Figure 1-17 illustrates the changes in the spectrum of photons that result from increasing tube current (mA) while maintaining constant tube voltage (kVp) and exposure time. 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. Therefore the quantity of radiation produced by an x-ray tube (i.e., the number of photons that reach the patient and film) is directly proportional to the increase in tube current.

**CHARACTERISTIC RADIATION**

Characteristic radiation occurs when an electron from the filament displaces an electron from a shell of a tungsten target atom, thereby ionizing the atom. When this happens, a higher energy electron in an outer shell of the tungsten atom is quickly attracted to the void in the deficient inner shell (Fig. 1-15). When the outer-shell electron replaces the displaced electron, a photon is emitted with an energy equivalent to the difference in the two orbital binding energies. Characteristic radiation from the K shell occurs only above 70 kVp with a tungsten target and occurs as discrete increments compared with bremsstrahlung radiation (see Fig. 1-14). The energies of characteristic photons are a function of the energy levels of various electron orbital levels and hence are characteristic of the target atoms. Characteristic radiation is only a minor source of radiation from an x-ray tube.

**FIG. 1-15** Characteristic radiation. A, An incident electron in an inner orbit ejects a photoelectron, creating a vacancy. B, This vacancy is filled by an electron from an outer orbit. C, A photon is emitted with energy equal to the difference in energy levels between the two orbits. D, Electrons from various orbits may be involved, giving rise to other photons. The energies of the photons thus created are characteristic of the target atom.
The spectrum of photon energies showing that as exposure time increases (kVp and tube voltage held constant), so does the total number of photons. The mean energy and maximal energy of the beams are unchanged.

The spectrum of photon energies showing that as the kVp is increased (tube current and exposure time 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.

The spectrum of photon energies showing that as tube current (mA) increases (kVp and exposure time held constant), so does the total number of photons. The mean energy and maximal energy of the beams are unchanged. Compare with Fig. 1-16.

The tube voltage (kVp) affects the spectrum of photon energies in an x-ray beam. Increasing the kVp increases the potential difference between the cathode and anode, thus increasing the energy of each electron when it strikes the target. This results in an increased efficiency of conversion of electron energy into x-ray photons, and thus an increase in (1) the number of photons generated, (2) their mean energy, and (3) their maximal energy. The increased number of photons produced per unit time by use of higher kVp results from the greater efficiency in the production of bremsstrahlung photons that occurs when increased numbers of higher-energy electrons interact with the target.

The ability of x-ray photons to penetrate matter depends on their energy. High-energy x-ray photons have a greater probability of penetrating matter, whereas relatively low-energy photons have a greater probability of being absorbed. Therefore the higher the kVp and mean energy of the x-ray beam, the greater the penetrability of the beam through matter. A useful way to characterize the penetrating quality of an x-ray beam (its energy) is by its half-value layer (HVL). The HVL is the thickness of an absorber, such as aluminum, required to reduce by one half the number of x-ray photons passing through it. As the average energy of an x-ray beam increases, so does its HVL. The term beam quality refers to the mean energy of an x-ray beam.

Filtration

Although an x-ray beam consists of a spectrum of x-ray photons of different energies, only photons with sufficient energy to penetrate through anatomic structures and reach the image receptor (usually film) are useful for diagnostic radiology. Those that are of low energy (long wavelength) contribute to patient exposure (and
Filtering an x-ray beam with aluminum preferentially removes low-energy photons, thereby reducing the beam intensity and increasing its mean energy.

Consequently, to reduce patient dose, the less-penetrating photons should be removed. This can be accomplished, in part, by placing an aluminum filter in the path of the beam. Fig. 1-19 illustrates how the addition of an aluminum filter alters the energy distribution of the unfiltered beam. The aluminum preferentially removes many of the lower-energy photons with lesser effect on the higher-energy photons that are able to penetrate to the film.

In determinations of the amount of filtration required for a particular x-ray machine, kVp and inherent filtration of the tube and its housing must be considered. **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. **Total filtration** is the sum of the inherent filtration plus any added **external filtration** supplied in the form of aluminum disks placed over the port in the head of the x-ray machine. Governmental regulations 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 to 70 kVp, and 2.5 mm of aluminum for all higher voltages (see Chapter 3).

**COLLIMATION**

A collimator is a metallic barrier with an aperture in the middle used to reduce the size of the x-ray beam (Fig. 1-20) and therefore the volume of irradiated tissue within the patient. Round and rectangular collimators are most frequently used in dentistry. Dental x-ray beams are usually collimated to a circle 23/4 inches (7 cm) in diameter. A round collimator (see Fig. 1-20, A) is a thick plate of radiopaque material (usually lead) 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. 1-20, B) further limit the size of the beam to just larger than the x-ray film. It is important to reduce the beam to the size of the film to reduce further unnecessary patient exposure. Some types of film-holding instruments also provide rectangular collimation of the x-ray beam (see Chapters 3 and 8).

Use of collimation also improves image quality. When an x-ray beam is directed at a patient, the tissues
absorb about 90% of the x-ray photons and 10% of the photons pass through the patient and reach the film. Many of the absorbed photons generate scattered radiation within the exposed tissues by a process called Compton scattering (see below). These scattered photons travel in all directions, and some reach the film and degrade image quality. Collimating the beam to reduce the exposure area and thus the number of scattered photons reaching the film can minimize the detrimental effect of scattered radiation on the images.

**INVERSE SQUARE LAW**

The intensity of an x-ray beam at a given point (number of photons per cross-sectional area per unit exposure time) depends on the distance of the measuring device from the focal spot. For a given beam the intensity is inversely proportional to the square of the distance from the source (Fig. 1-21). The reason for this decrease in intensity is that the x-ray beam spreads out as it moves from the source. The relationship is as follows:

\[
\frac{I_1}{I_2} = \left( \frac{D_2}{D_1} \right)^2
\]

where \( I \) is intensity and \( D \) is distance. Therefore if a dose of 1 gray \((\text{Gy})\) is measured at a distance of \(2\) m, a dose of \(4\) Gy will be found at \(1\) m, and \(0.25\) Gy at \(4\) m.

Therefore changing the distance between the x-ray tube and patient has a marked effect on beam intensity. Such a change requires a corresponding modification of the kVp or mAs if the exposure of the film is to be kept constant.

**Interactions of X Rays With Matter**

The intensity of an x-ray beam is reduced by interaction with the matter it encounters. This attenuation results from interactions of individual photons in the beam with atoms in the absorber. The x-ray photons are either absorbed or scattered out of the beam. In absorption, photons ionize absorber atoms and convert their energy into kinetic energy of the absorber electrons. In scattering, photons are ejected out of the primary beam as a result of interactions with the orbital electrons of absorber atoms. In a dental x-ray beam there are three means of beam attenuation: (1) coherent scattering, (2) photoelectric absorption, and (3) Compton scattering (Fig. 1-22). In addition, about 9% of the primary photons pass through the patient without interaction (Table 1-1).

**COHERENT SCATTERING**

Coherent scattering (also known as classical, elastic, or Thompson scattering) may occur when a low-energy incident photon (less than \(-10\) keV) passes near an outer electron of an atom (which has a low binding energy). The incident photon interacts with the electron by causing it to become momentarily excited at the same frequency as the incoming photon (Fig. 1-23). The incident photon ceases to exist. The excited electron then returns to the ground state and generates another x-ray photon with the same frequency and energy as in the incident beam. Usually the secondary photon is emitted at an angle to the path of the incident photon. In effect, the direction of the incident x-ray photon is altered. This interaction accounts for only about 8% of the total number of interactions (per exposure) in a dental examination (see Table 1-1). Coherent scattering contributes very little to film fog because the total quantity...
Scattered photon

FIG. 1-23 Coherent scattering resulting from the interaction of a low-energy incident photon with an outer electron, causing the outer electron to vibrate momentarily. After this, a scattered photon of the same energy is emitted at a different angle from the path of the incident photon.

<table>
<thead>
<tr>
<th>INTERACTION</th>
<th>PRIMARY PHOTONS</th>
<th>SCATTERED PHOTONS*</th>
<th>TOTAL†</th>
</tr>
</thead>
<tbody>
<tr>
<td>Coherent scattering</td>
<td>148,905</td>
<td>156,234</td>
<td>305,139</td>
</tr>
<tr>
<td>Photoelectric</td>
<td>536,208</td>
<td>522,082</td>
<td>1,058,290</td>
</tr>
<tr>
<td>absorption</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Compton scattering</td>
<td>1,131,878</td>
<td>1,098,720</td>
<td>2,230,598</td>
</tr>
<tr>
<td>Exit</td>
<td>183,009</td>
<td>758,701</td>
<td>941,710</td>
</tr>
<tr>
<td>TOTAL</td>
<td>2,000,000</td>
<td>2,535,737</td>
<td>4,535,737</td>
</tr>
</tbody>
</table>

From Gibbs SJ: Personal communication, 1986.
*Scattered photons result from primary, Compton, and coherent interactions.
†Note that the sum of the total number of photoelectric interactions and photons that exit the patient equals the total number of incident photons.

of scattered photons is small and its energy level is too low for much of it to reach the film.

PHOTOELECTRIC ABSORPTION

Photoelectric absorption is critical in diagnostic imaging. This process occurs when an incident photon collides with a bound electron in an atom of the absorbing medium. At this point the incident photon ceases to exist. The electron is ejected from its shell and becomes a recoil electron (photoelectron) (Fig. 1-24). The kinetic energy imparted to the recoil electron is equal to the energy of the incident photon minus that used to overcome the binding energy of the electron. The absorbing atom is now ionized because it has lost an electron. In the case of atoms with low atomic numbers (e.g., those in most biologic molecules), the binding energy is small. As a result the recoil electron acquires most of the energy of the incident photon. Most photoelectric interactions occur in the K shell because the density of the electron cloud is greater in this region and a higher probability of interaction exists. About 30% of photons absorbed from a dental x-ray beam are absorbed by the photoelectric process.

An atom that has participated in a photoelectric interaction is ionized as a result of the loss of an electron. This electron deficiency (usually in the K shell) is instantly filled, usually by an L-shell electron, with the release of characteristic radiation (see Fig. 1-15). Whatever the orbit 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 film.

Recoil electrons ejected during photoelectric absorption travel only short distances in the absorber before they give up their energy through secondary ionizations. As a consequence, all the energy of incident photons that undergo photoelectric interaction is deposited in the patient. Although this is beneficial in producing high-quality radiographs, because no scattered radiation fogs the film, it is potentially deleterious for patients because of increased radiation absorption.

The frequency of photoelectric interaction varies directly with the third power of the atomic number of the absorber. For example, because the effective atomic number of compact bone \( Z = 13.8 \) is greater than that of soft tissue \( Z = 7.4 \), the probability that a photon will be absorbed by a photoelectric interaction in bone is approximately 6.5 times greater than in an equal thickness of soft tissue. This difference is readily seen on dental radiographs as a difference in optical density of the image. It is this difference in the absorption that makes the production of a radiographic image possible.

COMPTON SCATTERING

Compton scattering occurs when a photon interacts with an outer orbital electron (Fig. 1-25). About 62% of the photons that are absorbed from a dental x-ray beam are absorbed by this process. In this interaction the incident photon collides with an outer electron, which
receives kinetic energy and recoils from the point of impact. The path of the incident photon is deflected by its interaction and is scattered from the site of the collision. The energy of the 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. As with photoelectric absorption, Compton scattering results in the loss of an electron and ionization of the absorbing atom. Scattered photons continue on their new paths, causing further ionizations. Similarly, the recoil electrons also give up their energy by ionizing other atoms.

The probability of a Compton interaction is directly proportional to the electron density of the absorber. The
number of electrons in bone \((5.55 \times 10^{23}/\text{cc})\) is greater than in soft tissue \((3.34 \times 10^{23}/\text{cc})\); therefore the probability of Compton scattering is correspondingly greater in bone than in tissue. In a dental x-ray beam, approximately 62% of the photons undergo Compton scattering.

Scattered photons travel in all directions. The higher the energy of the incident photon, however, the greater the probability that the angle of scatter of the secondary photon will be small and its direction will be forward. Approximately 30% of the scattered photons formed during a dental x-ray exposure (primarily from Compton scattering) exit through the patient’s head. This is advantageous to the patient because some of the energy of the incident x-ray beam escapes the tissue, but it is disadvantageous because it causes nonspecific film darkening. Scattered photons darken the film while carrying no useful information because their paths are altered.

**DIFFERENTIAL ABSORPTION**

The importance of photoelectric absorption and Compton scattering in diagnostic radiography relates to differences in the way photons are absorbed by various anatomic structures. The number of photoelectric and Compton interactions is greater in hard tissues than in soft tissues. As a consequence, more photons in the beam exit the patient after passing through soft tissue than through hard tissue. Thus while the *incident beam*, the beam striking the patient, is spatially homogenous, the *remnant beam*, the beam that exits the patient, is spatially heterogeneous. This remnant beam strikes the image receptor (film), resulting in greater exposure of the film behind soft tissue than behind hard tissues. It is this differential exposure of the film that allows a radiograph to reveal the morphology of enamel, dentin, bone, and soft tissues.

**SECONDARY ELECTRONS**

In both photoelectric absorption and Compton scattering, electrons are ejected from their orbits in the absorbing material after interaction with x-ray photons. These secondary electrons give up their energy in the absorber by either of two processes: (1) collisional interaction with other electrons, resulting in ionization or *excitation* of the affected atom, and (2) radiative interactions, which produce *bremsstrahlung* radiation, resulting in the emission of low-energy x-ray photons. Secondary electrons eventually dissipate all their energy, mostly as heat by collisional interactions, and come to rest.

**BEAM ATTENUATION**

As an x-ray beam travels through matter, its intensity is reduced (attenuated). This results from loss of individual photons, primarily through photoelectric absorption and Compton scattering interactions. The reduction of beam intensity is predictable because it depends on physical characteristics of the beam and absorber. A monochromatic beam of photons, a beam in which all the photons have the same energy, provides a good 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. Therefore 1.5 cm of water may reduce a beam intensity by 50%, the next 1.5 cm by another 50% (to 25% of the original intensity), and so on. This is an exponential pattern of absorption (Fig. 1-26). The HVL described earlier in this chapter is a measure of beam energy describing the amount of an absorber that reduces the beam intensity by half; in the

![FIG. 1-26 Exponential decay of intensity in a homogeneous photon beam through the absorber, where the HVL is 1.5 cm of absorber. The curve for a heterogeneous x-ray beam does not drop quite as precipitously because of the preferential removal of low-energy photons and the increased mean energy of the resulting beam.](#)
TABLE 1-2
Summary of Radiation Quantities and Units

<table>
<thead>
<tr>
<th>QUANTITY</th>
<th>SI UNIT</th>
<th>TRADITIONAL UNIT</th>
<th>CONVERSION</th>
</tr>
</thead>
<tbody>
<tr>
<td>Exposure</td>
<td>Air kerma (Gy)</td>
<td>Roentgen (R)</td>
<td>1 Gy = 100 rad</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>1 rad = 0.01 Gy (1 cGy)</td>
</tr>
<tr>
<td>Absorbed dose</td>
<td>Gray (Gy)</td>
<td>Rad</td>
<td>1 Gy = 100 rad</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>1 rad = 0.01 Gy (1 cGy)</td>
</tr>
<tr>
<td>Equivalent dose</td>
<td>Sievert (Sv)</td>
<td>Rem</td>
<td>1 Sv = 100 rem</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>1 rem = 0.01 Sv (1 cSv)</td>
</tr>
<tr>
<td>Effective dose</td>
<td>Sievert (Sv)</td>
<td>—</td>
<td>—</td>
</tr>
<tr>
<td>Radioactivity</td>
<td>Becquerel (Bq)</td>
<td>Curie (Ci)</td>
<td>1 Bq = 2.7 × 10^{-11} Ci</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>1 Ci = 3.7 × 10^{10} Bq</td>
</tr>
</tbody>
</table>

Data from The NIST Reference on Constants, Units, and Uncertainty: http://physics.nist.gov/cuu/Units/units.html.

The attenuation of a beam depends on both the energy of the incident beam and the composition of the absorber. In general, as the energy of the beam increases, so does the transmission of the beam through the absorber. When the energy of the incident photon is raised to the binding energy of the K-shell electrons of the absorber, however, the probability of photoelectric absorption increases sharply and the number of transmitted photons is greatly decreased. 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 K-shell electrons interact photoelectrically only with electrons in the L shell and in shells even farther from the nucleus. Rare earth elements are sometimes used as filters because their K edges (50.2 keV for gadolinium) greatly increase the absorption of high-energy photons. This is desirable because these high-energy photons are not as likely to contribute to a radiographic image as mid-energy photons.

**Dosimetry**

Determining the quantity of radiation exposure or dose is termed *dosimetry*. The term *dose* is used to describe the amount of energy absorbed per unit mass at a site of interest. *Exposure* is a measure of radiation based on its ability to produce ionization in air under standard conditions of temperature and pressure (STP).

**UNITS OF MEASUREMENT**

Table 1-2 presents some of the more frequently used units for measuring quantities of radiation. In recent years a move has occurred to use a modernized version of the metric system called the *SI system* (Système International d’Unités)*. This book uses SI units. The SI system uses *base units* including the kilogram (mass), the meter (length), the second (time), the amphere
Exposure is a measure of radiation quantity, the capacity of radiation to ionize air. The SI unit of exposure is air kerma, an acronym for kinetic energy released in matter. Kerma measures the kinetic energy transferred from photons to electrons and is expressed in units of dose (Gy), where 1 Gy equals 1 joule/kg. Kerma is the sum of the initial kinetic energies of all the charged particles liberated by uncharged ionizing radiation (neutrons and photons) in a sample of matter, divided by the mass of the sample. It has replaced the roentgen (R), the traditional unit of radiation exposure measured in air.

Absorbed Dose
Absorbed dose is a measure of the energy absorbed by any type of ionizing radiation per unit mass of any type of matter. The SI unit is the gray (Gy), where 1 Gy equals 1 joule/kg. The traditional unit of absorbed dose is the rad (radiation absorbed dose), where 1 rad is equivalent to 100 ergs/g of absorber. One gray equals 100 rads.

Equivalent Dose
The equivalent dose ($H_T$) is used to compare the biologic effects of different types of radiation to a tissue or organ. It is the sum of the products of the absorbed dose ($D_T$) averaged over a tissue or organ and the radiation weighting factor ($W_R$):

$$H_T = \sum W_R \times D_T$$

Equivalent dose is expressed as a sum to allow for the possibility that the tissue or organ is exposed to more than one type of radiation. The radiation weighting factor is chosen for the type and energy of the radiation involved. Thus high-LET radiations (which are more damaging to tissue than low-LET radiations) have a correspondingly higher $W_R$. For example, the $W_R$ of photons is 1; of 5 keV neutrons and high-energy protons, 5; and of alpha particles, 20. The unit of equivalent dose is the sievert (Sv). For diagnostic x-ray examinations 1 Sv equals 1 Gy. The traditional unit of equivalent dose is the rem (roentgen equivalent man). One sievert equals 100 rem.

Effective Dose
The effective dose ($E$) is used to estimate the risk in humans. It is the sum of the products of the equivalent dose to each organ or tissue ($H_T$) and the tissue weighting factor ($W_T$):

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

The tissue weighting factors include gonads, 0.20; red bone marrow, 0.12; esophagus, 0.05; thyroid, 0.05; skin, 0.01; and bone surface, 0.01. The unit of effective dose is the sievert (Sv). The use of this term is described more fully in Chapter 3.

Radioactivity
The measurement of radioactivity (A) describes the decay rate of a sample of radioactive material. The SI unit is the becquerel (Bq); 1 Bq equals 1 disintegration/second. The traditional unit is the curie (Ci), which corresponds to the activity of 1 g of radium ($3.7 \times 10^{10}$ disintegrations/second). Accordingly, 1 mCi equals 37 megaBq; and 1 Bq equals $2.7 \times 10^{-11}$ Ci.

BIBLIOGRAPHY