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 or all 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).

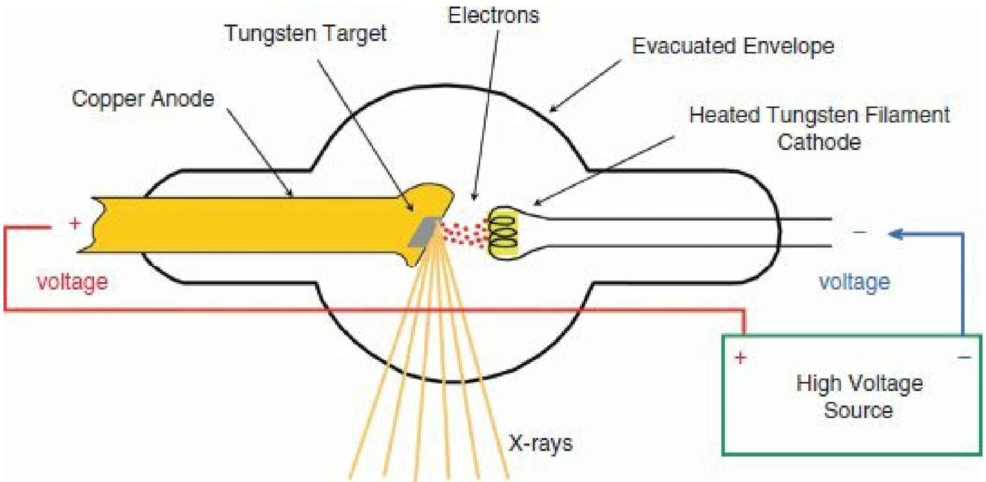


FIGURE 6-1 Minimum requirements for x-ray production include a source and target of electrons, an evacuated envelope, and connection of the electrodes to a high-voltage source.

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 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.

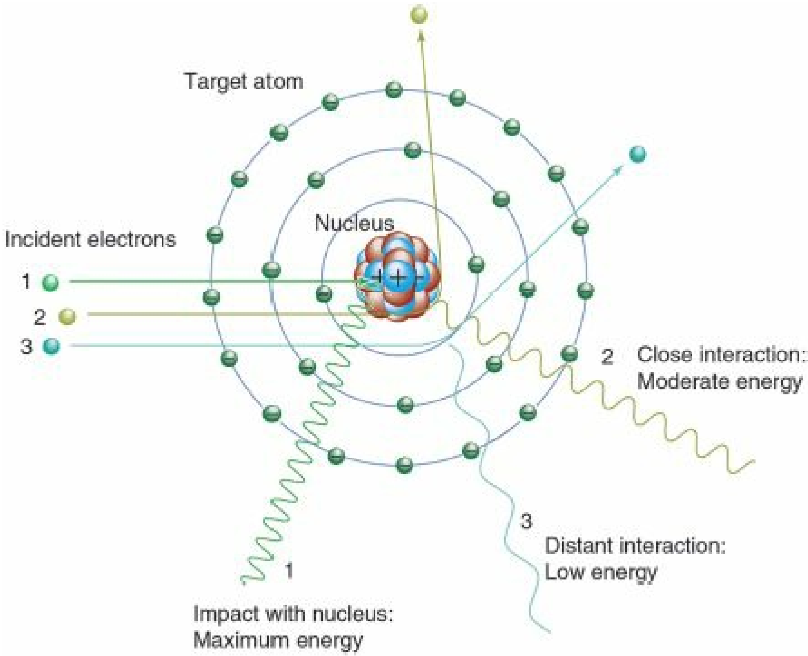


FIGURE 6-2 Bremsstrahlung radiation arises from energetic electron interactions with an atomic nucleus of the target material. In a "close" approach, the positive nucleus attracts the negative electron, causing deceleration and redirection, resulting in a loss of kinetic energy that is converted to an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

The 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 6.5).

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 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.

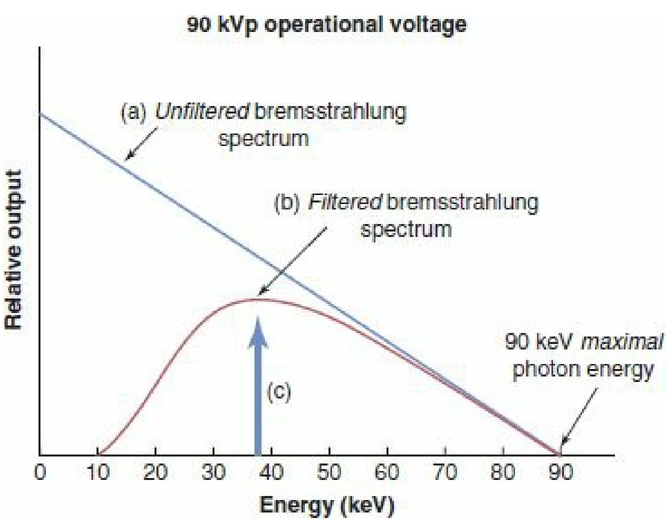


FIGURE 6-3 The bremsstrahlung energy distribution for a 90-kV acceleration potential difference. The unfiltered bremsstrahlung spectrum (a) shows a greater probability of low-energy x-ray photon production that is inversely linear with energy up to the maximum energy of 90 keV. The filtered spectrum (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.

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 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.

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

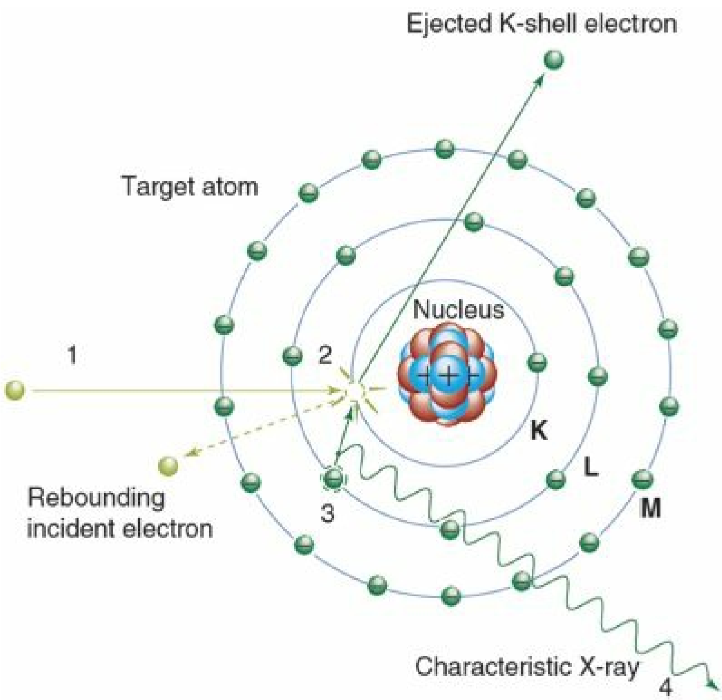


FIGURE 6-4 Generation of a characteristic x-ray in a target atom occurs in the following sequence: (1) The incident electron interacts with the *K*-shell electron via a repulsive electrical force. (2) The *K*-shell electron is removed (only if the energy of the incident electron is greater than the *K*-shell 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.

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.

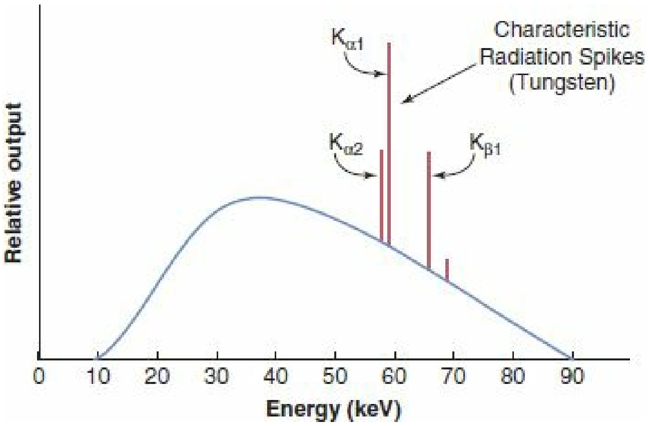
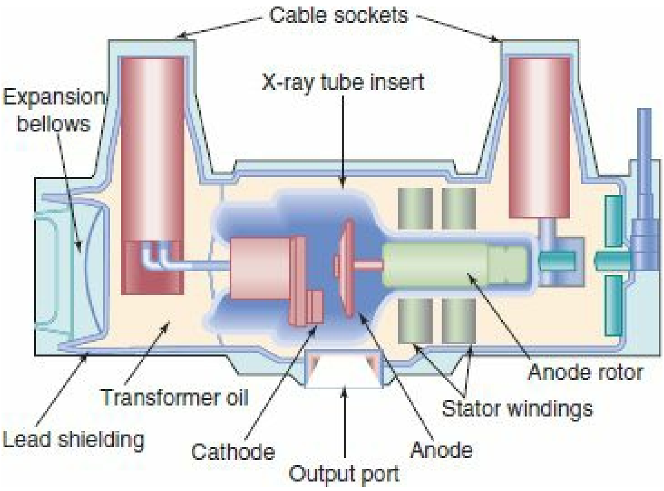


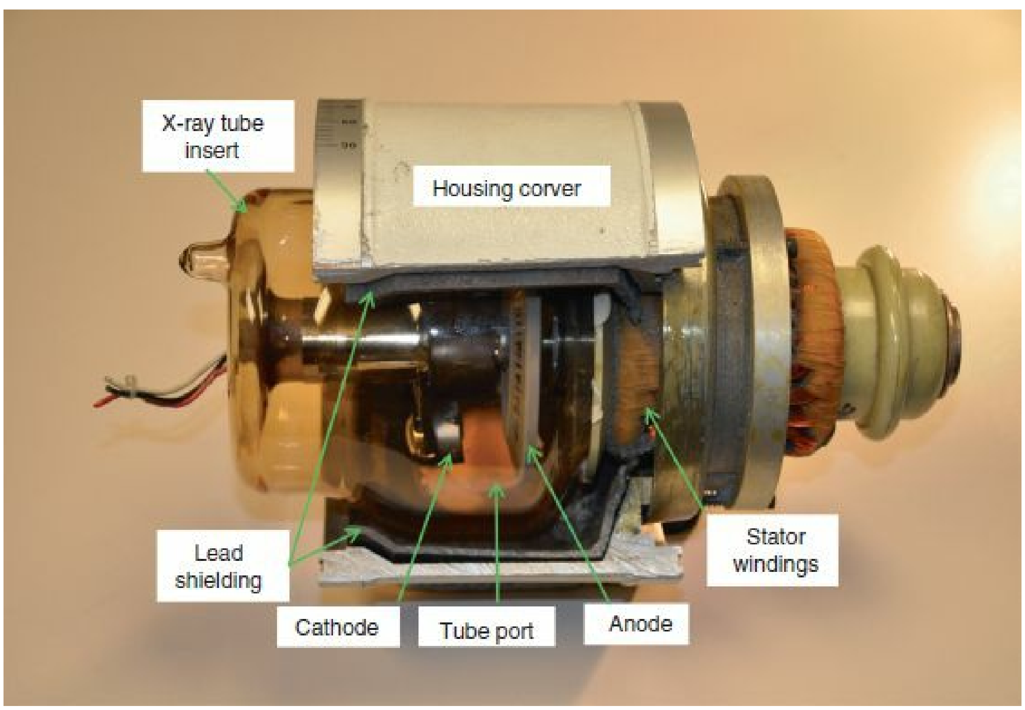
FIGURE 6-5 The filtered spectrum of bremsstrahlung and characteristic radiation from a tungsten target with a potential difference of 90 kV illustrates specific characteristic radiation energies from K_α and K_β transitions. Filtration (the preferential removal of low-energy photons as they traverse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



■ **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.

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 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.

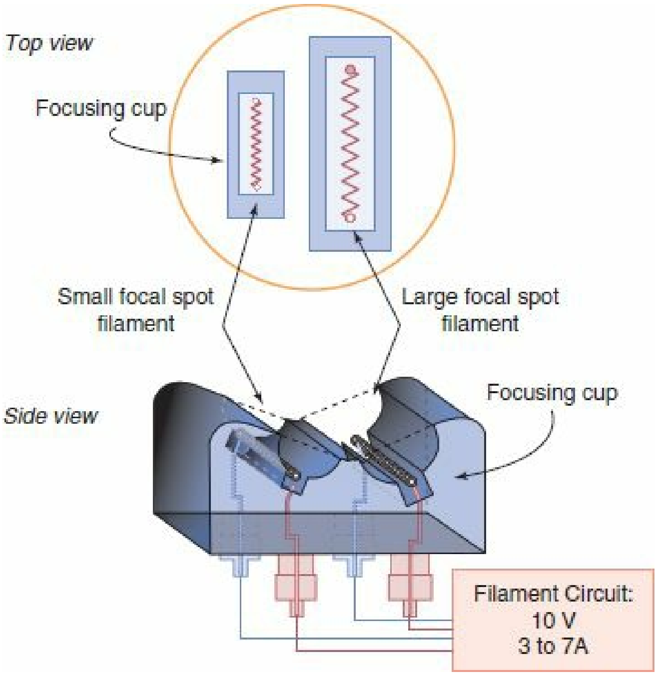
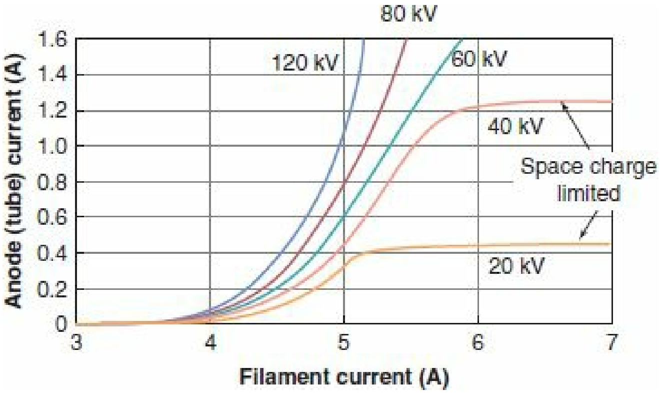


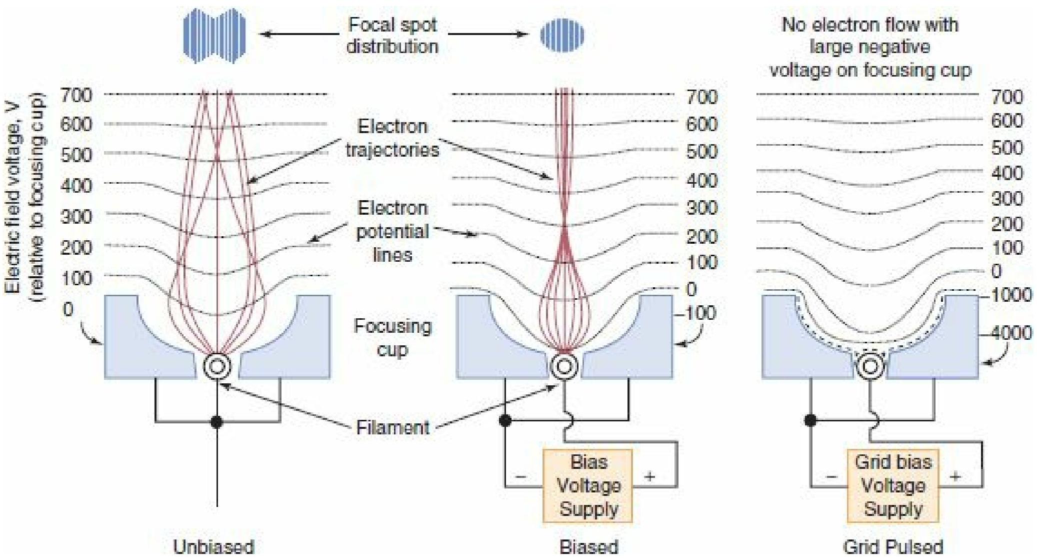
FIGURE 6-8 The x-ray tube cathode structure consists of the filaments and the focusing (or cathode) cup. Current from the filament circuit heats a filament, which releases electrons by thermionic emission.

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 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, produce lower average x-ray energies in the x-ray beam, and deliver unnecessary patient dose.



■ **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**).

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 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).

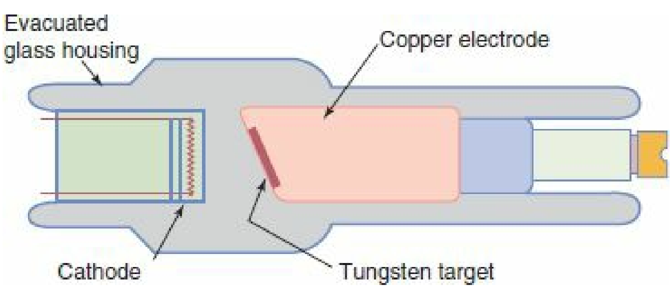


FIGURE 6-11 The anode of a fixed anode x-ray tube consists of a tungsten insert mounted in a copper block. Heat is removed from the tungsten target by conduction into the copper block.

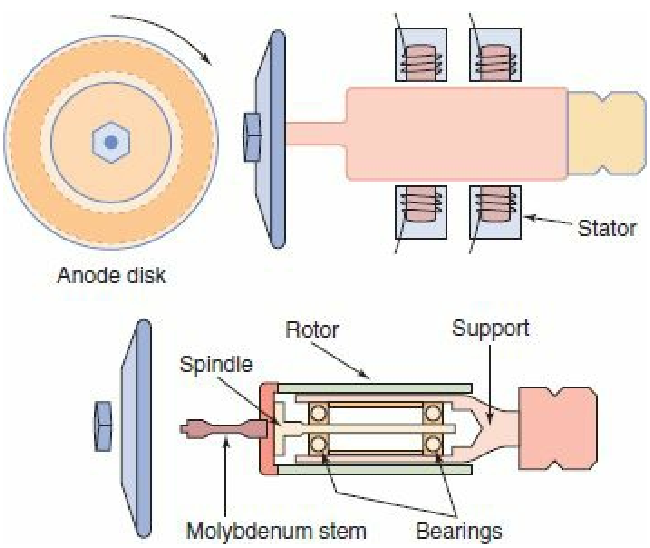
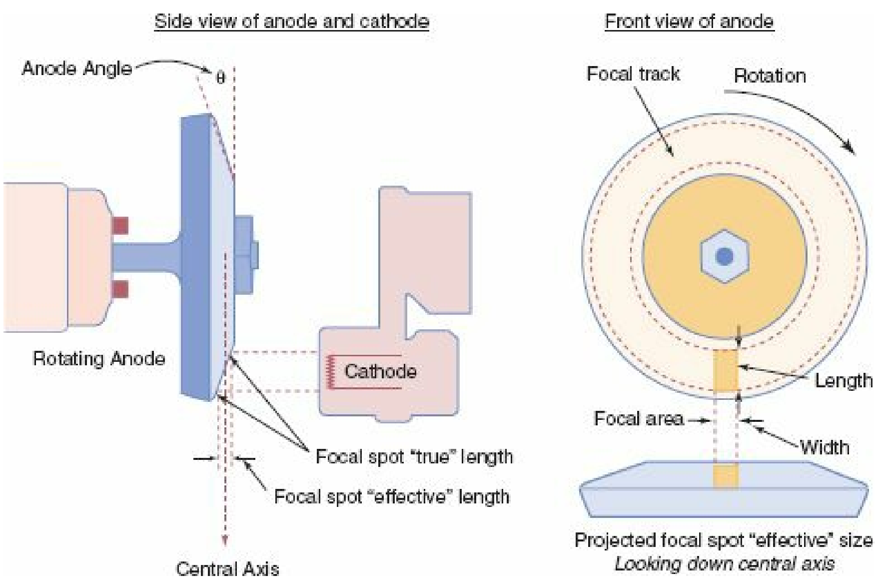


FIGURE 6-12 The anode of a rotating anode x-ray tube is a tungsten disk mounted on a

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**).

The focal track area of the rotating anode is approximately equal to the product of the circumferential track length (2πr) 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 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).



■ **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**).

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.

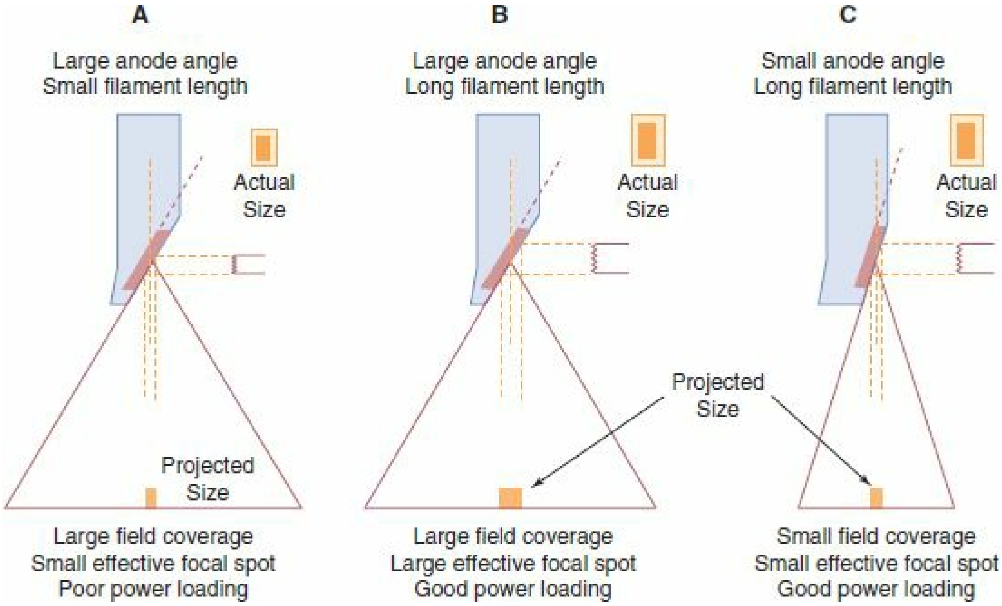
EXAMPLE 1: The actual anode focal area for a 20-degree anode angle is 4 mm (length) by 1.2 mm (width). What is the projected focal spot size at the central axis position?

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 (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.



■ **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.

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 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.

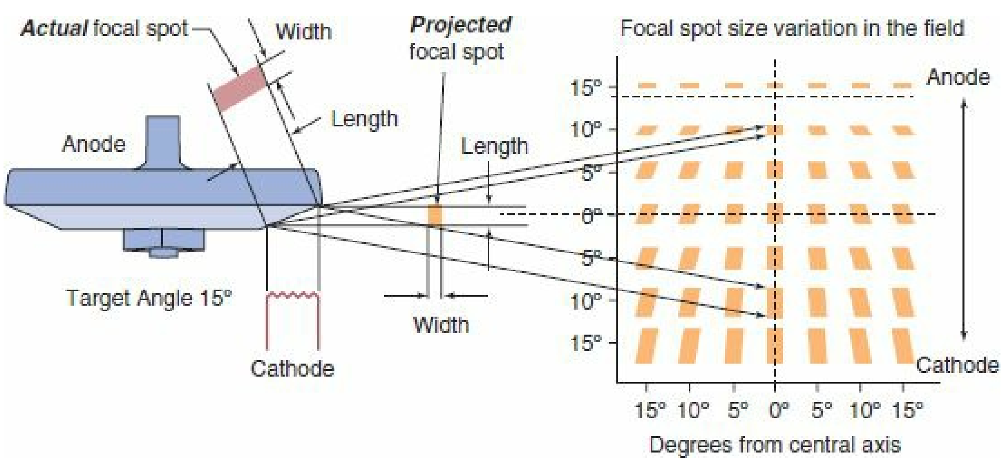


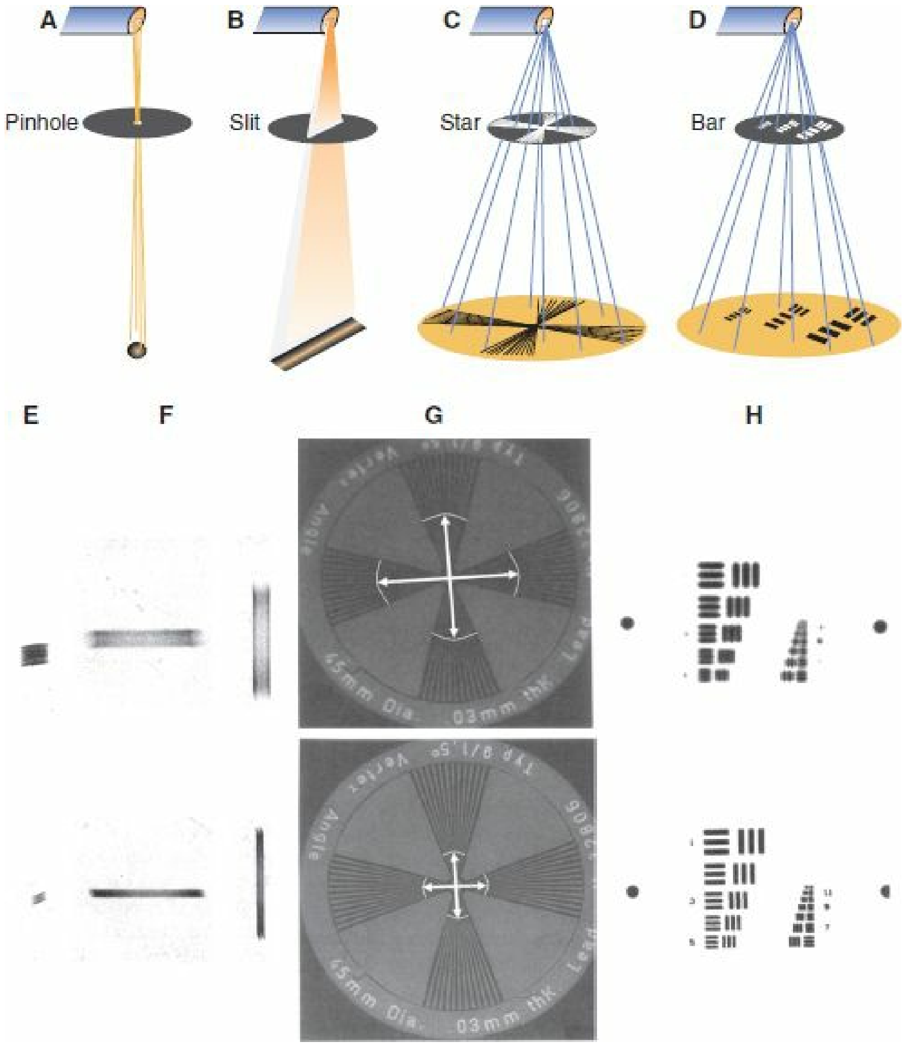
FIGURE 6-15 Variation of the effective focal spot size in the image field occurs along the anode-cathode direction. Focal spot distributions are plotted as a function of projection angle in degrees from the central axis, the parallel (vertical axis), and the perpendicular (horizontal axis).

The 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 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.



■ **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.

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.

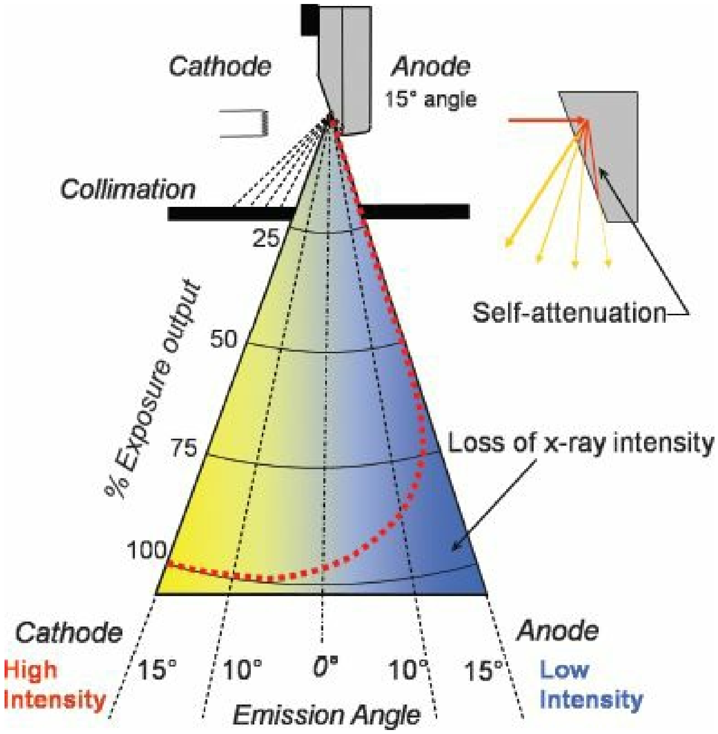


FIGURE 6-17 The heel effect is a loss of intensity on the anode side of the x-ray field of view. It is caused by attenuation of the x-ray beam by the anode. Upper right is an expanded view that shows electrons interacting at depth within the anode and the resultant "self attenuation" of produced x-rays 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 tube structures and degrade the vacuum. A "*getter*" circuit is used to trap gas in the insert and to maintain the vacuum.

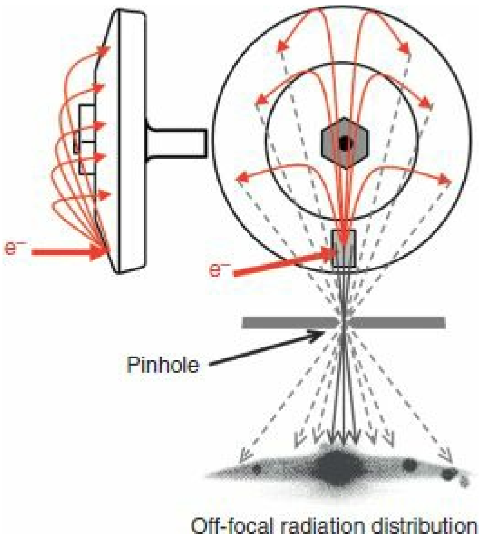


FIGURE 6-18 Off-focal radiation is produced from back-scattered electrons that are re-accelerated to the anode outside the focal spot. This causes a low-intensity, widespread radiation distribution pattern. Hotspots outside the focal spot indicate areas where the electrons are more likely to interact.

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 generator 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.

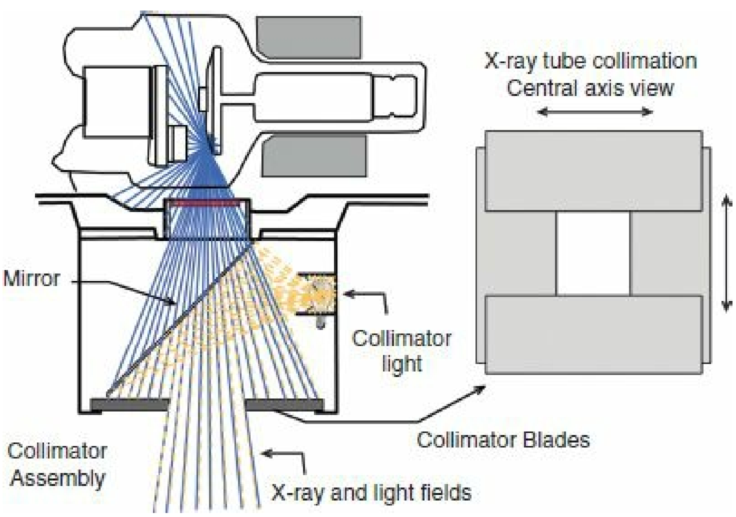


FIGURE 6-19 The x-ray tube collimator assembly is attached to the housing at the tube port, typically on a collar that allows it to be rotated. A light source, positioned at a virtual focal spot

location, illuminates the field from a 45-degree angle mirror. Lead collimator blades define both the x-ray and light fields.

Special X-ray Tube Designs

A grid-biased tube has a focusing cup that is electrically isolated from the cathode filament and maintained at a more negative voltage. When the bias voltage is sufficiently large, the resulting electric field stops the tube current. Turning off the grid bias allows the tube current to flow and x-rays to be produced. Grid biasing requires approximately −4,000 V applied to the focusing cup with respect to the filament to switch the x-ray tube current off (see Fig. 6-10). The grid-biased tube is used in applications such as pulsed fluoroscopy and cine-angiography, where rapid x-ray pulsing is necessary. Biased x-ray tubes are significantly more expensive than conventional, nonbiased tubes.

Mammography tubes are designed to provide the low-energy x-rays necessary to produce optimal mammographic images. As explained in Chapter 8, the main differences between a dedicated mammography tube and a conventional x-ray tube are the target materials (molybdenum, rhodium, and tungsten), the output port (beryllium versus glass or metal insert material), the smaller effective focal spot sizes (typically 0.3 and 0.1 mm nominal focal spot sizes), and the use of grounded anodes.

X-ray tubes for interventional fluoroscopy and CT require high instantaneous x-ray output and high heat loading and rapid cooling. Furthermore, in CT, with the fast x-ray tube rotation (as low as 0.3 s for a complete rotation about the patient) and the tremendous mechanical forces it places on the CT tube, planar surface cathode emitter designs different than the common helical filaments and enhanced bearing support for the rotating anode are often used. One manufacturer’s CT tube incorporates a novel design with the cathode and the anode as part of a metal vacuum enclosure that is attached to externally mounted bearings and drive motor to rotate the assembly as shown in Figure 6-20. Dynamic steering of the electron beam within the tube is achieved by external electromagnetic deflection coils to direct the electrons to distinct focal spots on the anode, which can produce slightly different projections and improve data sampling during the CT acquisition (refer to Chapter 10 on CT). Direct anode cooling by oil circulating within the housing provides extremely high cooling rates, and eliminates the need for high anode heat storage capacity. The cessation of imaging during clinical examinations to allow anode cooling is seldom necessary when using these advanced x-ray tubes.

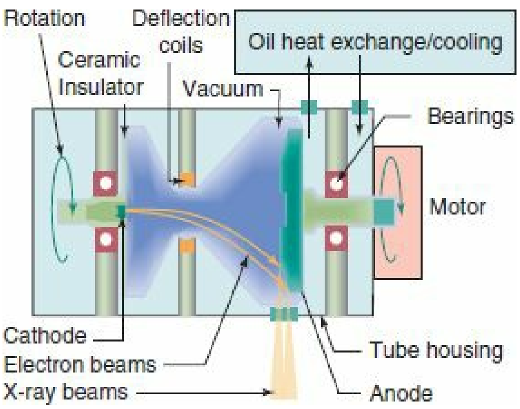


FIGURE 6-20 Diagram of an advanced CT x-ray tube, showing the anode and the planar cathode within a rotating vacuum enclosure. The bearings are mounted outside of the vacuum enclosure. Deflection coils magnetically direct the electron beam to specific areas on the target. Circulating oil rapidly removes excess heat from the anode. The electron beam can be rapidly deflected between two focal spots; this is known as a "flying focal spot."

Recommendations to Maximize X-ray Tube Life

X-ray tubes eventually must be replaced, but a long lifetime can be achieved with appropriate care and use. Several simple rules are discussed here. (1) Minimize filament boost "prep" time (the first detent of two on the x-ray exposure switch) especially when high mA is used. If applied for too long, filament life will be shortened, unstable operation will occur, and evaporated tungsten will be deposited on the glass envelope. (2) Use lower tube current with longer exposure times to arrive at the desired mAs if possible. (3) Avoid extended or repeated operation of the x-ray tube with high technique (kV and mAs) factors because, even though the x-ray generator has logic to prohibit single exposure settings that could damage the x-ray tube, multiple exposures could etch the focal track, resulting in less radiation output; transmit excessive heat to the bearings; and cause outgassing of the anode structure that will cause the tube to become unstable. (4) Always follow the manufacturer's recommended warm-up procedure. Do not make high mA exposures on a cold anode, because uneven expansion caused by thermal stress can cause cracks. (5) Limit rotor start and stop operations, which can generate significant heating and hot spots within the stator windings; when possible, a 30 to 40 s delay between exposures should be used.

Filtration

As mentioned earlier, filtration is the removal of x-rays as the beam passes through a layer of material. Filtration includes both the inherent filtration of the x-ray tube and added filtration. Inherent filtration includes the thickness (1 to 2 mm) of the glass or metal insert at the x-ray tube port. Glass (primarily silicon dioxide, SiO_2) and aluminum have similar attenuation properties (Z_{Si} = 14 and Z_{Al} = 13) and effectively attenuate all x-rays in the spectrum below about 15 keV. Dedicated mammography tubes, on the other hand, require beryllium (Z = 4) to permit the transmission of low-energy x-rays. Inherent filtration includes attenuation by housing oil and the field light mirror in the collimator assembly.

Added filtration refers to sheets of metal intentionally placed in the beam to change its effective energy. In general diagnostic radiology, added filters attenuate the low-energy x-rays in the spectrum that have almost no chance of penetrating the patient and reaching the x-ray detector. Because the low-energy x-rays are absorbed by the filters instead of the patient, radiation dose is reduced by beam filtration. Aluminum (Al) is the most commonly used added filter material. Other common filter materials include copper and plastic (e.g., acrylic). An example of the patient dose savings obtained with extra tube filtration is described in Section 6.5, point no. 4, beam filtration. In mammography, thin filters of Mo, Rh, and silver (Ag) are used to transmit bremsstrahlung x-rays in the intermediate energy range (15 to 25 keV), including characteristic radiation from Mo and Rh, and also to highly attenuate lowest and highest x-ray energies in the spectrum (see Chapter 8). Some have advocated using rare earth elements (K-absorption edges of 39 to 65 keV) such as erbium in filters in radiography to reduce patient dose and improve image contrast when contrast materials are used.

Compensation (equalization) filters are used to change the spatial pattern of the x-ray intensity incident on the patient, so as to deliver a more uniform x-ray exposure to the detector. For example, a trough filter used for chest radiography has a centrally located vertical band of reduced thickness and consequently produces greater x-ray fluence in the middle of the field. This filter compensates for the high attenuation of the mediastinum and reduces the exposure latitude incident on the image receptor. Wedge filters are useful for lateral projections in cervical-thoracic spine imaging, where the incident fluence is increased to match the increased tissue thickness encountered (e.g., to provide a low incident flux to the thin neck area and a high incident flux to the thick shoulders). “Bow-tie” filters are used in CT to reduce dose to the periphery of the patient, where x-ray paths are shorter and fewer x-rays are required. Compensation filters are placed close to the x-ray tube port or just external to the collimator assembly.

6.3 X-ray Generators

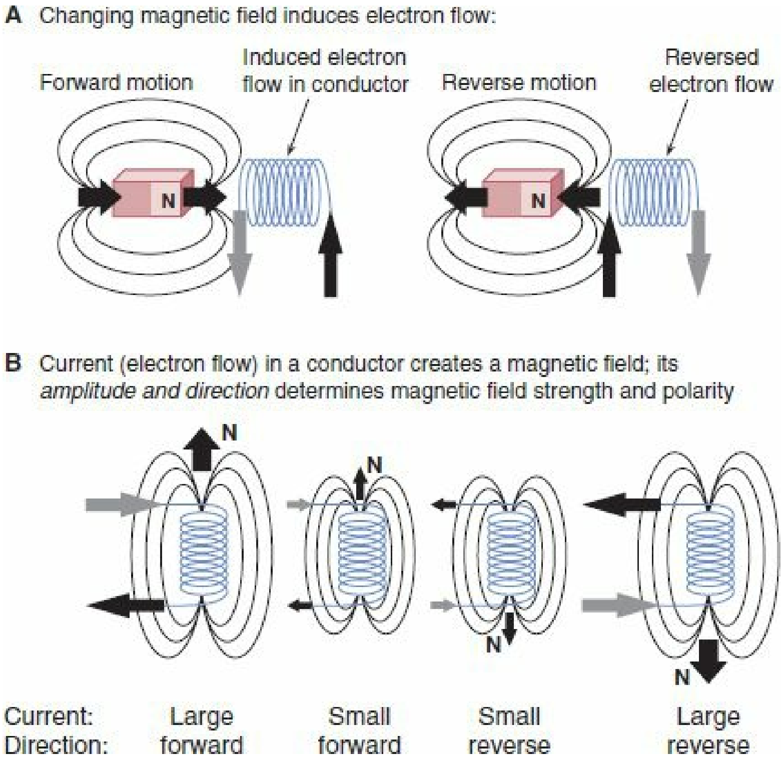
The principal function of an x-ray generator is to provide current at a high voltage to an x-ray tube. Electrical power available to a hospital or clinic provides up to about 480 V, much lower than the 20,000 to 150,000 V needed for x-ray production.

Transformers are principal components of x-ray generators; they convert low voltage into high voltage through a process called *electromagnetic induction*.

Electromagnetic Induction and Voltage Transformation

Electromagnetic induction is a phenomenon in which a changing magnetic field induces an electrical potential difference (voltage) in a nearby conductor and also in which a voltage is induced in a conductor moving through a stationary magnetic field. For example, the changing magnetic field from a moving bar magnet induces a voltage and a current in a nearby conducting wire (Fig. 6-21A). As the magnet moves in the opposite direction away from the wire, the induced current flows in the opposite direction. The magnitude of the induced voltage is proportional to the rate of change of the magnetic field strength.

Electrical current, such as the electrons flowing through a wire, produces a magnetic field whose magnitude (strength) is proportional to the magnitude of the current (see Fig. 6-21B). For a coiled wire geometry, superimposition of the magnetic fields from adjacent turns of the wire increases the amplitude of the overall magnetic field (the magnetic fields penetrate the wire’s insulation), and therefore the magnetic field strength is proportional to the number of wire turns. A constant current flowing through a wire or a coil produces a constant magnetic field, and a varying current produces a varying magnetic field. With an AC and voltage, such as the standard 60 cycles/s (Hz) AC in North America and 50 Hz AC in most other areas of the world, the induced magnetic field alternates with the current.



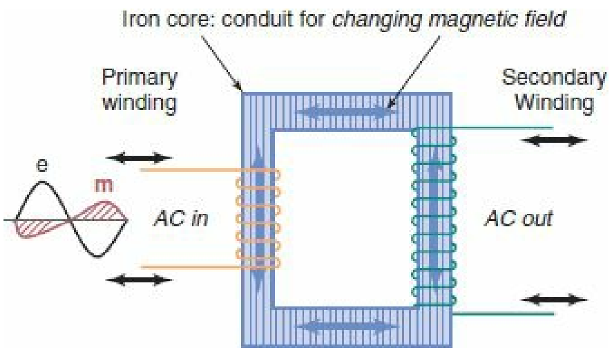
■ **FIGURE 6-21** Principles of electromagnetic induction are illustrated. **A**. Induction of an electrical current in a wire conductor coil by a moving (changing) magnetic field. The direction of the current is dependent on the direction of the magnetic field motion. **B**. Creation of a magnetic field by the current in a conducting coil. The polarity and magnetic field strength are determined by the amplitude and direction of the current.

A wire or a wire coil with a changing current will induce a voltage in a nearby wire or wire coil, and therefore, an AC applied to a wire or a wire coil will induce an alternating voltage in another wire or wire coil by electromagnetic induction. However, when a constant current, like that produced by a chemical battery, flows through a wire or a wire coil, although it creates a constant magnetic field, electromagnetic induction does not occur, and so it does not induce a voltage or current in a nearby conductor.

Transformers

Transformers use the principle of electromagnetic induction to change the voltage of an electrical power source. The generic transformer has two distinct, electrically insulated wires wrapped about a common iron core (Fig. 6-22). Input AC power produces an oscillating magnetic field on the "primary winding" of the transformer, where each turn of the wire amplifies the magnetic field that is unaffected by the electrical insulation and permeates the iron core. Contained within the core, the changing magnetic field induces a voltage on the "secondary winding," the magnitude of which is amplified by the number of turns of wire. The voltage induced in the second winding is proportional to the voltage on the primary winding and the ratio of the number of turns in the two windings, as stated by the *Law of Transformers*,

$$\frac{V_P}{V_S} = \frac{N_P}{N_S} \qquad [6\text{-}3]$$

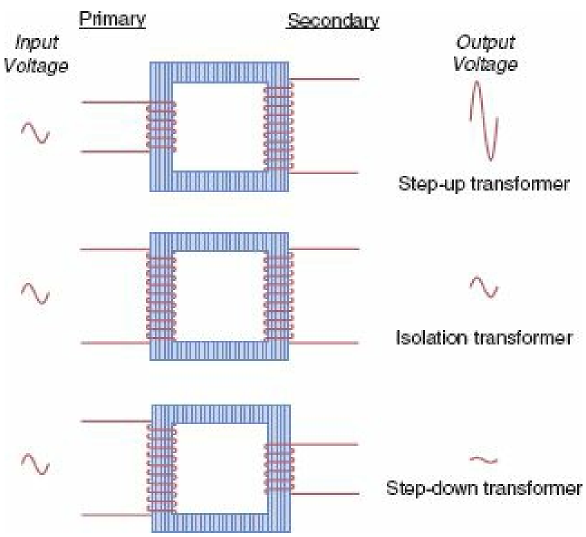


■ **FIGURE 6-22** The basic transformer consists of an iron core, a primary winding circuit, and a secondary winding circuit. An AC flowing through the primary winding produces a changing magnetic field, which permeates the core and induces an alternating voltage on the secondary winding. This mutual electromagnetic induction is mediated by the containment of the magnetic field in the iron core and permeability through wire insulation.

where N_p is the number of turns in the primary coil, N_s is the number of turns in the secondary coil, V_p is the amplitude of the input voltage on the primary side of the transformer, and V_s is the amplitude of the output voltage on the secondary side.

A transformer can increase, decrease, or isolate input voltage, depending on the ratio of the numbers of turns in the two coils. For $N_s > N_p$, a “step-up” transformer increases the secondary voltage; for $N_s < N_p$, a “step-down” transformer decreases the secondary voltage; and for $N_s = N_p$, an “isolation” transformer produces a secondary voltage equal to the primary voltage. Configurations of these transformers are shown in Figure 6-23. An input AC waveform is supplied to a transformer in order to produce a changing magnetic field and induce a voltage in the secondary winding. A step-up transformer circuit provides the high voltage necessary (20 to 150 kV) for a diagnostic x-ray generator.

For electrons to be accelerated to the anode in an x-ray tube, the voltage at the anode must be positive with respect to the cathode, but alternating waveforms change between negative and positive voltage polarity each half cycle. For continuous production of x-rays, the anode must be continuously at a positive voltage with respect to the cathode. However, this occurs only half of the time if an alternating voltage waveform is provided directly to the x-ray tube. A basic electrical component known as a *rectifier* will allow current to flow in one direction only. For instance, the x-ray tube itself can act as a rectifier, since current usually will flow only when the anode has positive and the cathode has negative polarity; however, if the anode becomes very hot from use and the accelerating voltage is applied with reverse polarity (tube cathode positive and anode negative), electrons can be released from the hot anode and accelerated into the cathode, possibly damaging it. A diode is a device with two terminals. When a voltage is applied between the terminals with a specific polarity, there is very little resistance and a large current flows; when the same voltage is applied with the opposite polarity, little or no current flows. Diodes come in all sizes, from large, x-ray tube–sized devices down to microscopic, solid-state components on an integrated circuit board. Clever use of diodes arranged in a *bridge rectifier circuit* can route the flow of electrons through an AC circuit to create a direct current (DC), a unidirectional movement of electrons in which the voltage polarity never reverses. Rectification is an important function of the x-ray generator.



■ **FIGURE 6-23** Transformers increase (step up), decrease (step down), or leave unchanged (isolate) the input voltage depending on the ratio of primary to secondary turns, according to the Law of Transformers. In all cases, the input and the output circuits are electrically isolated.

Power is the rate of energy production or expenditure per unit time. The SI unit of power is the watt (W), defined as 1 joule (J) of energy per second. For electrical devices, power is equal to the product of voltage and current.

$$P = I\,V \qquad [6\text{-}4]$$

Because a volt is defined as 1 joule per coulomb and an ampere is 1 coulomb per second,

$$1 \text{ watt} = 1 \text{ volt} \times 1 \text{ ampere}$$

For an ideal transformer, because the power output is equal to the power input, the product of voltage and current in the primary circuit is equal to that in the secondary circuit

$$V_P I_P = V_S I_S \qquad [6\text{-}5]$$

where I_P is the input current on the primary side and I_S is the output current on the secondary side. Therefore, a decrease in current must accompany an increase in voltage, and vice versa. Equations 6-3 and 6-5 describe ideal transformer performance. Power losses in an actual transformer due to inefficient coupling cause both the voltage and current on the secondary side of the transformer to be less than those predicted by these equations.

EXAMPLE: The ratio of primary to secondary turns is 1:1,000 in a transformer. If an input AC waveform has a peak voltage of 50 V, what is the peak voltage induced in the secondary side?

$$\frac{V_P}{V_S} = \frac{N_P}{N_S}; \frac{50}{V_S} = \frac{1}{1000}; V_S = 50 \times 1{,}000 = 50{,}000\ \text{V} = 50\text{kV}$$

What is the secondary current for a primary current of 10 A?

$$V_P\, I_P = V_S\, I_S; 50\ \text{V} \times 10\ \text{A} = 50{,}000\ \text{V} \times I_S; I_S = 0.001 \times 10\ \text{A} = 10\ \text{mA}$$

The high-voltage section of an x-ray generator contains a step-up transformer, typically with a primary-to-secondary turns ratio of 1:500 to 1:1,000. Within this range, a tube voltage of 100 kV requires an input line voltage of 200 to 100 V, respectively. The center of the secondary winding is usually connected to ground potential (“center tapped to ground”). Ground potential is the electrical potential of the earth. Center tapping to ground does not affect the maximum potential difference applied between the anode and cathode of the x-ray tube, but it limits the maximum voltage at any point in the circuit relative to ground to one half of the peak voltage applied to the tube. Therefore, the maximum voltage at any point in the circuit for a center-tapped transformer of 150 kV is −75 kV or +75 kV, relative to ground. This reduces electrical insulation requirements and improves electrical safety. In some x-ray tube designs (e.g., mammography and CT), the anode is maintained at the same potential as the body of the insert, which is maintained at ground potential. Even though this places the cathode at peak negative voltage with respect to ground, the low kV (less than 50 kV) used in mammography does not present a big electrical insulation problem, while in modern CT systems (up to 140 kV) the x-ray generator is placed adjacent to the x-ray tube in the enclosed gantry.

X-ray Generator Modules

Modules of the x-ray generator (Fig. 6-24) include the high-voltage power circuit, the stator circuit, the filament circuit, the focal spot selector, and automatic exposure control (AEC) circuit. Generators typically have circuitry and microprocessors that monitor the selection of potentially damaging overload conditions in order to protect the x-ray tube. Combinations of kV, mA, and exposure time delivering excessive power to the anode are identified, and such exposures are prohibited. Heat load monitors calculate the thermal loading on the x-ray tube anode, based on kV, mA, and exposure time, and taking cooling into account. Some x-ray systems are equipped with sensors that measure the temperature of the anode. These systems protect the x-ray tube and housing from excessive heat buildup by prohibiting exposures that would damage them. This is particularly important for CT scanners and high-powered interventional fluoroscopy systems.

Operator Console

For radiographic applications, a technologist at the operator's console can select the tube voltage (kV), the tube current (mA), the exposure time (s), or the product of mA and time (mAs), the AEC mode, the AEC sensors to be used, and the focal spot. If AEC is used, exposure time is not set. The focal spot size (i.e., large or small) is usually determined by the mA setting; low mA selections allow the small focal spot to be used, and higher mA settings require the use of the large focal spot due to anode heating concerns. On some x-ray generators, preprogrammed techniques can be selected for various examinations (e.g., chest; kidneys, ureter and bladder; cervical spine; and extremities). For fluoroscopic procedures, although kV and mA can be manually selected, the generator's automatic exposure rate control circuit, sometimes called the automatic brightness control (ABC), is commonly activated. It automatically sets the kV and mA from feedback signals from a sensor that indicates the radiation intensity at the image receptor. All console circuits have relatively low voltages and currents to minimize electrical hazards.

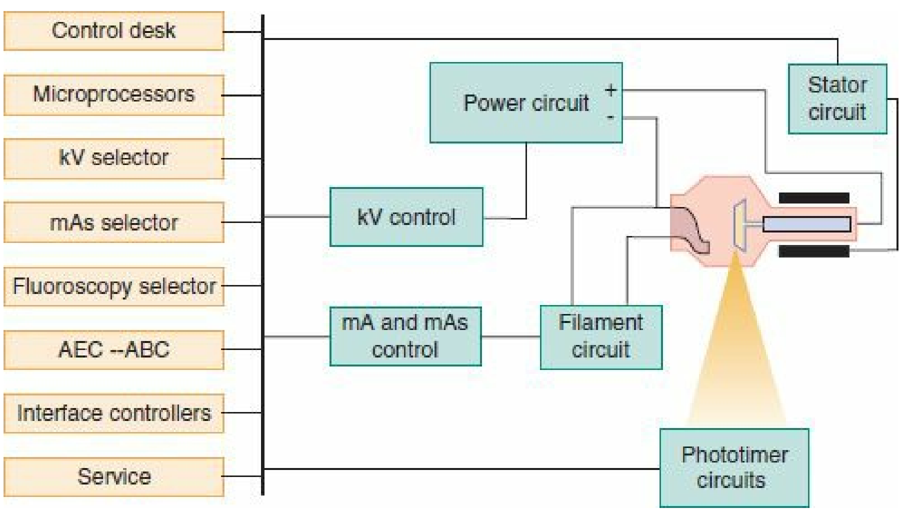
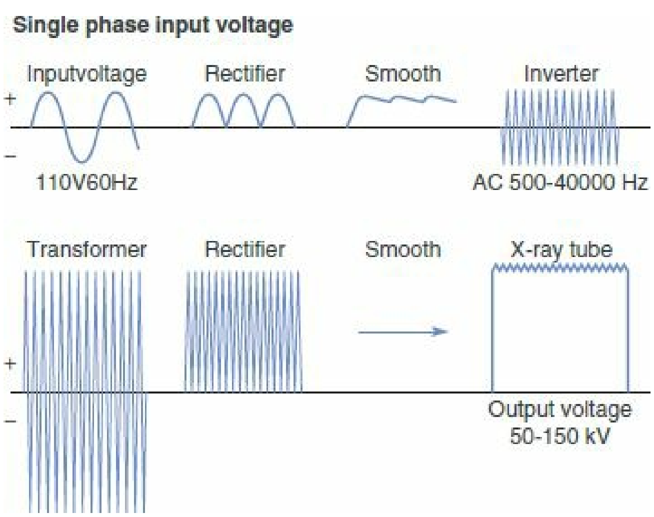


FIGURE 6-24 A modular schematic view shows the basic x-ray generator components. Most systems are now microprocessor controlled and include service support diagnostics.

High-Frequency X-ray Generator

Several x-ray generator circuit designs are in use, including single-phase, three-phase, constant potential, and high-frequency inverter generators. The high-frequency generator is now the contemporary state-of-the-art choice for diagnostic x-ray systems. Its name describes its function, whereby a high-frequency alternating waveform (up to 50,000 Hz) is used for efficient conversion of low to high voltage by a step-up transformer. Subsequent rectification and voltage smoothing produce a nearly constant output voltage. These conversion steps are illustrated in Figure 6-25. The operational frequency of the generator is variable, depending on the exposure settings (kV, mA, and time), the charge/discharge characteristics of the high-voltage capacitors on the x-ray tube, and the frequency-to-voltage characteristics of the transformer.

Figure 6-26 shows the components and circuit diagram of a general-purpose high-frequency inverter generator. Low-frequency, low-voltage input power (50 to 60 cycles/s AC) is converted to a low voltage, direct current. Next, an inverter circuit creates a high-frequency AC waveform, which supplies the high-voltage transformer to create a high-voltage, high-frequency waveform. Rectification and smoothing produces high-voltage DC power that charges the high-voltage capacitors placed across the anode and cathode in the x-ray tube circuit. Accumulated charge in the capacitors will produce a voltage to the x-ray tube according to the relationship $V = Q/C$, where V is the voltage (volts), Q is the charge (coulombs), and C is the capacitance (farads). During the x-ray exposure, feedback circuits monitor the tube voltage and tube current and continuously supply charge to the capacitors as needed to maintain a nearly constant voltage.



■ **FIGURE 6-25** In a high-frequency inverter generator, a single- or three-phase AC input voltage is rectified and smoothed to create a DC waveform. An inverter circuit produces a high-frequency AC waveform as input to the high-voltage transformer. Rectification and capacitance smoothing provide the resultant high-voltage output waveform, with properties similar to those of a three-phase system.

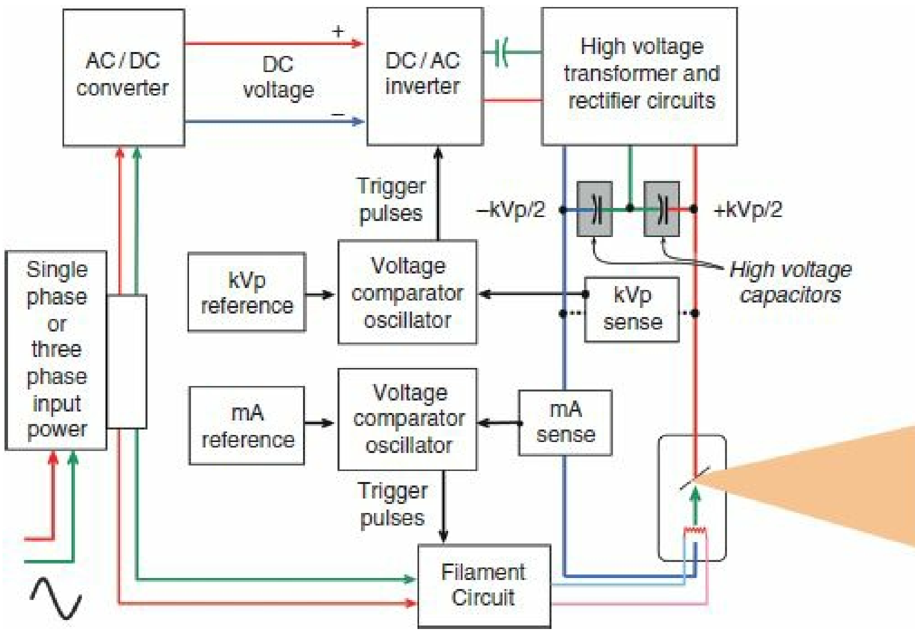


FIGURE 6-26 Modular components and circuits of the high-frequency generator. The selected high voltage across the x-Ray tube is created by charging high voltage capacitors to the desired potential difference. During the exposure when the x-ray circuit is energized, tube current is kept constant by the "mA sense" circuit that maintains the proper filament current by sending trigger pulses to the filament circuit , and tube voltage is kept constant by the "kV sense" circuit that sends trigger pulse signals to the DC/AC inverter to maintain the charge of high voltage capacitors.

For kV adjustment, a voltage comparator measures the difference between the reference voltage (a calibrated value proportional to the requested kV) and the actual kV measured across the tube by a voltage divider (the kV sense circuit). Trigger pulses generated by the comparator circuit produce a frequency that is proportional to the voltage difference between the reference signal and the measured signal. A large discrepancy in the compared signals results in a high trigger-pulse frequency, whereas no difference produces few or no trigger pulses. For each trigger pulse, the DC/AC inverter circuit produces a corresponding output pulse, which is subsequently converted to a high-voltage output pulse by the transformer. The high-voltage capacitors store the charge and increase the potential difference across the x-ray tube. When the requested x-ray tube voltage is reached, the output pulse rate of the comparator circuit settles down to a constant value, and recharging of the high-voltage capacitors is constant. When the actual tube voltage drops below a predetermined limit, the pulse rate increases. The feedback pulse rate (generator frequency) strongly depends on the tube current (mA), since the high-voltage capacitors discharge more rapidly with higher mA, thus actuating the kV comparator circuit. Because of the closed-loop voltage regulation, input line voltage compensation is not necessary, unlike older generator designs.

The mA is regulated in an analogous manner to the kV, with a resistor circuit sensing the actual mA (the voltage across a resistor is proportional to the current) and comparing it with a reference voltage. If the mA is too low, the mA comparator circuit increases the trigger frequency, which boosts the power to the filament to raise its temperature and increase the thermionic emission of electrons. The feedback circuit eliminates the need for space charge compensation circuits and automatically corrects for filament aging effects.

There are several advantages to the high-frequency inverter generator. Single-phase or three-phase input voltage can be used. Closed-loop feedback and regulation circuits ensure reproducible and accurate kV and mA values. Transformers operating at high frequencies are efficient, compact, and less costly to manufacture than those in other generator designs such as single-phase, three-phase, or constant-potential generators. Modular and compact design makes equipment installation and repairs relatively easy.

The high-frequency inverter generator is the preferred system for all but a few applications (e.g., those requiring extremely high power, extremely fast kV switching, or submillisecond exposure times provided by a *constant-potential* generator, which is very costly and requires more space).

Voltage Ripple

Ideally, the voltage applied to an x-ray tube would be constant. However, variations occur in the high voltage produced by an x-ray generator and applied to the x-ray tube. In an electronic waveform, voltage ripple is defined as the difference between the peak voltage and the minimum voltage, divided by the peak voltage and multiplied by 100%:

$$\%\ \text{voltage ripple} = \frac{V_{max} - V_{min}}{V_{max}} \times 100 \qquad [6\text{-}6]$$

The voltage ripple for various types of x-ray generators is shown in Figure 6-27. In theory, a single-phase generator, whether one-pulse or two-pulse output, has 100% voltage ripple. Actual voltage ripple for a single-phase generator is less than 100% because of cable capacitance effects (longer cables produce a greater capacitance, whereby the capacitance “borrows” voltage from the cable and returns it a short time later, smoothing the peaks and valleys of the voltage waveform). Three-phase 6-pulse and 12-pulse generators (not discussed in this chapter) have voltage ripples of 3% to 25%. High-frequency generators have a ripple that is kV and mA dependent, typically similar to a three-phase generator, ranging from 4% to 15%. Higher kV and mA settings result in less voltage ripple for these generators. Constant-potential generators have an extremely low ripple, less than 2%, but are expensive and bulky.

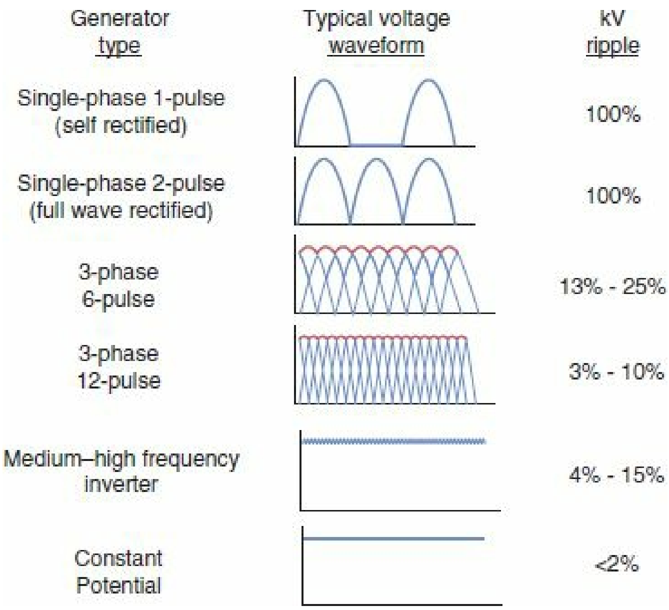


FIGURE 6-27 Typical voltage ripple for various x-ray generators used in diagnostic radiology varies from 100% voltage ripple for a single-phase generator to almost no ripple for a constant-potential generator.

Timers

Digital timers have largely replaced electronic timers based on resistor-capacitor circuits and charge-discharge timers in older systems. Digital timer circuits have extremely high reproducibility and microsecond accuracy, but the precision and accuracy of the x-ray exposure time depends chiefly on the type of switching (high voltage, low voltage, or x-ray tube switching) employed in the x-ray system. A countdown timer, also known as a backup timer, is used as a safety mechanism to terminate the exposure in the event of an exposure switch or timer failure.

Switches

The high-frequency inverter generator typically uses electronic switching on the primary side of the high-voltage transformer to initiate and stop the exposure. A relatively rapid response resulting from the high-frequency waveform characteristics of the generator circuit allows exposure times as short as 2 ms.

Alternatively, a grid-controlled x-ray tube can be used with any type of generator to switch the exposure on and off by applying a bias voltage (about ~4,000 V) to the focusing cup. This is the fastest switching method, with minimal turn-on–turnoff “lag”; however, there are extra expenses and high-voltage insulation issues to be considered.

Phototimer—Automatic Exposure Control

The phototimer, also known as the automatic exposure control (AEC) system, is often used instead of manual exposure time settings in radiography. Phototimers measure the actual amount of radiation incident on the image receptor (i.e., screen-film or digital radiography detector) and terminate x-ray production when the proper radiation exposure is obtained. Compensation for patient thickness and other variations in attenuation are achieved at the point of imaging. A phototimer system consists of one or more radiation detectors, an amplifier, a film density or digital SNR variable selector, a signal integrator circuit, a comparator circuit, a termination switch, and a backup timer safety shutoff switch (Fig. 6-28). X-rays transmitted through the patient and antiscatter grid, if present, generate ion pairs in one to three selectable ionization chambers positioned prior to the detector. An amplifier boosts the signal, which is fed to a voltage comparator and integration circuit. When the accumulated signal equals a preselected reference value, an output pulse terminates the exposure. A user-selectable “film density” or “SNR” selector on the generator control panel increases or decreases the reference voltage about 10% to 15% per step to modify the total accumulated x-ray exposure. For general diagnostic radiography, the phototimer sensors are placed in front of the image receptor to measure the transmitted x-ray flux through the patient (see Fig. 6-28). Positioning in front of the image receptor is possible because of the high transparency of the ionization chambers at the high kV values (greater than 50 kV) used for diagnostic radiography exams. In the event of a phototimer detector or circuit failure, the backup timer safety switch terminates the x-ray exposure after a preset “on” time.

To facilitate radiography of various anatomic projections, wall-mounted chest and table cassette stands and DR image receptors typically have three photocells arranged as shown in Figure 6-28 (front view). The technologist can select which photocells are used for each radiographic application. For instance, in posteroanterior chest imaging, the two outside chambers are usually activated, and the x-ray beam transmitted through the lungs determines the exposure time. This prevents signal saturation in the lung areas, which can occur when the transmitted x-ray flux is otherwise measured under the highly attenuating mediastinum with the center chamber.

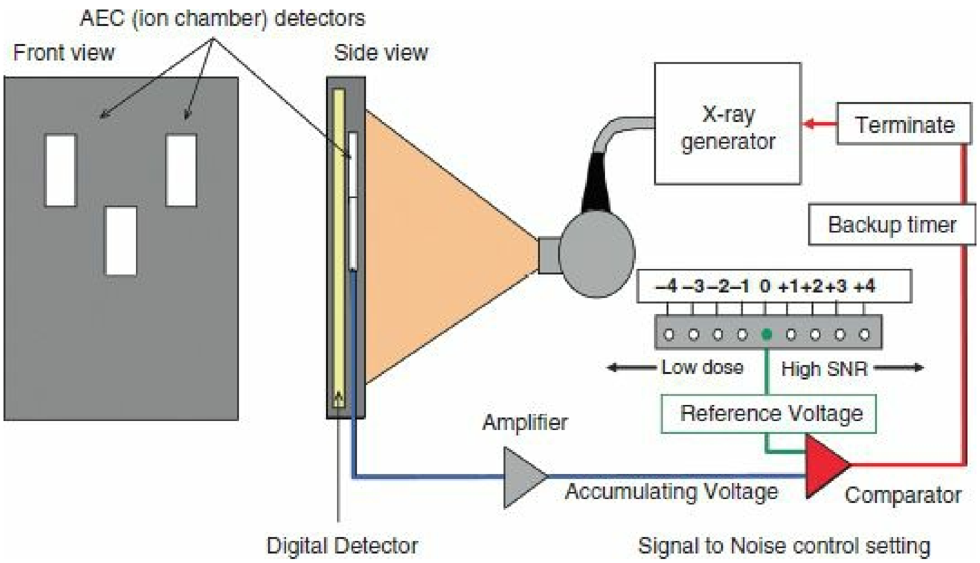


FIGURE 6-28 AEC detectors measure the radiation incident on the detector and terminate the exposure according to a preset optical density or signal-to-noise ratio achieved in the analog or digital image. A front view (**left**) and side view (**middle**) of a chest cassette stand and the locations of ionization chamber detectors are shown. The desired signal to the image receptor and thus the signal-to-noise ratio may be selected at the operator's console with respect to a normalized reference voltage.

6.4 Power Ratings and Heat Loading and Cooling

The *power rating* of an x-ray tube or generator is the maximal power that an x-ray tube focal spot can accept or the generator can deliver. General diagnostic x-ray tubes and x-ray generators use 100 kV and the maximum ampere rating available for a 0.1 s exposure as the benchmark, as

$$\text{Power (kW)} = 100\ \text{kV} \times \text{I}\ (\text{A}_{max}\ \text{for a 0.1 s exposure}) \qquad [6\text{-}7]$$

For instance, a generator that can deliver 800 mA (0.8 A maximum) of tube current at 100 kV for 0.1 second exposure has a power rating of 80 kW according to Equation 6-7. For applications such as computed tomography and interventional fluoroscopy, higher- power x-ray generators are specified, typically of 80 to 100 kW or higher. However, this power capability must be matched to the power deposition capability of the focal spot; otherwise, the power cannot be realized without exceeding the x-ray tube heat limitations. Focal spot dimensions of approximately 1.2 mm × 1.2 mm, large diameter anodes, and fast rotation speeds are necessary to achieve 80 to 100 kW of power deposition. Most medium focal spot dimensions (0.6 mm × 0.6 mm) have moderate power ratings (30 to 50 kW), and smaller focal spots (0.3 mm × 0.3 mm) have low power ratings (5 to 15 kW), but power ratings are dependent on several engineering, operational, and x-ray tube design factors. For instance, in modern CT x-ray tubes, a shallow anode angle (7°), bearings mounted outside of the vacuum of the insert, and heat exchangers allow for extended high power operation (e.g., continuous operation at 800 mA at 120 kV for minutes). X-ray generators have circuits or programs to prohibit combinations of kV, mA, and exposure time that will exceed the single-exposure power deposition tolerance of the focal spot.

X-ray generator power ratings vary considerably. The highest generator power ratings (80 to 120 kW) are found in interventional radiology and cardiovascular fluoroscopic imaging suites, while modern multirow detector CT scanners have 85- to 100-kW generators. General radiographic or radiographic/fluoroscopic systems use generators providing 30 to 80 kW. Lower powered generators (5 to 30 kW) are found in mobile radiography and fluoroscopy systems, dental x-ray systems, and other systems that have fixed anode x-ray tubes. When purchasing an x-ray generator and x-ray tube system, the clinical application must be considered, as well as the matching of the power ratings of the x-ray tube large focal spot and the x-ray generator. Otherwise, there will be a mismatch in terms of capability, and most likely needless expense.

Heat Loading

The Heat Unit

The heat unit (HU) is a traditional unit that provides a simple way of expressing the energy deposition on and dissipation from the anode of an x-ray tube. The number of HU can be calculated from the parameters defining the radiographic technique

$$\text{Energy (HU)} = \text{peak voltage (kV)} \times \text{tube current (mA)} \times \text{exposure time (s)} \qquad [6\text{-}8]$$

Although Equation 6-8 is correct for single-phase generator waveforms, it underestimates the energy deposition of three-phase, high-frequency, and constant-potential generators because of their lower voltage ripple and higher average voltage. A multiplicative factor of 1.35 to 1.40 compensates for this difference, the latter value being applied to constant-potential waveforms. For example, an exposure of 80 kV, 250 mA, and 100 ms results in the following HU deposition on the anode:

Single-phase generator: $80 \times 250 \times 0.100 = 2{,}000$ HU
Three-phase or high-frequency inverter generator: 80×250
$\times\, 0.100 \times 1.35 = 2{,}700$ HU

For continuous x-ray production (fluoroscopy), the HU/s is defined as follows:

$$\text{HU / s} = \text{kV} \times \text{mA} \qquad [6\text{-}9]$$

Heat accumulates by energy deposition and simultaneously disperses by anode cooling. The anode cools faster at higher temperatures, so that for most fluoroscopic and long CT procedures, steady-state equilibrium exists and Equation 6-9 will overestimate the heat energy delivered to the anode. This is taken into consideration by the anode thermal characteristics chart for heat input and heat output (cooling) curves.

The Joule

The joule (J) is the SI unit of energy. One joule is deposited by a power of one watt acting for 1 s (1 J = 1 W × 1 s). The energy, in joules, deposited in the anode is calculated as follows:

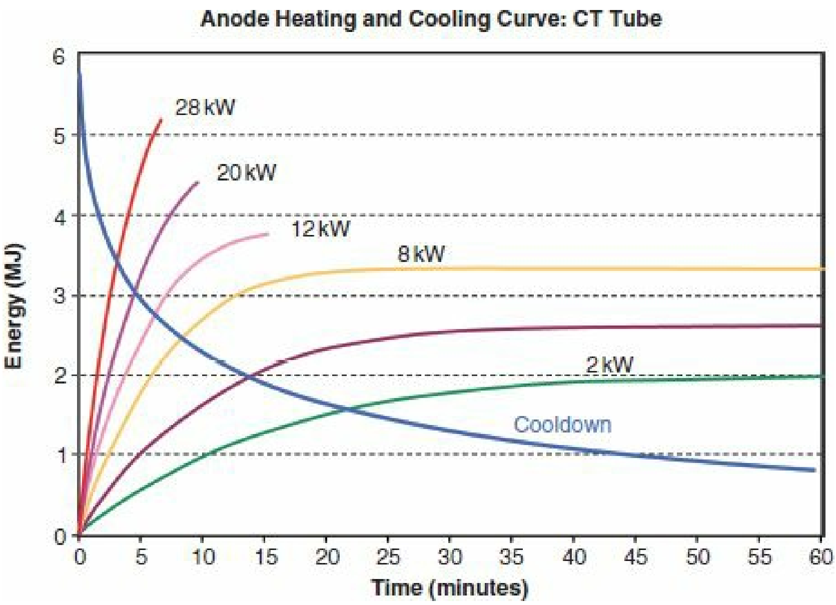
$$\text{Energy (J)} = \text{Voltage (V)} \times \text{Tube current (A)} \times \text{Exposure time(s)} \qquad [6\text{-}10]$$

The root-mean-square voltage, V_{RMS}, is the constant voltage that would deliver the same power as a varying voltage waveform (e.g., for a single phase generator), which would be substituted for V, and, from Equation 6-8 and 6-10, the relationship between the deposited energy in joules and HU can be approximated as

$$\text{Energy (HU)} \cong 1.4 \times \text{Heat input (J)} \qquad [6\text{-}11]$$

Anode Heating and Cooling Chart

Multiple x-ray exposures, continuous x-ray tube operation with CT, and prolonged fluoroscopy in interventional procedures deliver significant heat energy to the anode. An *anode heating and cooling chart* (Fig. 6-29) shows the anode heat loading for various input powers (kW or HU/s) as the x-ray tube is operating, taking into account the cooling that simultaneously occurs. For low-input power techniques, the heat loading increases to a plateau, where the cooling rate equals the heating rate. At higher power inputs, heating exceeds cooling, and after a certain time, a limit is reached where x-ray production must be stopped and anode cooling must be allowed prior to using the x-ray tube again. With larger anode heat capacities, the anode cooling curve is steeper because higher temperatures result in more rapid cooling (the "black-body" effect, where the radiative cooling rate is proportional to T^4, for absolute temperature T). The maximal anode heat capacity is indicated by the peak value of the cooling curve on the y-axis (0 time) of the chart. Each anode heating and cooling chart is specific to a particular x-ray tube and housing assembly. Some high-powered x-ray tubes for CT exceed 8 MHU anode heat loading with extremely high anode and x-ray tube housing cooling rates, achieved through advanced anode, rotor bearing, and housing designs. An example of a heating and cooling chart for a modern x-ray CT tube, with a maximum anode heat capacity of 5.7 megajoules (MJ) (equivalent to 8.0 MHU), is shown in Figure 6-29. If the x-ray technique is high or a large number of sequential or continuous exposures have been taken (e.g., during operation of a CT scanner), it can be necessary to wait before reenergizing the x-ray tube to avoid damage to the anode. Use of the specific anode cooling curve can determine how long to wait, based upon the total amount of accumulated heat, the simultaneous cooling that occurs, and the amount of power that will be deposited during the next acquisition. Modern x-ray systems have sensors and protection circuits to prevent overheating.



■ **FIGURE 6-29** Anode heating and cooling curve chart for a CT scanner plots *energy* in megajoules (MJ) on the vertical axis and *time* in minutes on the horizontal axis. A series of power input curves from low (2 kW) to high (28 kW) are determined by the kV and mA settings with continuous x-ray tube operation as a function of time. The cooling curve shows the rate of cooling and indicates faster cooling with higher anode heat load (temperature). In this example the maximum capacity is 5.7 MJ. For low power inputs, heating and cooling rates eventually equilibrate and reach a steady state, as shown for the 2, 4, and 8 kW curves.

6.5 Factors Affecting X-ray Emission

The output of an x-ray tube is often described by the terms quality, quantity, and exposure. *Quality* describes the penetrability of an x-ray beam, with higher energy x-ray photons having a larger half-value layer (HVL) and higher "quality." (The HVL is discussed in Chapter 3.) *Quantity* refers to the number of photons comprising the beam. *Exposure,* defined in Chapter 3, is nearly proportional to the energy fluence of the x-ray beam. X-ray production efficiency, exposure, quality, and quantity are determined by six major factors: x-ray tube target material, tube voltage, tube current, exposure time, beam filtration, and generator waveform.

1. **Anode target material** affects the *efficiency* of bremsstrahlung radiation production, with output exposure roughly proportional to atomic number. Incident electrons are more likely to have radiative interactions in higher-Z materials (see Equation 6-1). The energies of characteristic x-rays produced in the target depend on the target material. Therefore, the target material affects the

quantity of bremsstrahlung photons and the quality of the characteristic radiation.

2. **Tube voltage** (kV) determines the maximum energy in the bremsstrahlung spectrum and affects the quality of the output spectrum. In addition, the efficiency of x-ray production is directly related to tube voltage. Exposure is approximately proportional to the square of the kV in the diagnostic energy range.

$$\text{Exposure} \propto \text{kV}^2 \qquad [6\text{-}12]$$

For example, according to Equation 6-12, the relative exposure of a beam generated with 80 kV compared with that of 60 kV for the same tube current and exposure time is calculated as follows:

$$\left(\frac{80}{60}\right)^2 \cong 1.78$$

Therefore, the output exposure increases by approximately 78% (Fig. 6-30). *An increase in kV increases the efficiency of x-ray production and the quantity and quality of the x-ray beam.*

Changes in the kV must be compensated by corresponding changes in mAs to maintain the same exposure. At 80 kV, 1.78 units of exposure occur for every 1 unit of exposure at 60 kV. To achieve the original 1 unit of exposure, the mAs must be adjusted to 1/1.78 = 0.56 times the original mAs, which is a *reduction* of 44%. An additional consideration of technique adjustment concerns the x-ray attenuation characteristics of the patient. To achieve equal transmitted exposure through a typical patient (e.g., 20-cm tissue), the compensatory mAs varies with approximately the fifth power of the kV ratio

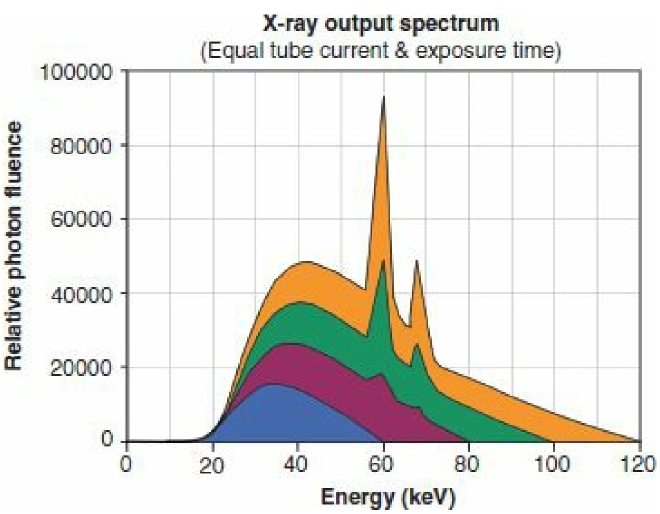


FIGURE 6-30 x-Ray tube output intensity varies as the square of tube voltage (kV). In this example, the same tube current and exposure times (mAs) are compared for 60 to 120 kV. The relative area under each spectrum roughly follows a squared dependence (characteristic

radiation is ignored).

$$\left(\frac{kVp_1}{kVp_2}\right)^5 \times \ mAs_1 = mAs_2 \qquad [6\text{-}13]$$

According to Equation 6-13, if a 60-kV exposure requires 40 mAs for a proper exposure through a typical adult patient, at 80 kV the adjusted mAs is approximately

$$\left(\frac{60}{80}\right)^5 \times 40 \text{ mAs} \cong 9.5 \text{ mAs}$$

or about one fourth of the original mAs. The value of the exponent (between four and five) depends on the thickness and attenuation characteristics of the patient.

3. **Tube current** (mA) is proportional to the number of electrons flowing from the cathode to the anode per unit time. The exposure of the beam for a given kV and filtration is proportional to the tube current. Also the exposure time is the duration of x-ray production. The quantity of x-rays is directly proportional to the product of tube current and exposure time (mAs), as shown in Figure 6-31.

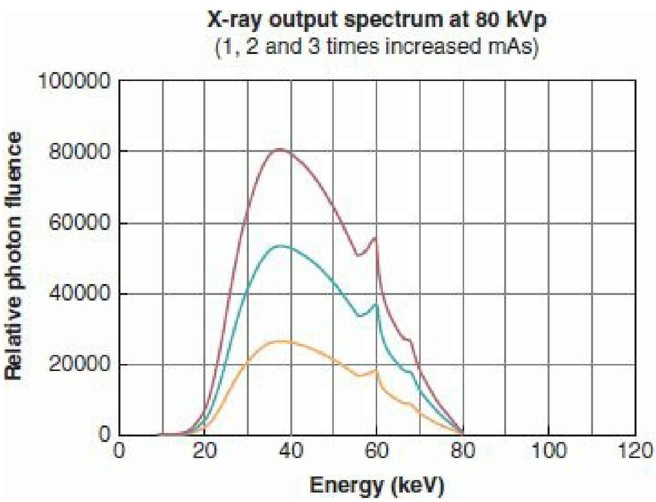


FIGURE 6-31 X-ray tube output intensity is proportional to the mAs (tube current and exposure time). Shown is the result of increasing the mAs from a baseline value (orange curve) by a factor of two (blue curve) and three (red curve), with a proportional change in the number of x-rays produced.

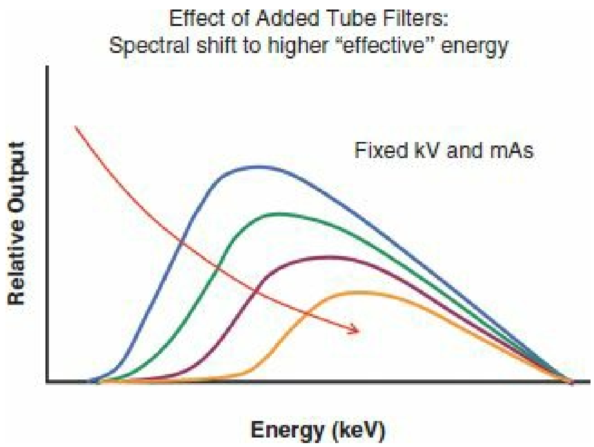


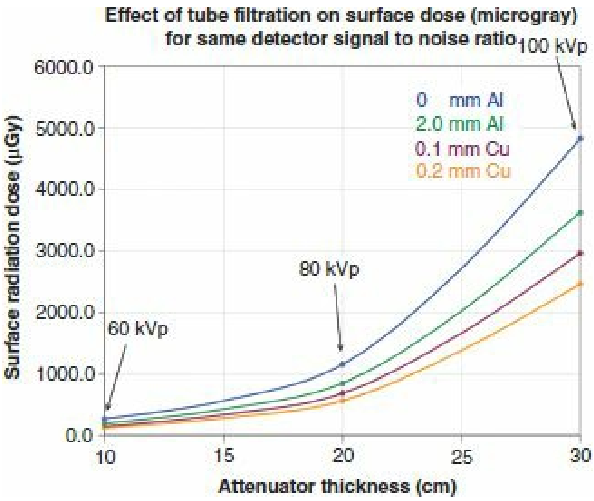
FIGURE 6-32 X-ray tube output intensity decreases and spectral quality (effective energy) increases with increasing thickness of added tube filters. Shown are spectra with added filtration at the same kV and mAs.

4. **Beam filtration** modifies the quantity and quality of the x-ray beam by preferentially removing the low-energy photons in the spectrum. This reduces the number of photons (quantity) and increases the average energy, also increasing the quality (Fig. 6-32). For highly filtered beams, the mAs required to achieve a particular x-ray intensity will be much higher than for lightly filtered beams; therefore, it is necessary to know the HVL of the beam and the normalized x-ray tube intensity per mAs (in addition to the geometry) to calculate the incident exposure and radiation dose to the patient. One cannot simply use kV and mAs for determining the "proper technique" or to estimate the dose to the patient without this information. In the United States, x-ray system manufacturers must comply with minimum HVL requirements specified in the Code of Federal Regulations, 21CFR 1020.30 (Table 6-3), which have been adopted by many state regulatory agencies. A pertinent clinical example of the radiation dose savings achievable with added filtration compared to the minimum filtration is shown in Figure 6-33, comparing the surface entrance radiation dose to the *same signal* generated in the detector for 0 mm Al (minimum filtration), 2 mm Al, 0.1 mm Cu, and 0.2 mm Cu. With more filtration, the dose savings become greater. For instance, as listed in Table 6-4, the comparison of 30 cm polymethylmethacrylate (PMMA) at 100 kV indicates a dose savings of 49% when a 0.2 mm Cu filter is added to the beam. To achieve this requires an increase in mAs of 37%, from 14.5 to 19.8 mAs at 100 kV. Even though the acquisition technique is much higher, the entrance dose is much lower, without any loss of image quality, particularly for digital imaging devices where image contrast and brightness enhancements are easily applied. Using added filtration in a modern x-ray collimator assembly is as simple as pushing the filter selection button, or automatically selecting the filter with anatomical programming.

TABLE 6-3 MINIMUM HVL REQUIREMENTS FOR X-RAY SYSTEMS IN THE UNITED STATES (21 CFR 1020.30)

DESIGNED OPERATING RANGE	MEASURED X-RAY TUBE VOLTAGE (kV)	MINIMUM HVL (mm OF ALUMINUM)
<51 kV	30	0.3
	40	0.4
	50	0.5
51–70 kV	51	1.3
	60	1.5
	70	1.8
>70 kV	71	2.5
	80	2.9
	90	3.2
	100	3.6
	110	3.9
	120	4.3
	130	4.7
	140	5.0
	150	5.4

Note: This table refers to systems manufactured after June 2006. It does not include values for dental or mammography systems.

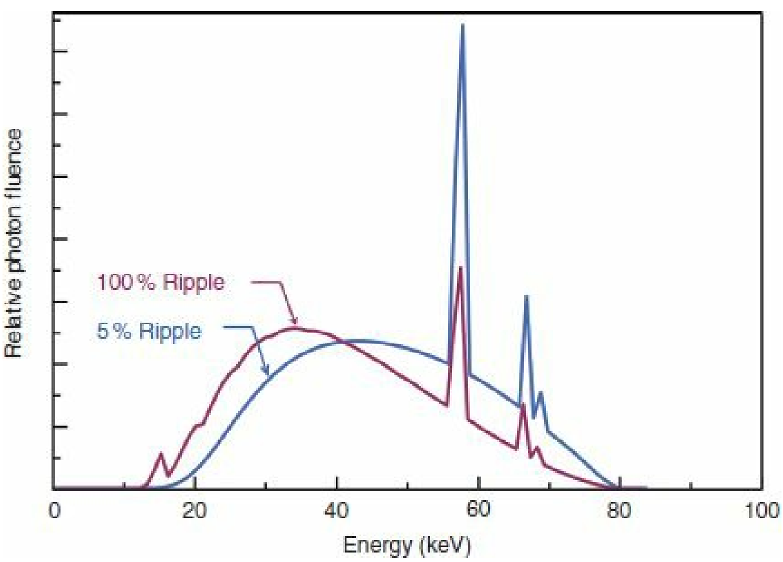


■ **FIGURE 6-33** Added x-ray tube filters can significantly reduce patient dose. Compared are measured entrance doses (air kerma with backscatter) to 10, 20, and 30 cm sheets of PMMA when using phototimed exposures at 60, 80, and 100 kV, respectively. The top curve represents the nominal beam condition, then 2.0 mm Al, 0.1 mm Cu + 1 mm Al, and 0.2 mm Cu + 1 mm Al for the lowest curve. For constant signal-to-noise ratio, the mAs is increased to compensate for added filter attenuation, as listed in Table 6-4.

TABLE 6-4 TUBE FILTRATION, MEASURED CHANGES IN REQUIRED MAS, AND MEASURED SURFACE DOSE (MGY) FOR EQUIVALENT SIGNAL IN THE OUTPUT DIGITAL IMAGE

	10 cm PMMA (60 kV)				20 cm PMMA (80 kV)				30 cm PMMA (100 kV)			
	TUBE CURRENT		DOSE (µGy)		TUBE CURRENT		DOSE (µGy)		TUBE CURRENT		DOSE (µGy)	
FILTRATION	mAs	% Δ	DOSE	% Δ	mAs	% Δ	DOSE	% Δ	mAs	% Δ	DOSE	% Δ
0 mm Al	3.8	0	264	0	6.8	0	1,153	0	14.5	0	4,827	0
2 mm Al	5.0	32	188	−29	8.2	21	839	−27	16.5	14	3,613	−25
0.1 mm Cu + 1 mm Al	6.2	63	150	−43	9.3	37	680	−41	17.6	21	2,960	−39
0.2 mm Cu + 1 mm Al	8.8	132	123	−53	11.2	65	557	−52	19.8	37	2,459	−49

Note: % Δ indicates the percentage change from the *no added filtration* measurement. For mAs, there is an increase in the mAs required to compensate for increased attenuation of the beam by the added filters. For surface dose, there is a decrease in the percentage surface dose change with more added filters for equivalent absorbed signal in the detector because of less attenuation of the beam in the object.



■ **FIGURE 6-34** Output intensity (bremsstrahlung) spectra for the same tube voltage (kV) and the same tube current and exposure time (mAs) demonstrate the higher effective energy and greater output of a three-phase or high-frequency generator voltage waveform (~5% voltage ripple), compared with a single-phase generator waveform (100% voltage ripple).

5. **Generator waveform** affects the quality of the emitted x-ray spectrum. For the same kV, a single-phase generator provides a lower average potential difference than does a three-phase or high-frequency generator. Both the quality and quantity of the x-ray spectrum are affected (Fig. 6-34).

The x-ray *quantity* is approximately proportional to $Z_{target} \times kV^2 \times mAs$. The x-ray *quality* depends on the kV, the generator waveform, and the tube filtration. Exposure depends on both the quantity and quality of the x-ray beam. Compensation for changes in kV with radiographic techniques requires adjustments of mAs on the order of the fourth to fifth power of the kV ratio, because kV determines quantity, quality, and transmission through the object, whereas mAs determines quantity only. Added filters to the x-ray tube can significantly lower patient dose by selectively attenuating the low energy x-rays in the bremsstrahlung spectrum, but require a compensatory increase in the mAs.

In summary, x-rays are the basic radiologic tool for a majority of medical diagnostic imaging procedures. Knowledge of x-ray production, x-ray generators, and x-ray beam control is important for further understanding of the image formation process and the need to obtain the highest image quality at the lowest possible radiation exposure.

SUGGESTED READING

Krestel E, ed. *Imaging systems for medical diagnostics*. Berlin/Munich, Germany: Siemens Aktiengesellschaft, 1990.

McCollough CH. The AAPM/RSNA physics tutorial for residents: x-ray production. *Radiographics* 1997;17:967–984.

Seibert JA. The AAPM/RSNA physics tutorial for residents: x-ray generators. *Radiographics* 1997;17: 1533–1557.

"Nope … no sign of your kitten, Ma'am. But to be absolutely certain, we'd better perform a CAT scan."