CHAPTER 6

X-ray Production, X-ray Tubes, and X-ray Generators

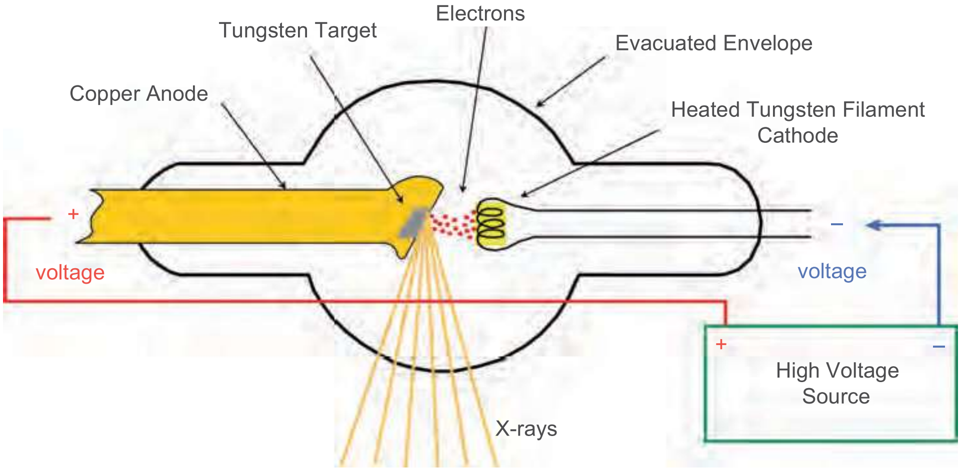
X-rays are produced when highly energetic electrons interact with matter, converting some of their kinetic energy into electromagnetic radiation. A device that produces x-rays in the diagnostic energy range typically contains an electron source, an evacuated path for electron acceleration, a target electrode, and an external power source to provide a high voltage (potential difference) to accelerate the electrons. Specifically, the *x-ray tube insert* contains the electron source and target within an evacuated glass or metal envelope; the *tube housing* provides protective radiation shielding and cools the x-ray tube insert; the *x-ray generator* supplies the voltage to accelerate the electrons; *x-ray beam filters* at the tube port shape the x-ray energy spectrum; and *collimators* define the size and shape of the x-ray field incident on the patient. The generator also permits control of the x-ray beam characteristics through the selection of voltage, current, and exposure time. These components work in concert to create a beam of x-ray photons of well-defined intensity, penetrability, and spatial distribution. In this chapter, the x-ray creation process, characteristics of the x-ray beam, and equipment components are discussed.

6.1 Production of X-rays

Bremsstrahlung Spectrum

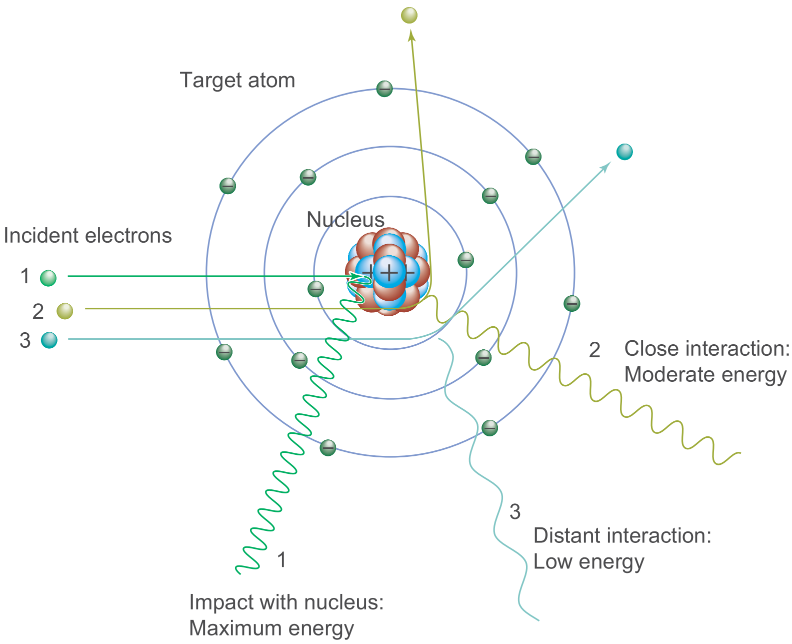
X-rays are created from the conversion of kinetic energy of electrons into electromagnetic radiation when they are decelerated by interaction with a target material. A simplified diagram of an x-ray tube (Fig. 6-1) illustrates these components. For diagnostic radiology, a large electric potential difference (the SI unit of potential difference is the volt, V) of 20,000 to 150,000 V (20 to 150 kV) is applied between two electrodes (the cathode and the anode) in the vacuum. The *cathode* is the *source* of electrons, and the *anode,* with a positive potential with respect to the cathode, is the *target* of electrons. As electrons from the cathode travel to the anode, they are accelerated by the voltage between the electrodes and attain kinetic energies equal to the product of the electrical charge and potential difference (see Appendix A). A common unit of energy is the electron volt (eV), equal to the energy attained by an electron accelerated across a potential difference of 1 V. Thus, the kinetic energy of an electron accelerated by a potential difference of 50 kV is 50 keV. One eV is a very small amount of energy, as there are 6.24×10^{18} eV/J.

On impact with the target, the kinetic energy of the electrons is converted to other forms of energy. The vast majority of interactions are *collisional,* whereby energy exchanges with electrons in the target give rise to heat. A small fraction of the accelerated electrons comes within the proximity of an atomic nucleus and is influenced by its positive electric field. As discussed in Chapter 3, electrical (Coulombic) forces attract and decelerate an electron and change its direction, causing a loss of kinetic energy, which is emitted as an x-ray photon of equal energy (i.e., bremsstrahlung radiation).



■ **FIGURE 6-1** Minimum requirements for x-ray production include a source and target of electrons, an evacuated envelope, and connection of the electrodes to a high-voltage source.

The amount of energy lost by the electron and thus the energy of the resulting x-ray are determined by the distance between the incident electron and the target nucleus, since the Coulombic force is proportional to the inverse of the square of the distance. At relatively large distances from the nucleus, the Coulombic attraction is weak; these encounters produce low x-ray energies (Fig. 6-2, electron no. 3). At closer interaction distances, the force acting on

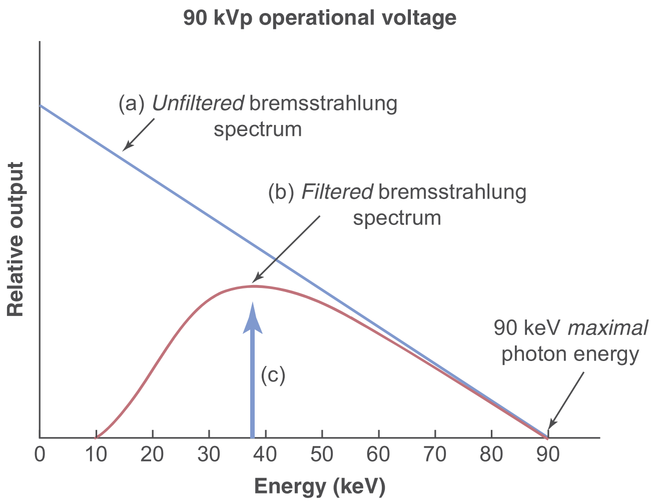


■ **FIGURE 6-2** Bremsstrahlung radiation arises from energetic electron interactions with an atomic nucleus of the target material. In a "close" approach, the positive nucleus attracts the negative electron, causing deceleration and redirection, resulting in a loss of kinetic energy that is converted to an x-ray. The x-ray energy depends on the interaction distance between the electron and the nucleus; it decreases as the distance increases.

the electron increases, causing a greater deceleration; these encounters produce higher x-ray energies (see Fig. 6-2, electron no. 2). A nearly direct impact of an electron with the target nucleus results in loss of nearly all of the electron's kinetic energy (see Fig. 6-2, electron no. 1). In this rare situation, the highest x-ray energies are produced.

The probability of electron interactions that result in production of x-ray energy E is dependent on the radial interaction distance, r, from the nucleus, which defines a circumference, $2\pi r$. With increasing distance from the nucleus, the circumference increases, and therefore the probability of interaction increases, but the x-ray energy decreases. Conversely, as the interaction distance, r, decreases, the x-ray energy increases because of greater electron deceleration, but the probability of interaction decreases. For the closest electron-atomic nuclei interactions, the highest x-ray energies are produced. However, the probability of such an interaction is very small, and the number of x-rays produced is correspondingly small. The number of x-rays produced decreases linearly with energy up to the maximal x-ray energy, which is equal to the energy of the incident electrons. A *bremsstrahlung spectrum* is the probability distribution of x-ray photons as a function of photon energy (keV). The *unfiltered* bremsstrahlung spectrum (Fig. 6-3A) shows an inverse linear relationship between the number and the energy of the x-rays produced, with the highest x-ray energy determined by the peak voltage (kV) applied across the x-ray tube. A typical *filtered* bremsstrahlung spectrum (Fig. 6-3B) has no x-rays below about 10 keV; the numbers increase to a maximum at about one third to one half the maximal x-ray energy and then decrease to zero as the x-ray energy increases to the maximal x-ray energy. Filtration in this context refers to the removal of x-rays by attenuation in materials that are inherent in the x-ray tube (e.g., the glass window of the tube insert), as well as by materials that are purposefully placed in the beam, such as thin aluminum and copper sheets, to remove lower energy x-rays and adjust the spectrum for optimal low-dose imaging (see Section 6.7).

Major factors that affect x-ray production efficiency include the atomic number of the target material and the kinetic energy of the incident electrons. The approximate ratio of radiative energy loss caused by bremsstrahlung production to



■ **FIGURE 6-3** The bremsstrahlung energy distribution for a 90-kV acceleration potential difference. The unfiltered bremsstrahlung spectrum (a) shows a greater probability of low-energy x-ray photon production that is inversely linear with energy up to the maximum energy of 90 keV. The filtered spectrum (b) shows the preferential attenuation of the lowest-energy x-ray photons. The vertical arrow (c) indicates the average energy of the spectrum, which is typically 1/3 to 1/2 the maximal energy.

collisional (excitation and ionization) energy loss within the diagnostic x-ray energy range (potential difference of 20 to 150 kV) is expressed as follows:

$$\frac{\text{Radiative energy loss}}{\text{Collisional energy loss}} \cong \frac{E_K Z}{820{,}000} \qquad [6\text{-}1]$$

where E_K is the kinetic energy of the incident electrons in keV, and Z is the atomic number of the target electrode material. The most common target material is tungsten (W, Z = 74); in mammography, molybdenum (Mo, Z = 42) and rhodium (Rh, Z = 45) are also used. For 100-keV electrons impinging on tungsten, the approximate ratio of radiative to collisional losses is $(100 \times 74)/820{,}000 \cong 0.009 \cong 0.9\%$; therefore, more than 99% of the incident electron energy on the target electrode is converted to heat and nonuseful low-energy electromagnetic radiation. At much higher electron energies produced by radiation therapy systems (millions of electron volts), the efficiency of x-ray production is dramatically increased. However, Equation 6-1 is not applicable beyond diagnostic imaging x-ray energies.

Characteristic X-ray Spectrum

In addition to the continuous bremsstrahlung x-ray spectrum, discrete x-ray energy peaks called "characteristic radiation" can be present, depending on the elemental composition of the target electrode and the applied x-ray tube voltage. Electrons in an atom are distributed in shells, each of which has an electron binding energy. The innermost shell is designated the K shell and has the highest electron binding energy, followed by the L, M, and N shells, with progressively less binding energy. Table 6-1 lists the common anode target materials and the corresponding binding energies of their *K*, *L*, and *M* electron shells. The electron binding energies are "characteristic" of the elements. When the energy of an incident electron, determined by the voltage applied to the x-ray tube, exceeds the binding energy of an electron shell in a target atom, a collisional interaction can eject an electron from its shell, creating a vacancy. As discussed in Chapter 2, an outer shell electron with less binding energy immediately transitions to fill the vacancy, and a characteristic x-ray is emitted with an energy equal to the difference in the electron binding energies of the two shells (Fig. 6-4).

For tungsten, an *L*-shell (binding energy = 10.2 keV) electron transition to fill a *K*-shell (binding energy = 69.5 keV) vacancy produces a characteristic x-ray with a discrete energy of

$$E_{bK} - E_{bL} = 69.5\ \text{keV} - 10.2\ \text{keV} = 59.3\ \text{keV}$$

Electron transitions occur from adjacent and nonadjacent electron shells in the atom, giving rise to several discrete characteristic energy peaks superimposed on the

TABLE 6-1 ELECTRON BINDING ENERGIES (KeV) OF COMMON X-RAY TUBE TARGET MATERIALS

ELECTRON SHELL	TUNGSTEN	MOLYBDENUM	RHODIUM
K	69.5	20.0	23.2
L	12.1/11.5/10.2	2.8/2.6/2.5	3.4/3.1/3.0
M	2.8–1.9	0.5–0.4	0.6–0.2

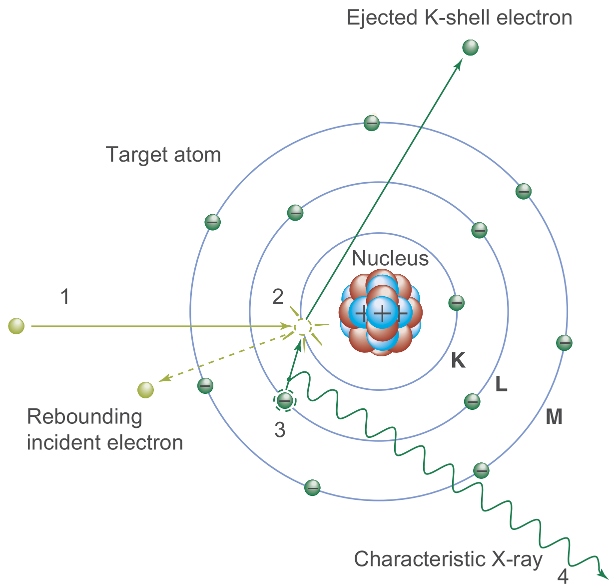


FIGURE 6-4 Generation of a characteristic x-ray in a target atom occurs in the following sequence: (1) The incident electron interacts with the *K*-shell electron via a repulsive electrical force. (2) The *K*-shell electron is removed (only if the energy of the incident electron is greater than the *K*-shell binding energy), leaving a vacancy in the *K*-shell. (3) An electron from the adjacent *L*-shell (or possibly a different shell) fills the vacancy. (4) A K_α characteristic x-ray photon is emitted with energy equal to the difference between the binding energies of the two shells. In this case, a 59.3-keV photon is emitted.

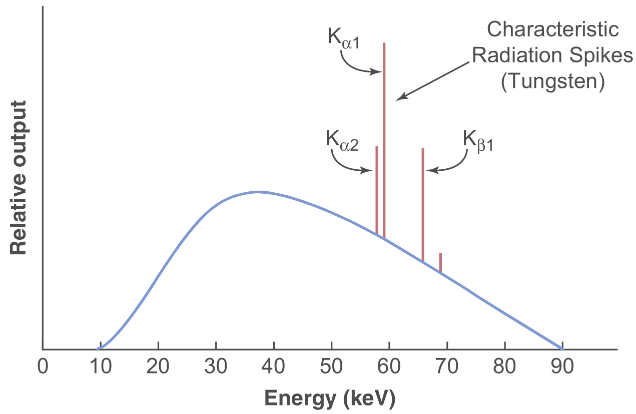
bremsstrahlung spectrum. Characteristic x-rays are designated by the shell in which the electron vacancy is filled, and a subscript of α or β indicates whether the electron transition is from an adjacent shell (α) or nonadjacent shell (β). For example, K_α refers to an electron transition from the *L* to the *K* shell, and K_β refers to an electron transition from the *M*, *N*, or *O* shell to the *K* shell. A K_β x-ray is more energetic than a K_α x-ray. Characteristic x-rays other than those generated by *K*-shell transitions are too low in energy for any useful contributions to the image formation process and are undesirable for diagnostic imaging. Table 6-2 lists electron shell binding energies and corresponding *K*-shell characteristic x-ray energies of W, Mo, and Rh anode targets.

Characteristic *K* x-rays are produced *only* when the electrons impinging on the target *exceed* the binding energy of a *K*-shell electron. x-Ray tube voltages must therefore be greater than 69.5 kV for W targets, 20 kV for Mo targets, and 23 kV for Rh targets to produce *K* characteristic x-rays. In terms of intensity, as the x-ray tube voltage increases, so does the ratio of characteristic to bremsstrahlung x-rays. For example, at 80 kV, approximately 5% of the total x-ray output intensity for a tungsten target is composed of characteristic radiation, which increases to about 10% at 100 kV. Figure 6-5 illustrates a bremsstrahlung plus characteristic radiation spectrum. In mammography, characteristic x-rays from Mo and Rh target x-ray tubes are particularly useful in optimizing image contrast and radiation dose (See Chapter 8 for further information).

TABLE 6-2 K-SHELL CHARACTERISTIC X-RAY ENERGIES (keV) OF COMMON X-RAY TUBE TARGET MATERIALS

SHELL TRANSITION	TUNGSTEN	MOLYBDENUM	RHODIUM
$K_{\alpha 1}$	59.32	17.48	20.22
$K_{\alpha 2}$	57.98	17.37	20.07
$K_{\beta 1}$	67.24	19.61	22.72

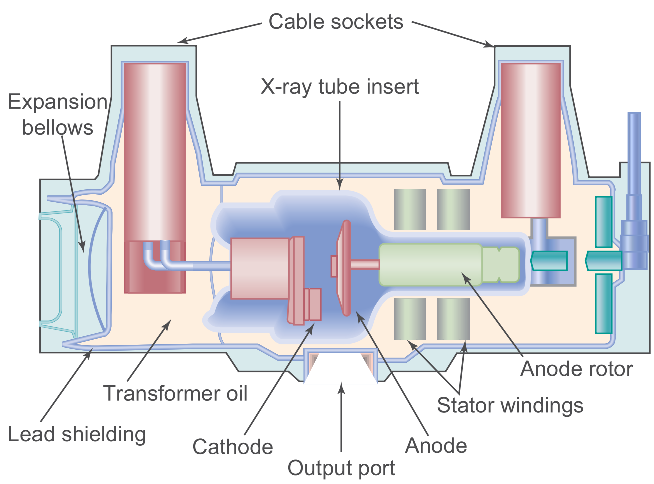
Note: Only prominent transitions are listed. The subscripts 1 and 2 represent energy levels that exist within each shell.



■ **FIGURE 6-5** The filtered spectrum of bremsstrahlung and characteristic radiation from a tungsten target with a potential difference of 90 kV illustrates specific characteristic radiation energies from K_α and K_β transitions. Filtration (the preferential removal of low-energy photons as they traverse matter) is discussed in Section 6.5.

6.2 X-ray Tubes

The x-ray tube provides an environment for the production of bremsstrahlung and characteristic x-rays. Major tube components are the *cathode, anode, rotor/stator, glass or metal envelope, tube port, cable sockets,* and *tube housing*, illustrated in Figure 6-6. An actual x-ray tube showing the x-ray tube insert and part of the housing is shown in Figure 6-7. The x-ray generator (Section 6.3) supplies the power and permits selection of tube voltage, tube current, and exposure time. Depending upon the type of imaging examination and the characteristics of the anatomy being imaged, the *x-ray tube voltage* is set to values from 40 to 150 kV for diagnostic imaging, and 25 to 40 kV for mammography. The *x-ray tube current*, measured in milliamperes (mA), is proportional to the number of electrons per second flowing from the cathode to the anode, where 1 mA = 6.24×10^{15} electrons/s. For continuous fluoroscopy, the tube current is relatively low, from 1 to 5 mA, and for projection radiography, the tube current is set from 50 to 1,200 mA in conjunction with short exposure times (typically less than 100 ms). (In pulsed fluoroscopy, the tube current is commonly delivered in short pulses instead of being continuous; the average tube current is typically in the range of 10 to 50 mA, while the overall number of electrons delivered through the tube is about the same per image.) The kV, mA, and exposure time are the three major selectable parameters on the x-ray generator control panel that



■ **FIGURE 6-6** A diagram of the major components of a modern x-ray tube and housing assembly is shown.

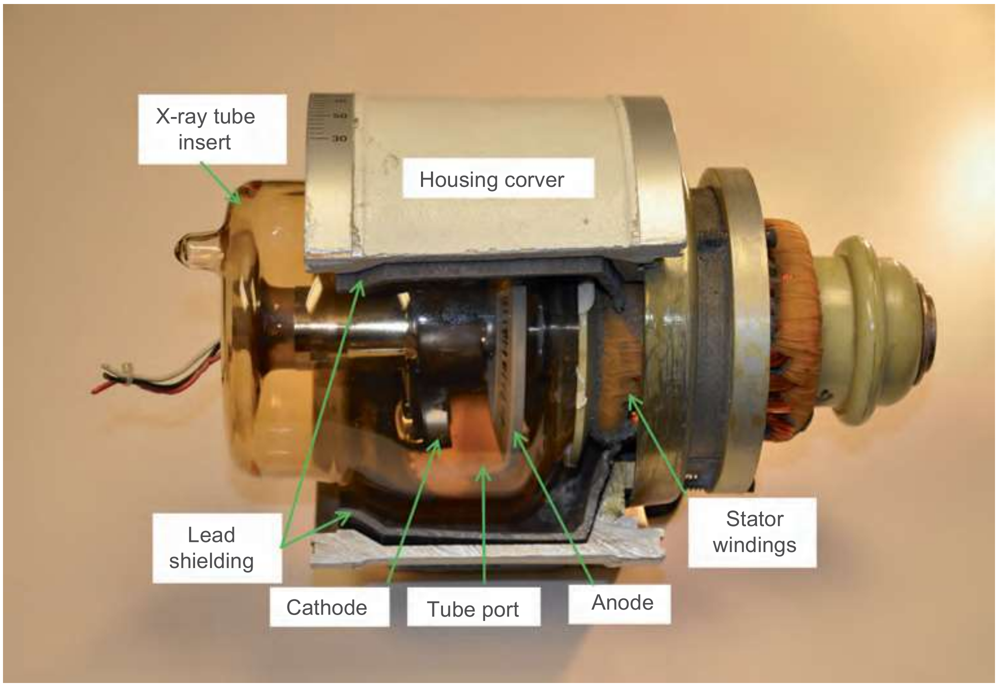


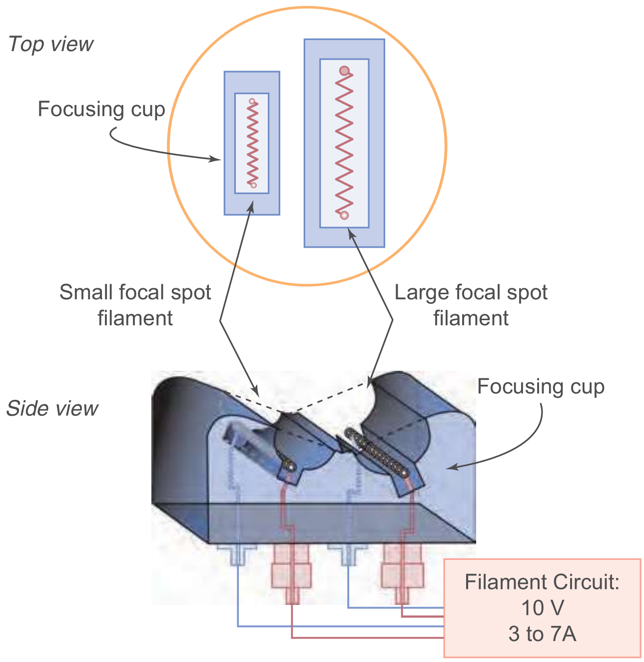
FIGURE 6-7 Picture of an x-ray tube insert and partially cut-away housing, shows the various components of the x-ray tube. For this housing, the lead shielding thickness is 2 mm.

determine the x-ray beam characteristics. Often, the product of the tube current and exposure time are considered as one entity, the mAs (milliampere-second; technically, mAs is a product of two units but, in common usage, it serves as a quantity). These parameters are discussed further in the following sections.

Cathode

The cathode is the negative electrode in the x-ray tube, comprised of a *filament* or filaments and a *focusing cup* (Fig. 6-8). A filament is made of tungsten wire wound in a helix, and is electrically connected to the filament circuit, which provides a voltage of approximately 10 V and variable current up to 7,000 mA (7 A). Most x-ray tubes for diagnostic imaging have two filaments of different lengths, each positioned in a slot machined into the focusing cup, with one end directly connected to the focusing cup, and the other end electrically insulated from the cup by a ceramic insert. Only one filament is energized for an imaging examination. On many x-ray systems, the small or the large filament can be manually selected, or automatically selected by the x-ray generator depending on the technique factors (kV and mAs).

When energized, the filament circuit heats the filament through electrical resistance, and the process of *thermionic emission* releases electrons from the filament surface at a rate determined by the filament current and corresponding filament temperature. Heat generated by resistance to electron flow in the filament raises the temperature to a point where electrons can leave the surface. However, electrons flow from the cathode to the anode *only when the tube voltage is applied between these electrodes*. The numbers of electrons that are available are adjusted by the filament current and filament temperature, as shown in Figure 6-9, where small changes in the filament current can produce relatively large changes in tube current. Output

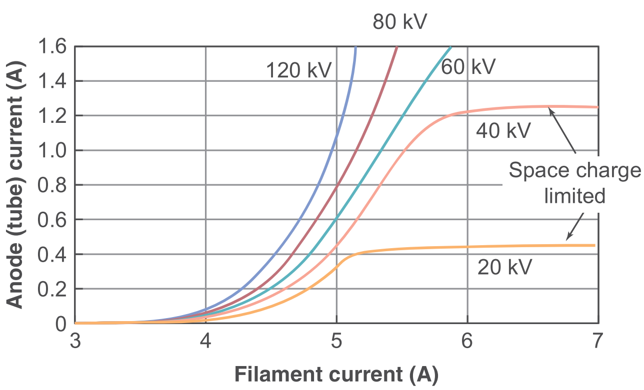


■ **FIGURE 6-8** The x-ray tube cathode structure consists of the filaments and the focusing (or cathode) cup. Current from the filament circuit heats a filament, which releases electrons by thermionic emission.

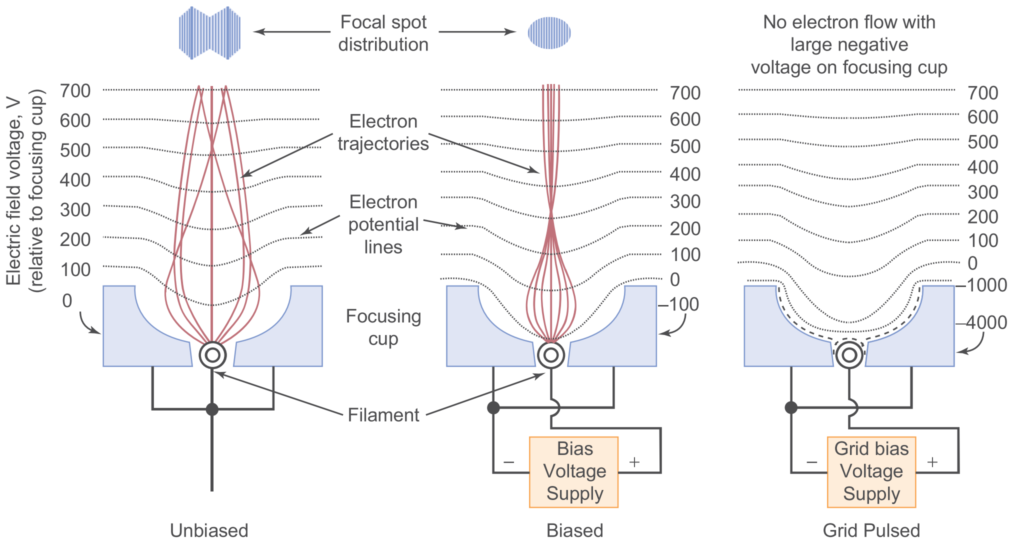
of the x-ray tube is *emission-limited*, meaning that the filament current determines the x-ray tube current; at any kV, the x-ray flux is proportional to the tube current. Higher tube voltages also produce slightly higher tube current for the same filament current. A filament current of 5 A at a tube voltage of 80 kV produces a tube current of about 800 mA, whereas the same filament current at 120 kV produces a tube current of about 1,100 mA.

In most x-ray tubes, the focusing cup is maintained at the same potential difference as the filament relative to the anode, and at the edge of the slot, an electric field exists that repels and shapes the cloud of emitted electrons from the filament surface. As a large voltage is applied between the cathode and anode in the correct polarity, electrons are accelerated into a tight distribution and travel to the anode, striking a small area called the focal spot (Fig. 6-10). The focal spot dimensions are determined by the length of the filament in one direction and the width of electron distribution in the perpendicular direction.

A *biased* x-ray tube has a focusing cup totally insulated from the filament wires so that its voltage is independent of the filament. Because high voltages are applied



■ **FIGURE 6-9** Relationship of tube current to filament current for various tube voltages shows a dependence of approximately $kV^{1.5}$. For tube voltages 40 kV and lower, a space charge cloud shields the electric field so that further increases in filament current do not increase the tube current. This is known as "space charge–limited" operation. Above 40 kV, the filament current limits the tube current; this is known as "emission-limited" operation.



■ **FIGURE 6-10** The focusing cup shapes the electron distribution when it is at the same voltage as the filament (**left**). Isolation of the focusing cup from the filament and application of a negative bias voltage (~ –100 V) reduces electron distribution further by increasing the repelling electric fields surrounding the filament and modifying the electron trajectories (**middle**). At the top are typical electron distributions incident on the target anode (the focal spot) for the unbiased and biased focusing cups. Application of –4,000 V on an isolated focusing cup completely stops electron flow, even with high voltage applied on the tube; this is known as a grid biased or grid pulsed tube (**right**).

to the cathode, electrical insulation of the focusing cup and the bias supply voltage is necessary, and can add significant expense to the x-ray system. A voltage of about 100 V negative is applied with respect to the filament voltage to further reduce the spread of electrons and produce a smaller focal spot width (Fig. 6-10 middle). Even greater negative applied voltage (about –4,000 V) to the focusing cup actually stops the flow of electrons, providing a means to rapidly switch the x-ray beam on and off (Fig. 6-10 right); a tube with this capability is referred to as a *grid-biased* x-ray tube. Grid-biased x-ray tube switching is used by more expensive fluoroscopy systems for pulsed fluoroscopy and angiography to rapidly and precisely turn on and turn off the x-ray beam. This eliminates the build-up delay and turnoff lag of x-ray tubes switched at the generator, which cause motion artifacts and produce lower average x-ray energies and unnecessary patient dose.

Ideally, a focal spot would be a point, eliminating geometric blurring. However, such a focal spot is not possible and, if it were, would permit only a tiny tube current. In practice, a finite focal spot area is used with an area large enough to permit a sufficiently large tube current and short exposure time. For magnification studies, a small focal spot is necessary to limit geometric blurring and achieve adequate spatial resolution (see Figure 6-16 and Chapter 7 on magnification).

Anode

The anode is a metal target electrode that is maintained at a large positive potential difference relative to the cathode. Electrons striking the anode deposit most of their energy as heat, with only a small fraction emitted as x-rays. Consequently, the production of x-rays, in quantities necessary for acceptable image quality, generates a large amount of heat in the anode. To avoid heat damage to the x-ray tube, the rate of x-ray production (proportional to the tube current) and, at large tube currents, the duration of x-ray production, must be limited. Tungsten (W, Z = 74) is the most widely used anode material because of its high melting point and high atomic number. A tungsten anode can handle substantial heat deposition without cracking or pitting of its surface. An alloy of 10% rhenium and 90% tungsten provides added resistance to surface damage. Tungsten provides greater bremsstrahlung production than elements with lower atomic numbers (Equation 6-1).

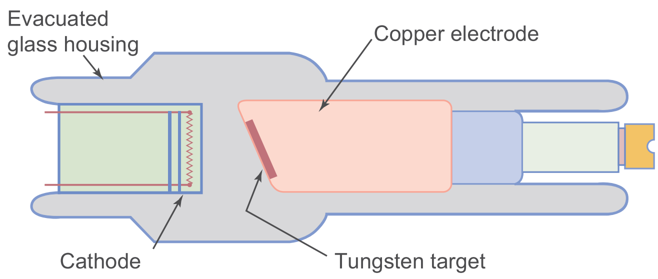
Molybdenum (Mo, Z = 42) and rhodium (Rh, Z = 45) are used as anode materials in mammographic x-ray tubes. These materials provide useful characteristic x-rays for breast imaging (see Table 6-2). Mammographic tubes are described further in Chapter 8.

Anode Configurations: Stationary and Rotating

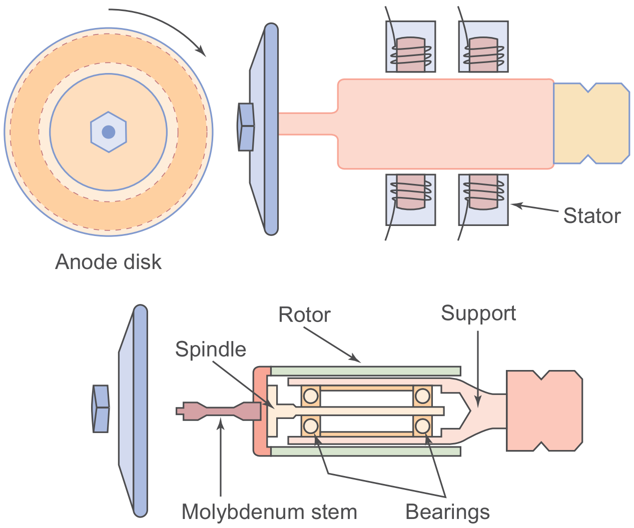
A simple x-ray tube design has a stationary anode, consisting of a tungsten insert embedded in a copper block (Fig. 6-11). Copper serves a dual role: it mechanically supports the insert and efficiently conducts heat from the tungsten target. However, the small area of the focal spot on the stationary anode limits the tube current and x-ray output that can be sustained without damage from excessive temperature. Dental x-ray units and some low-output mobile x-ray machines and mobile fluoroscopy systems use fixed anode x-ray tubes.

Rotating anodes are used for most diagnostic x-ray applications, mainly because of greater heat loading and higher x-ray intensity output. This design spreads the heat over a much larger area than does the stationary anode design, permitting much larger tube currents and exposure durations. The anode is a beveled disk mounted on a *rotor* assembly supported by bearings in the x-ray tube insert (Fig. 6-12). The rotor consists of copper bars arranged around a cylindrical iron core. A donut-shaped *stator* device, comprised of electromagnets, surrounds the rotor and is mounted outside of the x-ray tube insert. Alternating current (AC), the periodic reversal of electron movement in a conductor, passes through the stator windings and produces a rotating magnetic field (see electromagnetic induction, Section 6.3). This induces an electrical current in the rotor's copper bars, which creates an opposing magnetic field that causes it to spin. Rotation speeds are 3,000 to 3,600 (low speed) or 9,000 to 10,000 (high speed) revolutions per minute (rpm). X-ray systems are designed such that the x-ray tube will not be energized if the anode is not at full speed; this is the cause for the short delay (1 to 2 s) when the x-ray tube exposure button is pushed.

Rotor bearings are heat sensitive and are often the cause of x-ray tube failure. Bearings require special heat insensitive, nonvolatile lubricants because of the vacuum inside the x-ray tube insert and also require thermal insulation from the anode, achieved by using a molybdenum (a metal with poor heat conductivity) stem attaching the anode to the rotor. Most rotating anodes are cooled by infrared radiation emission, transferring heat to the x-ray tube insert and to the surrounding oil bath



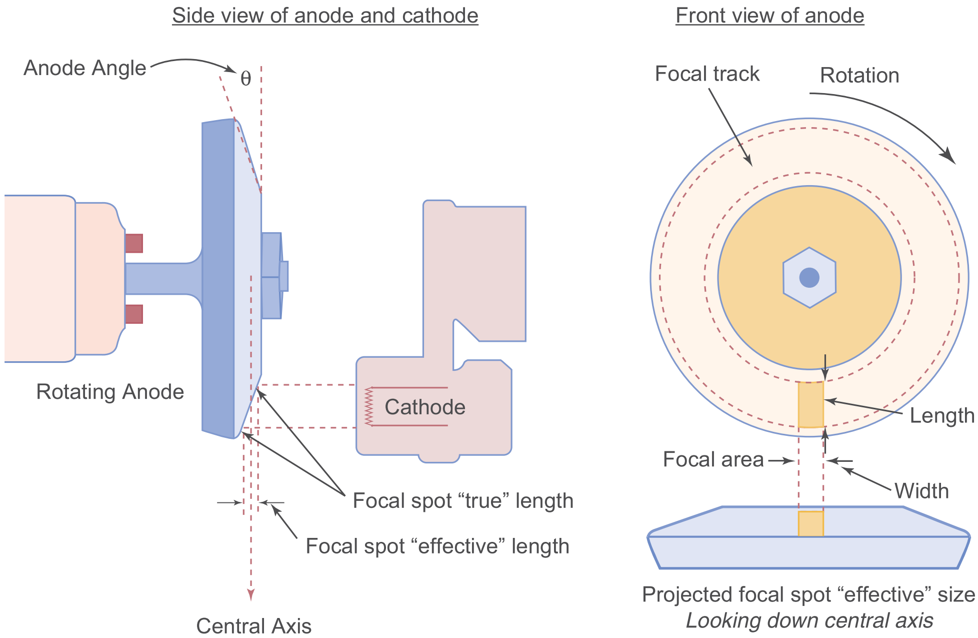
■ **FIGURE 6-11** The anode of a fixed anode x-ray tube consists of a tungsten insert mounted in a copper block. Heat is removed from the tungsten target by conduction into the copper block.



■ **FIGURE 6-12** The anode of a rotating anode x-ray tube is a tungsten disk mounted on a bearing-supported rotor assembly (front view, **top left**; side view, **top right**). The rotor consists of a copper and iron laminated core and forms part of an induction motor. The other component is the stator, which exists outside of the insert, **top right**. A molybdenum stem (molybdenum is a poor heat conductor) connects the rotor to the anode to reduce heat transfer to the rotor bearings (**bottom**).

and tube housing. In imaging situations demanding higher heat loads and more rapid cooling, such as interventional fluoroscopy and computed tomography (CT), sophisticated designs with externally mounted bearings and oil or water heat exchangers are employed (see special x-ray tube designs in this section).

The focal track area of the rotating anode is approximately equal to the product of the circumferential track length ($2\pi r$) and the track width (Δr), where r is the radial distance from the axis of the x-ray tube to the center of the track (Fig. 6-13). Thus, a rotating anode with a 5-cm focal track radius and a 1-mm track width provides a focal track with an annular area 314 times greater than that of a fixed anode with a focal spot area of 1×1 mm. The allowable instantaneous heat loading depends on



■ **FIGURE 6-13** The anode (target) angle, θ, is defined as the angle of the target surface in relation to the central ray. The focal spot length, as projected down the central axis, is foreshortened, according to the line focus principle (**lower right**).

the anode rotation speed and the focal spot area. Faster rotation distributes the heat load over a greater portion of the focal track area for short exposure times. A larger focal spot allows a greater x-ray beam intensity but causes a loss of spatial resolution that increases with distance of the imaged object from the image receptor. A large focal spot, which permits high x-ray output and short exposure times, should be used in situations where motion is expected to be a problem and geometric magnification is small (the object is close to the image receptor).

Anode Angle, Field Coverage, and Focal Spot Size

The anode angle is defined as the angle of the target surface with respect to the central ray (central axis) in the x-ray field (Fig. 6-13, left diagram). Anode angles in diagnostic x-ray tubes typically range from 7 to 20 degrees, with 12- to 15-degree angles being most common. Major factors affected by the anode angle include the effective focal spot size, tube output intensity, and x-ray field coverage provided at a given focal spot to detector distance.

The actual focal spot size is the area on the anode that is struck by electrons, and is primarily determined by the length of the cathode filament and the width of the focusing cup slot. However, the projected length of the focal spot area at the x-ray field central ray is much smaller, because of geometric foreshortening of the distribution from the anode surface. Thus, the effective and actual focal spot lengths are geometrically related as

$$\text{Effective focal length} = \text{Actual focal length} \times \sin\theta \qquad [6\text{-}2]$$

where θ is the anode angle. Foreshortening of the focal spot length at the central ray is called the *line focus principle*, as described by Equation 6-2. An ability to have a smaller effective focal spot size for a large actual focal spot increases the power loadings for smaller effective focal spot sizes.

EXAMPLE 1: The actual anode focal area for a 20-degree anode angle is 4 mm (length) by 1.2 mm (width). What is the projected focal spot size at the central axis position?

Answer: Effective length = actual length × sin θ = 4 mm × sin 20 degrees = 4 mm × 0.34 = 1.36 mm; therefore, the projected focal spot size is 1.36 mm (length) by 1.2 mm (width).

EXAMPLE 2: If the anode angle in Example 1 is reduced to 10 degrees and the actual focal spot size remains the same, what is the projected focal spot size at the central axis position?

Answer: Effective length = 4 mm × sin 10 degrees = 4 mm × 0.174 = 0.69 mm; thus, the smaller anode angle results in a projected size of 0.69 mm (length) by 1.2 mm (width) for the same actual target area.

As the anode angle decreases (approaches 0 degrees), the *effective* focal spot becomes smaller for the same *actual* focal area, providing better spatial resolution of the object when there is geometric image magnification. Also, for larger actual focal areas, greater x-ray output intensity with shorter exposure times is possible. However, a small anode angle limits the usable x-ray size at a given source to image receptor distance, because of cutoff of the beam on the anode side of the beam. *Field coverage* is also less for short focus-to-detector distances (Fig. 6-14). Therefore, the optimal anode angle depends on the clinical imaging application. A small anode angle (~7 to 9 degrees) is desirable for small field-of-view devices, such as some small fluoroscopy detectors, where field coverage is limited by the image receptor diameter

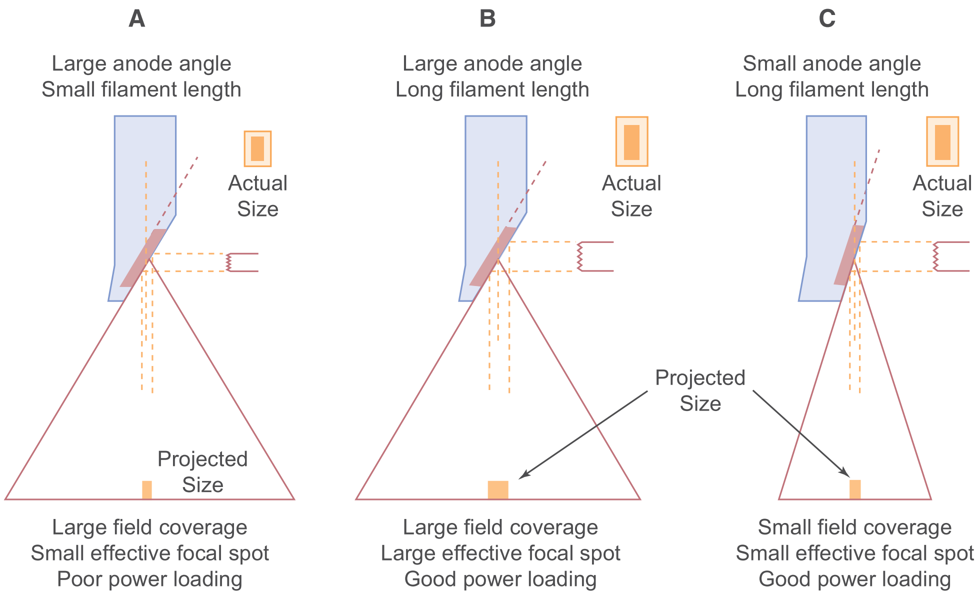
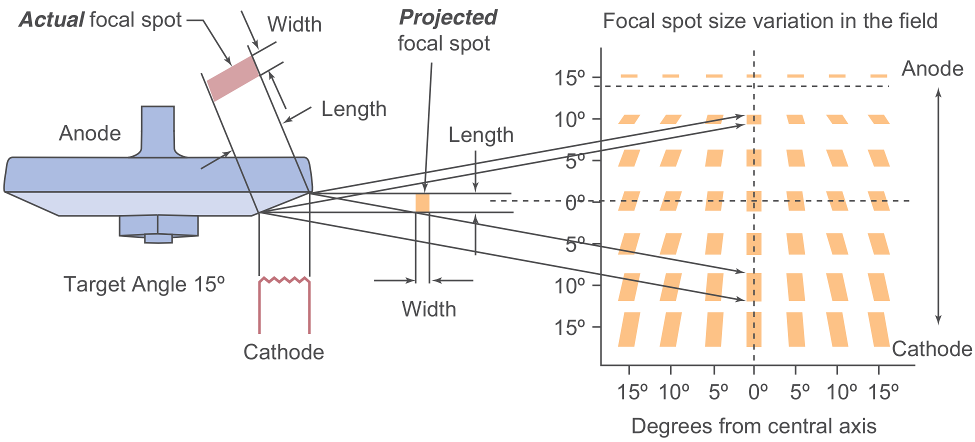


FIGURE 6-14 Field coverage and effective focal spot length vary with the anode angle. **A.** A large anode angle provides good field coverage at a given distance; however, to achieve a small effective focal spot, a small actual focal area limits power loading. **B.** A large anode angle provides good field coverage, and achievement of high power loading requires a large focal area; however, geometric blurring and image degradation occur. **C.** A small anode angle limits field coverage at a given distance; however, a small effective focal spot is achieved with a large focal area for high power loading.

(e.g., 23 cm). Larger anode angles (~12 to 15 degrees) are necessary for general radiographic imaging to achieve sufficiently large field area coverage at typical focal spot-to-detector distances such as 100 cm.

The effective focal spot length varies with the position in the image plane, in the anode-cathode (A–C) direction. Toward the anode side of the field, the projected length of the focal spot shortens, whereas it lengthens towards the cathode side of the field (Fig. 6-15). The width of the focal spot does not change appreciably with position in the image plane. Nominal focal spot size (width and length) is specified at the central ray of the beam, from the focal spot to the image receptor, perpendicular to the anode-cathode axis and bisecting the plane of the image receptor. x-Ray mammography is an exception, where "half-field" geometry is employed, as explained in Chapter 8.

Measurement and verification of focal spot size can be performed in several ways. Common tools for measuring focal spot size are the pinhole camera, slit camera, star pattern, and resolution bar pattern (Fig. 6-16). The *pinhole camera* uses a very small circular aperture (10 to 30 μm diameter) in a thin, highly attenuating metal (e.g., lead, tungsten, or gold) disk to project a magnified image of the focal spot onto an image receptor. With the pinhole camera positioned on the central axis between the x-ray source and detector, an image of the focal spot is recorded. Figure 6-16E shows magnified (2×) pinhole pictures of the large (top row) and small (bottom row) focal spots with a typical "bi-gaussian" intensity distribution. Correcting for the known image magnification allows measurement of the focal spot dimensions. The *slit camera* consists of a highly attenuating metal (usually tungsten) plate with a thin slit, typically 10 μm wide. In use, the slit camera is positioned above the image receptor, with the center of the slit on the central axis, and with the slit either parallel or perpendicular to the A-C axis. Measuring the width of the x-ray distribution in the image and correcting for magnification yields one dimension of



■ **FIGURE 6-15** Variation of the effective focal spot size in the image field occurs along the anode-cathode direction. Focal spot distributions are plotted as a function of projection angle in degrees from the central axis, the parallel (vertical axis), and the perpendicular (horizontal axis).

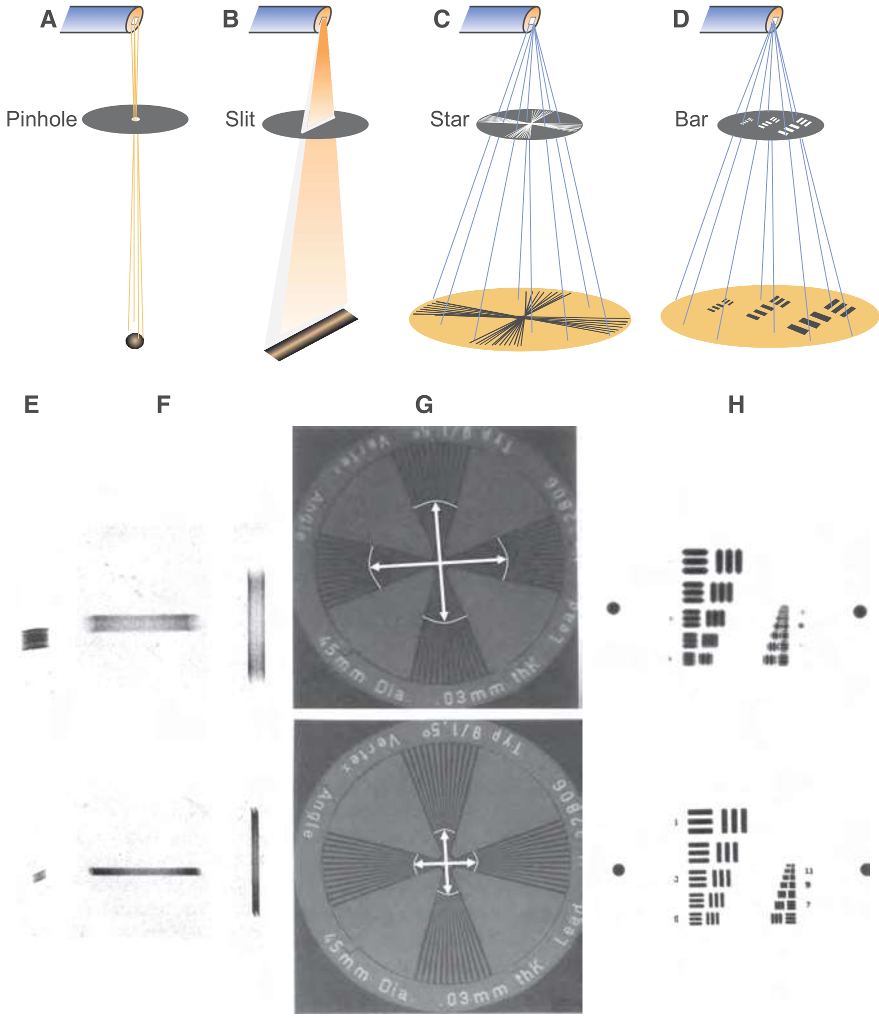
the focal spot. A second radiograph, taken with the slit perpendicular to the first, yields the other dimension of the focal spot, as shown in Figure 6-16F. The *star pattern* test tool (Fig. 6-16G) contains a radial pattern of lead spokes of diminishing width and spacing on a thin plastic disk. Imaging the star pattern at a known magnification and measuring the distance between the outermost blur patterns (location of the outermost unresolved spokes as shown by the arrows) on the image allows the calculation of the effective focal spot dimensions in the directions perpendicular and parallel to the A-C axis. A large focal spot will have a greater blur diameter than a small focal spot, as shown in the figure. A resolution bar pattern is a simple tool for evaluation of focal spot size (Fig. 6-16H). Bar pattern images demonstrate the effective resolution parallel and perpendicular to the A-C axis for a given magnification geometry, determined from the number of the bar pattern that can be resolved.

The National Electrical Manufacturers Association (NEMA) has published tolerances for measured focal spot sizes. For focal spot nominal (indicated) sizes less than 0.8 mm, the measured focal spot size can be larger by 50% (e.g., for a 0.6-mm focal spot, the measured size can be up to 0.9 mm), but not smaller than the nominal size. For focal spots between 0.8 and 1.5 mm nominal size, the measured focal spot size can be 0% smaller to 40% larger; and for focal spots greater than 1.5 mm, 0% smaller to 30% larger.

Focal spot "blooming" is an increase in the size of the focal spot resulting from high tube current (mA), and is caused by electron repulsion in the electron beam between the cathode and anode. It is most pronounced at low kVs. Focal spot "thinning" is a slight decrease in the measured size with increasing kV (electron repulsion and spreading in the x-ray tube is reduced). NEMA standards require measurement at 75 kV using 50% of the maximal rated mA for each focal spot.

Heel Effect

The *heel effect* refers to a reduction in the x-ray beam intensity toward the anode side of the x-ray field (Figure 6-17), caused by the greater attenuation of x-rays directed toward the anode side of the field by the anode itself. The heel effect is less prominent with a longer source-to-image distance (SID). Since the x-ray beam intensity is greater on the cathode side of the field, the orientation of the x-ray tube cathode over thicker parts of the patient can result in a better balance of x-ray photons transmitted

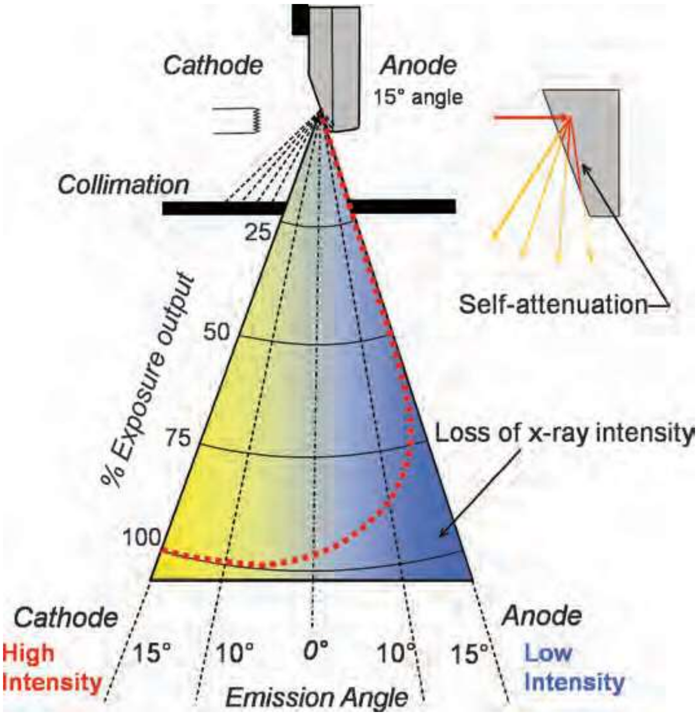


■ **FIGURE 6-16** Various tools allow measurement of the focal spot size, either directly or indirectly. **A** and **E**: Pinhole camera and images. **B** and **F**: Slit camera and images. **C** and **G**: Star pattern and images. **D** and **H**: Resolution bar pattern and images. For **E–H**, the top row of images represents the measurements of the large focal spot (1.2 mm × 1.2 mm), and the bottom row the small focal spot (0.6 mm × 0.6 mm). The star and bar patterns provide an "equivalent" focal spot dimension based upon the resolvability of the equivalent spatial frequencies.

through the patient and onto the image receptor. For example, the preferred orientation of the x-ray tube for a chest x-ray of a standing patient would be with the A-C axis vertical, and the cathode end of the x-ray tube down.

Off-Focal Radiation

Off-focal radiation results from electrons that scatter from the anode, and are re-accelerated back to the anode, outside of the focal spot area. These electrons cause low-intensity x-ray emission over the entire face of the anode, as shown in Figure 6-18, increasing patient exposure, causing geometric blurring, reducing image contrast, and increasing random noise. A small lead collimator aperture placed near the x-ray tube output port can reduce off-focal radiation by intercepting x-rays that are produced away from the focal spot. An x-ray tube that has a metal enclosure and the anode at electrical ground potential will have less off-focal radiation, because many of the scattered electrons are attracted to the metal envelope instead of the anode.

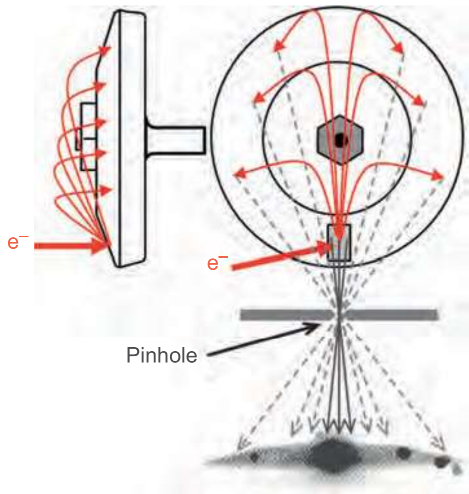


■ **FIGURE 6-17** The heel effect is a loss of intensity on the anode side of the x-ray field of view. It is caused by attenuation of the x-ray beam by the anode. Upper right is an expanded view that shows electrons interacting at depth within the anode and the resultant "self attenuation" of produced x-rays that have a trajectory towards the anode side of the field.

X-ray Tube Insert

The *x-ray tube insert* contains the cathode, anode, rotor assembly, and support structures sealed in a glass or metal enclosure under a high vacuum. The high vacuum prevents electrons from colliding with gas molecules and is necessary in most electron beam devices. As x-ray tubes age, trapped gas molecules percolate from

■ **FIGURE 6-18** Off-focal radiation is produced from back-scattered electrons that are re-accelerated to the anode outside the focal spot. This causes a low-intensity, widespread radiation distribution pattern. Hotspots outside the focal spot indicate areas where the electrons are more likely to interact.



tube structures and degrade the vacuum. A "*getter*" circuit is used to trap gas in the insert and to maintain the vacuum.

X-rays are emitted in all directions from the focal spot; however, the x-rays that emerge through the *tube port* constitute the useful beam. Except for mammography and special-purpose x-ray tubes, the port is typically made of the same material as the tube enclosure. Mammography tubes use beryllium (Z = 4) in the port to minimize absorption of the low-energy x-rays used in mammography.

X-ray Tube Housing

The x-ray tube housing supports, insulates, and protects the x-ray tube insert from the environment. Between the x-ray tube insert and housing is oil that provides heat conduction and electrical insulation. In many radiographic x-ray tubes, an expanding bellows inside the housing accommodates oil expansion due to heat absorption during operation. If the oil heats excessively, a microswitch disables the operation of the x-ray tube until sufficient cooling has occurred. X-ray tubes used in interventional fluoroscopy and CT commonly have heat exchangers to allow prolonged operation at high output.

Lead shielding inside the housing attenuates nearly all x-rays that are not directed to the tube port (see Fig. 6-7 for the typical lead sheet thickness and location within the housing). A small fraction of these x-rays, known as leakage radiation, penetrates the housing. Federal regulations (21 CFR 1020.30) require manufacturers to provide sufficient shielding to limit the leakage radiation exposure rate to 0.88 mGy air kerma per hour (equivalent to 100 mR/h) at 1 m from the focal spot when the x-ray tube is operated at the leakage technique factors for the x-ray tube. Leakage techniques are the maximal operable kV (kV_{max}, typically 125 to 150 kV) at the highest possible continuous current (typically 3 to 5 mA at kV_{max} for most diagnostic tubes). Each x-ray tube housing assembly has a maximal rated tube potential that must not be exceeded during clinical operation of the x-ray tube source assembly. The x-ray equipment is designed to prevent the selection of x-ray tube kV greater than the maximal rating.

Collimators

Collimators adjust the size and shape of the x-ray field emerging from the tube port. The collimator assembly typically is attached to the tube housing at the tube port with a swivel joint. Two pairs of adjustable parallel-opposed lead shutters define a rectangular x-ray field (Fig. 6-19). In the collimator housing, a beam of light reflected by a mirror of low x-ray attenuation mimics the x-ray beam. Thus, the collimation of the x-ray field is identified by the collimator's shadows. Federal regulations (21 CFR 1020.31) require that the light field and x-ray field be aligned so that the sum of the misalignments, along either the length or the width of the field, is within 2% of the SID. For example, at a typical SID of 100 cm (40 inches), the sum of the misalignments between the light field and the x-ray field at the left and right edges must not exceed 2 cm, and the sum of the misalignments at the other two edges also must not exceed 2 cm.

Positive beam limitation (PBL) collimators automatically limit the field size to the useful area of the detector. Mechanical sensors in the film cassette holder detect the cassette size and location and automatically adjust the collimator blades so that the x-ray field matches the cassette dimensions. Adjustment to a smaller field area is possible; however, a larger field area requires disabling the PBL circuit.