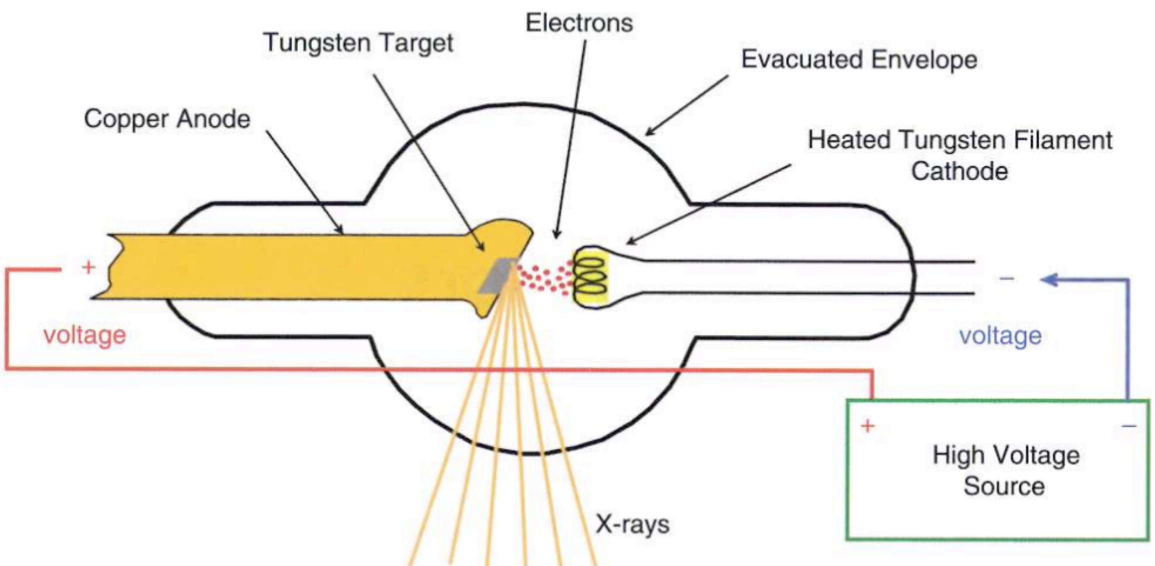
Production of X-rays

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

Production of X-rays

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 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.



■ 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.

Electrical potential difference

Electrical current is the flow of electrical charge
1 Ampere (A) is 1 Coulomb (C) passing a point in an electric circuit per second

Voltage or electrical potential (V) is the difference in the electrical potential energy of an electrical charge at 2 position (a,b) divided by the charge

$$V_{ab} = (E_{pa} - E_{pb})/q$$

1 V = 1 J/C

According to the principle of conservation of energy, the gain in kinetic energy of a charged particle is equal to its loss of potential energy

$$K_{final} - K_{initial} = E_{p\ initial} - E_{p\ final}$$

Assuming that $K_{initial} = 0$

$$K_{final} = q\,V$$

For example, the kinetic energy of an electron (charge = 1.602×10^{-19} C) accelerated through a electrical potential difference of 100 kV is:
$K_{final} = q\,V = (1.602 \times 10^{-19}\ \text{C})\ (100\ \text{kV}) = 1.602 \times 10^{-14}\ \text{J}$

1 eV is the kinetic energy of an electron accelerated across a potential of 1V:

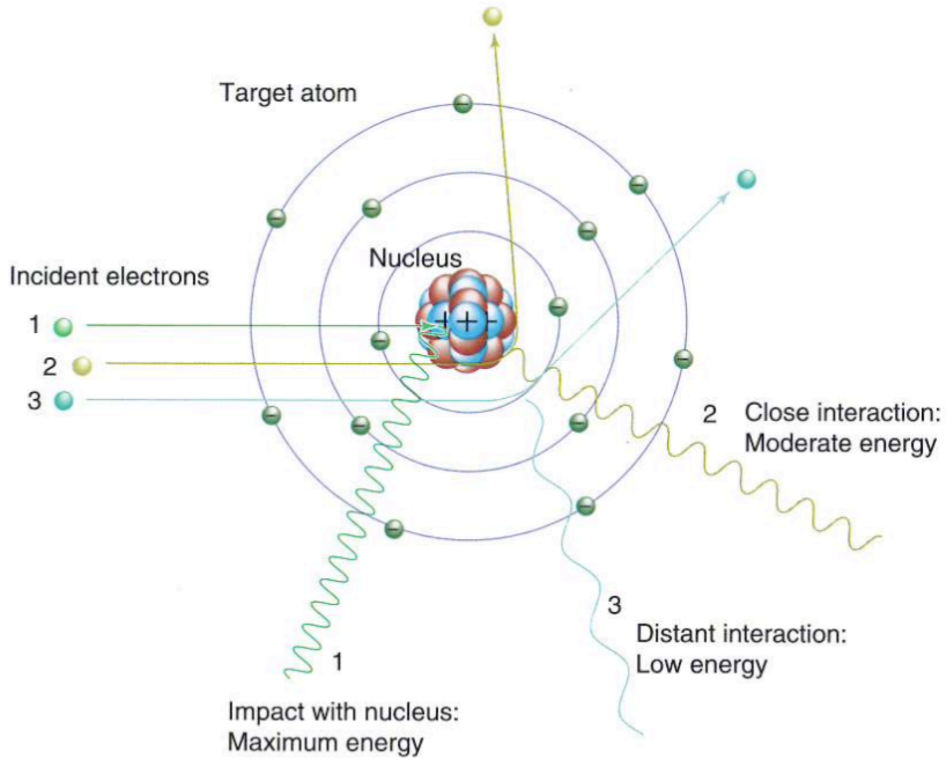
$$1\ \text{eV} = 1.602 \times 10^{-19}\ \text{J}$$

For example, kinetic energy of 1 electron through a potential of 100 kV is:
$K_{final} = q\,V$ = (1 electron charge) (100 kV) = 100 keV

Bremsstrahlung spectrum

On impact with the target the kinetic energy of electrons is converted to other form of energy. Most of the interaction are collisional and energy exchange gives rise to heat. A small fraction of the accelerated electrons comes in proximity of the atomic nucleus and it is influenced by its positive electric filed. Electrical forces attract and decelerate an electron and change its direction, causing a loss of kinetic energy, emitted as an x-ray photon of equal energy (Bremsstrahlung radiation).

The energy loss and thus the energy of the x-rays depends on the distance between the incident electron and the target nucleus (Coulomb force ~ $1/r^2$)



■ 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.

Bremsstrahlung spectrum

The total intensity of bremsstrahlung radiation (integrated over all angles and all energies) resulting from a charged particle of mass m and charge ze incident onto target nuclei with charge Ze is proportional to:

$$I_{\text{bremsstrahlung}} \propto \frac{Z^2 z^4 e^6}{m^2}. \tag{1.3}$$

The bremsstrahlung efficiency is markedly reduced if a massive particle such as a proton or alpha particle is the charged particle. Relative to an electron, protons and α particles are over 3 million times less efficient (1836^{-2}) than electrons at producing bremsstrahlung x rays. Electrons therefore become the practical choice for producing bremsstrahlung. The Z^2 term in Eq. (1.3) also indicates that bremsstrahlung production increases rapidly as the atomic number of the target increases, suggesting that high-Z targets are preferred.

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

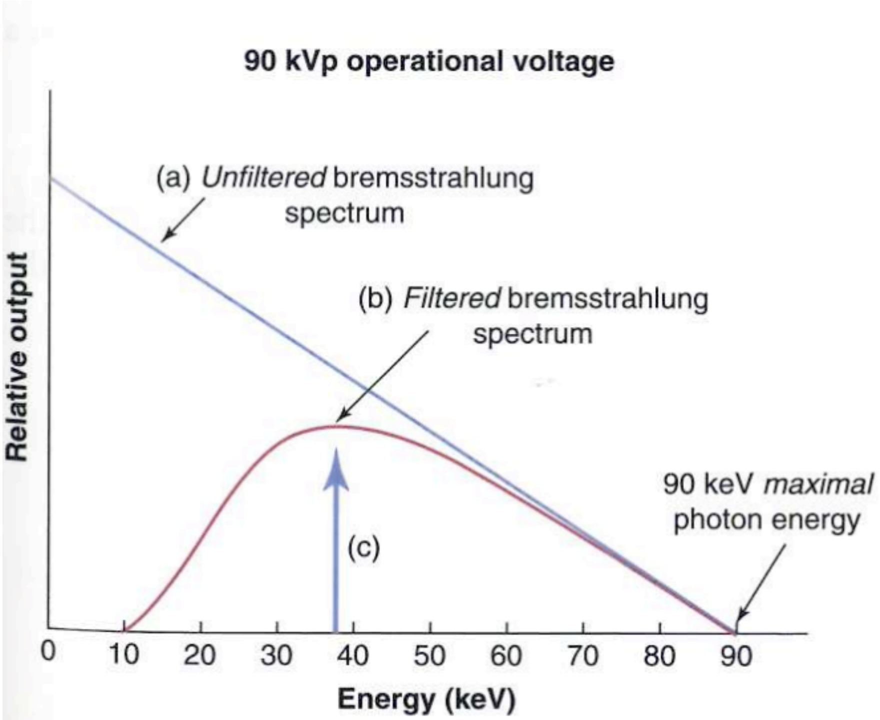


FIGURE 6-3 The bremsstrahlung energy distribution for a 90-kV acceleration potential difference. The unfiltered bremsstrahlung spectrum (a) shows a greater probability of low-energy x-ray photon production that is inversely linear with energy up to the maximum energy of 90 keV. The filtered spectrum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

Bremsstrahlung spectrum

X-ray production efficiency includes atomic number Z of the target material and the kinetic energy of the incident electrons.

The approximate ratio of the radiative energy loss caused by bremsstrahlung production to collisional (excitation and ionization) energy loss within diagnostic x-ray energy (potential difference of 20 to 150 kV) is:

$$\frac{\text{Radiative energy loss}}{\text{Collisional energy loss}} \cong \frac{E_K Z}{820{,}000}$$

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.

This equation is not applicable beyond diagnostic imaging x-ray energies

Characteristic X-ray spectrum

In addition to the continuous x-ray spectrum, discrete x-ray energy peaks (characteristic radiation) can be present depending on the anode target element and the x-ray tube voltage.
The peaks depend on the electron binding energy "characteristic" of the element.
The energy of the incident electron is determined by the x-ray tube voltage and, if exceeds the binding energy of an electron shell, can eject the electron and create a vacancy.

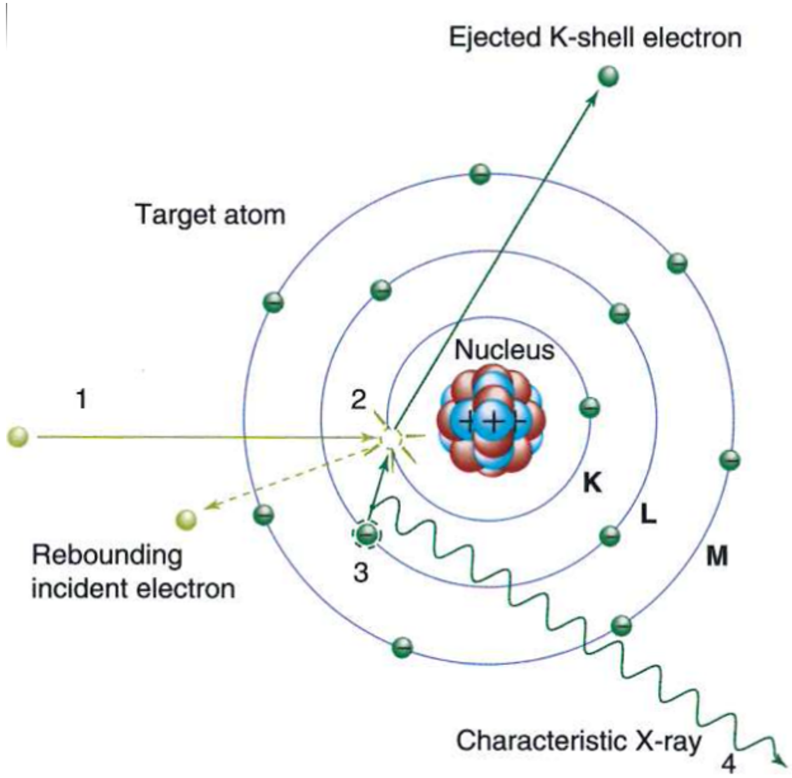


FIGURE 6-4 Generation of a characteristic x-ray in a target atom occurs in the following sequence: (1) The incident electron interacts with the *K*-shell electron via a repulsive electrical force. (2) The *K*-shell electron is removed (only if the energy of the incident electron is greater than the *K*-shell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

For tungsten , an L-shell (binding energy =10.2 keV) electron transition to fill an K-shell (binding energy =69.5 keV) vacancy produces a characteristic x-ray with a discrete energy of:
$E_bK - E_bL = 69.5\ \text{keV} - 10.2\ \text{keV} = 59.3\ \text{keV}$

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

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.

Characteristic X-ray spectrum

Electron transitions occur from adjacent and non adjacent electron shell in the atom, giving rise to several discrete characteristic energy peak 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.

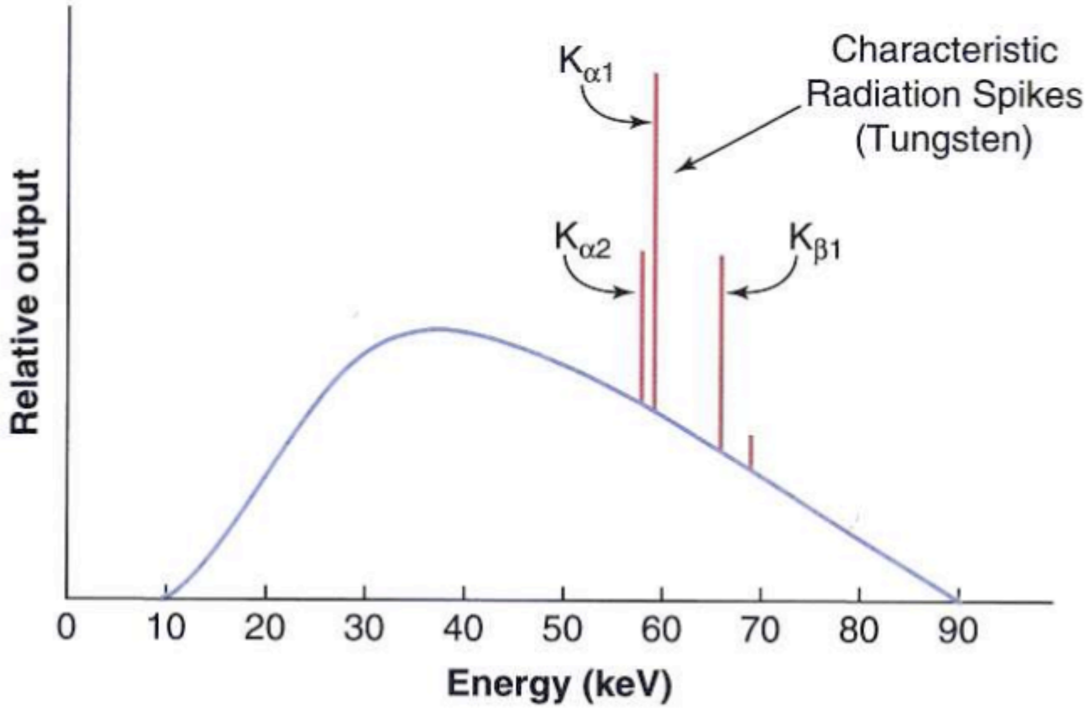
TABLE 6-2 K-SHELL CHARACTERISTIC X-RAY ENERGIES (keV) OF COMMON X-RAY TUBE TARGET MATERIALS

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.

Characteristic X-ray spectrum

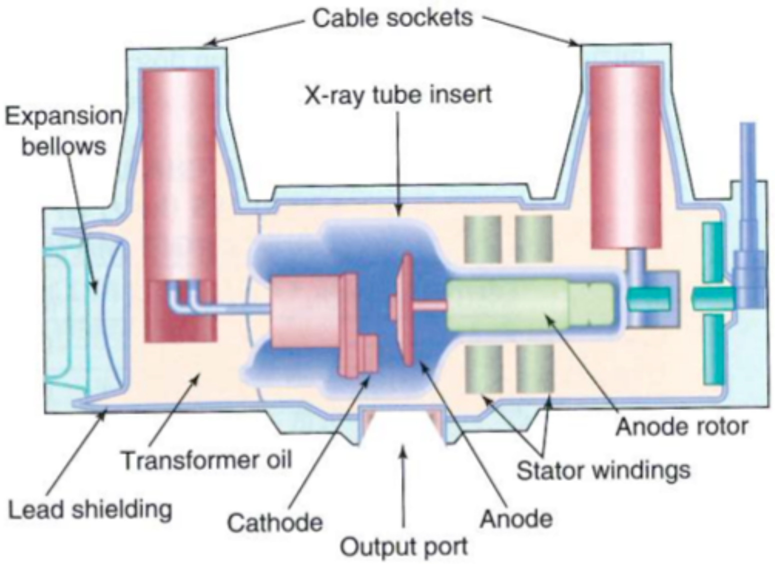
■ **FIGURE 6-5** The filtered spectrum of bremsstrahlung and characteristic radiation from a tungsten target with a potential difference of 90 kV illustrates specific characteristic radiation energies from K_α and K_β transitions. Filtration (the preferential removal of low-energy photons as they traverse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



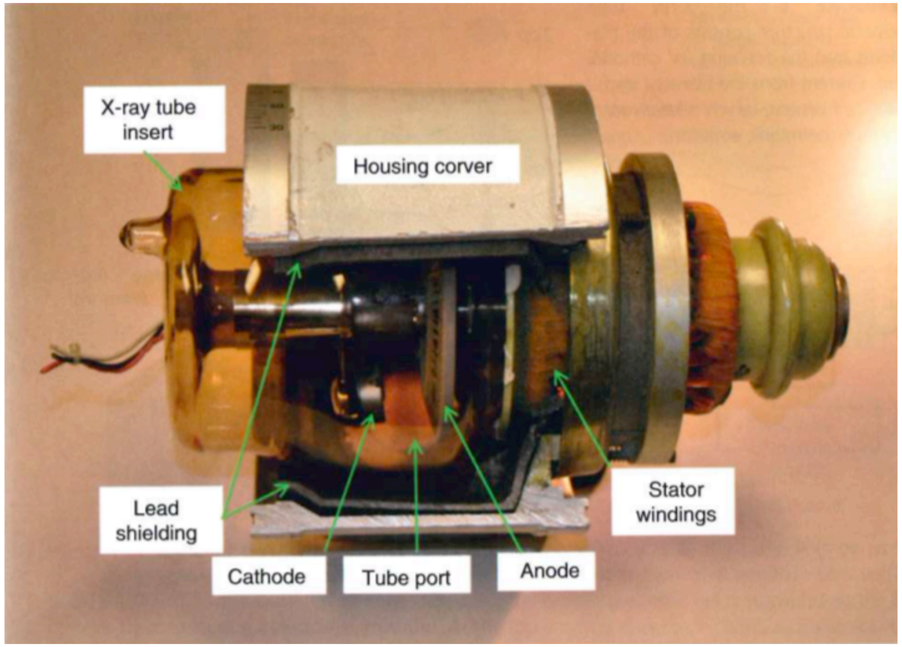
In terms of intensity, as the x-ray tube voltage increases, so does the ratio of characteristic to bremsstrahlung x-rays. For example, at 80 kV, approximately 5% of the total x-ray output intensity for a tungsten target is composed of characteristic radiation, which increases to about 10% at 100 kV. 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

X-ray tube

Major components: cathode, anode, rotor/stator, glass or metal envelope, tube port, cable sockets, tube housing. The x-ray tube generator supplies the power and permits the selection of tube voltage, tube current and exposure time. Depending on the kind of examination tube voltage is set from 40 to 150 kV for diagnostic imaging and 25 to 40 kV for mammography. The x-ray tube current, measured in mA, is proportional to the number of electrons per second flowing from the cathode to the anode. 1 mA = 6.24×10^{15} electrons/s



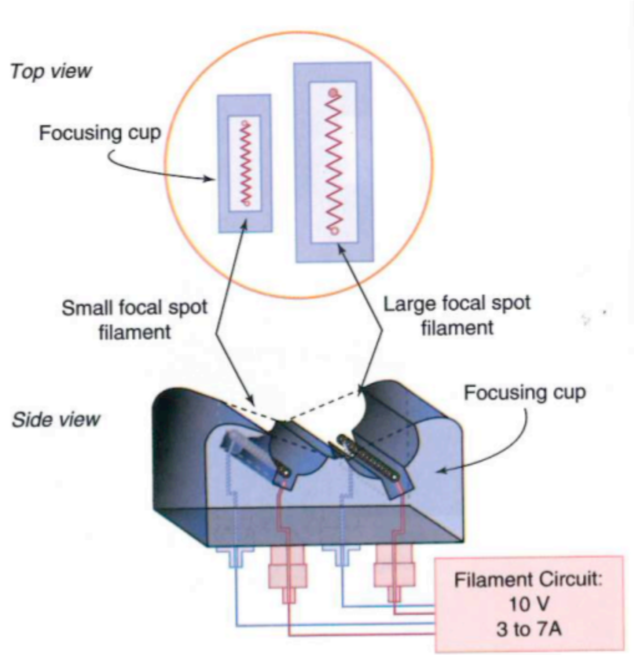
A diagram of the major components of a modern x-ray tube and housing assembly



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

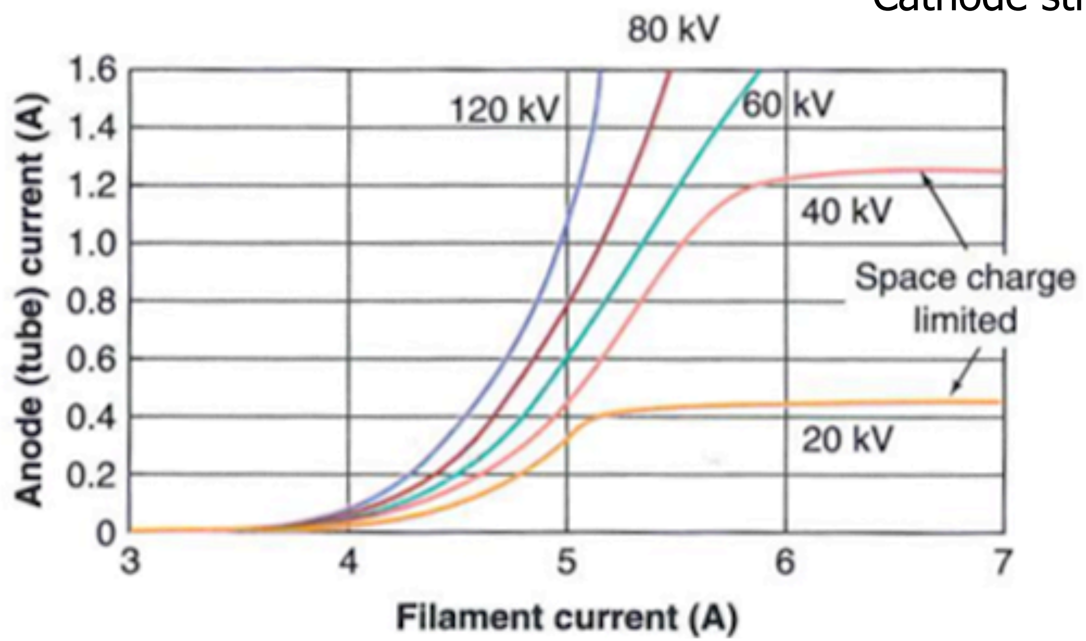
X-ray tube: cathode

Cathode is the negative electrode comprised of a filaments and a focusing cup. A filament is made of tungsten wire wound in a helix and connected to a filament circuit which provides a voltage of about 10 V and variable current up to 7 A. When energized the filament circuit heats the filament through electrical resistance and electrons are releases by thermionic emission. **Electrons flow to the anode only when the tube voltage is applied between the electrodes.** Number of electrons are adjusted by the filament current and the filament temperature.
At any kV, x-ray flux is proportional to the tube current. Higher tube voltages produce a slightly higher tube current for the same filament current



Cathode structure

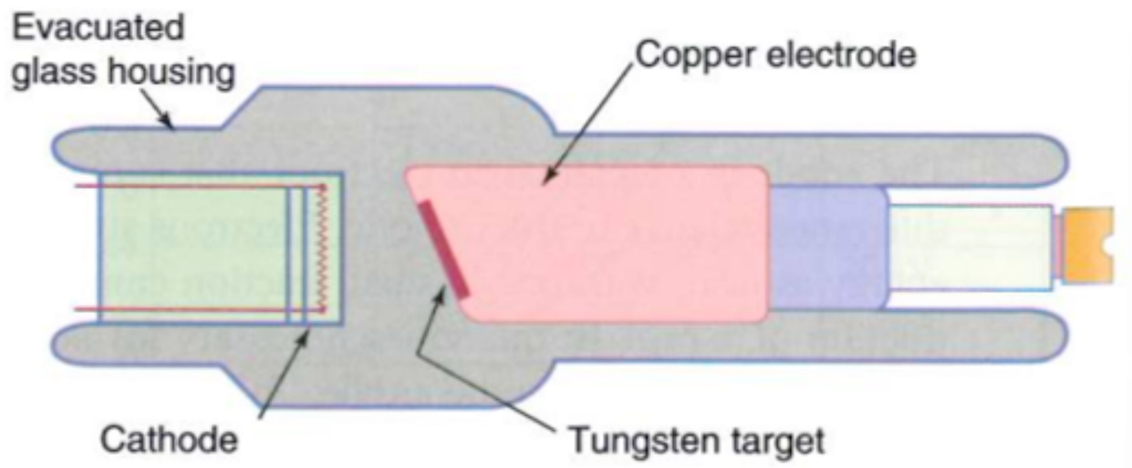
■ FIGURE 6-9 Relationship of tube current to filament current for various tube voltages shows a dependence of approximately $kV^{1.5}$. For tube voltages 40 kV and lower, a space charge cloud shields the electric field so that further increases in filament current do not increase the tube current. This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



X-ray tube: anode

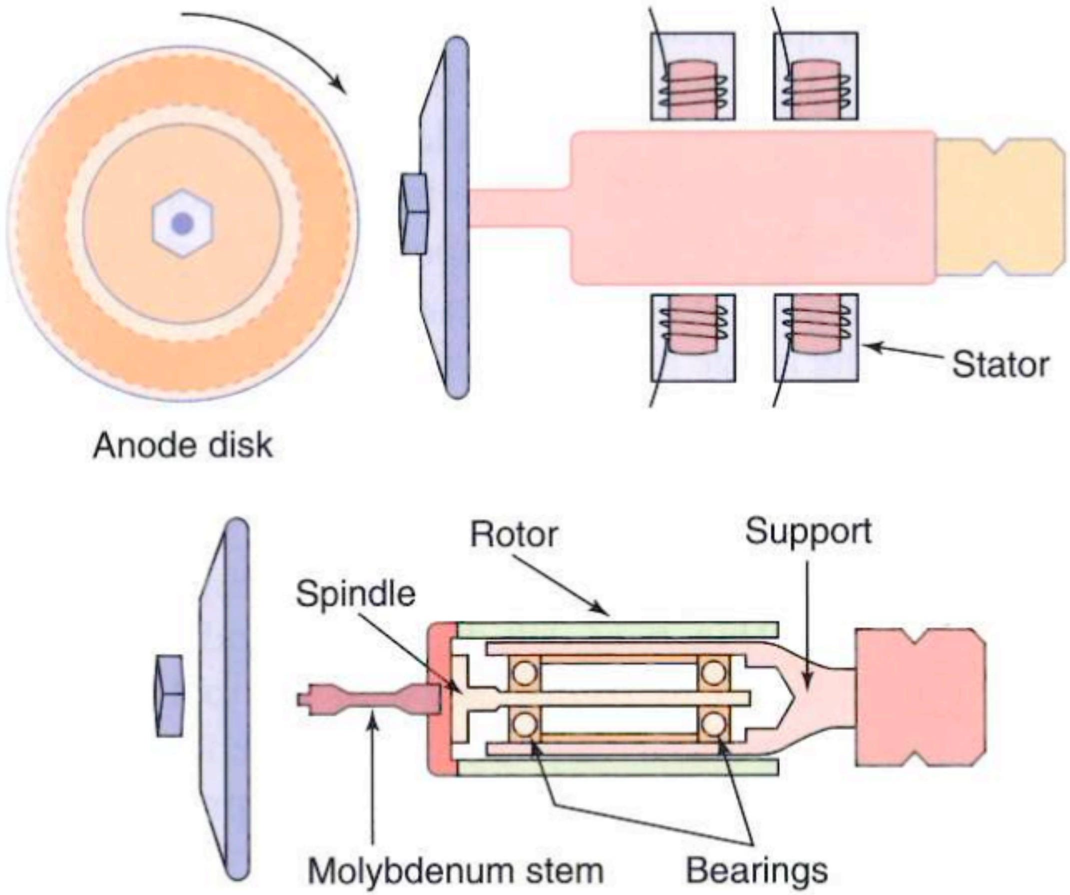
A simple x-ray tube design has a stationary anode, consisting of a tungsten insert embedded in a copper block (Fig. 6-11). Copper serves a dual role: it mechanically supports the insert and efficiently conducts heat from the tungsten target. However, the small area of the focal spot on the stationary 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.

■ 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.



X-ray tube: anode

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.



■ 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**).

Rotation speeds are:
3000 to 3600 (low speed) or 9000 to 10000 (high speed) revolutions per minute (rpm)

X-ray tube: anode

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

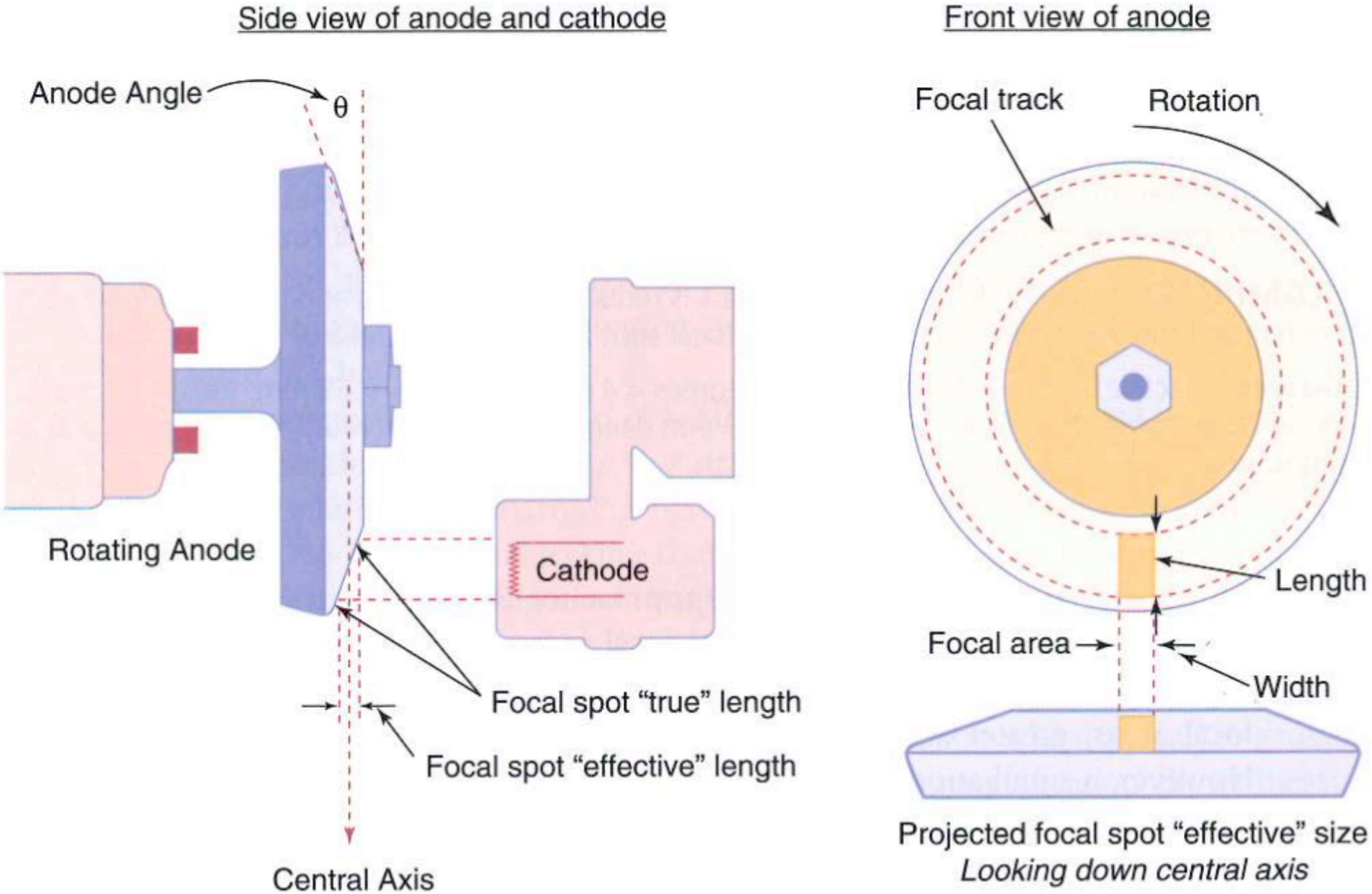


FIGURE 6-13 The anode (target) angle, θ, is defined as the angle of the target surface in relation to the central ray. The focal spot length, as projected down the central axis, is foreshortened, according to the line focus principle (**lower right**).

anode angle – focal spot size

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

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

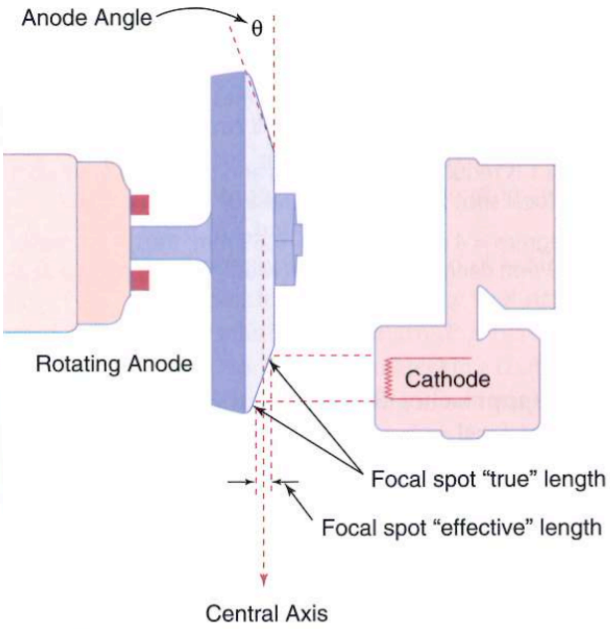
where θ is the anode angle. Foreshortening of the focal spot length at the central ray is called the *line focus principle*, as described by Equation 6-2. An ability to have a smaller effective focal spot size for a large actual focal spot increases the power loadings for smaller effective focal spot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

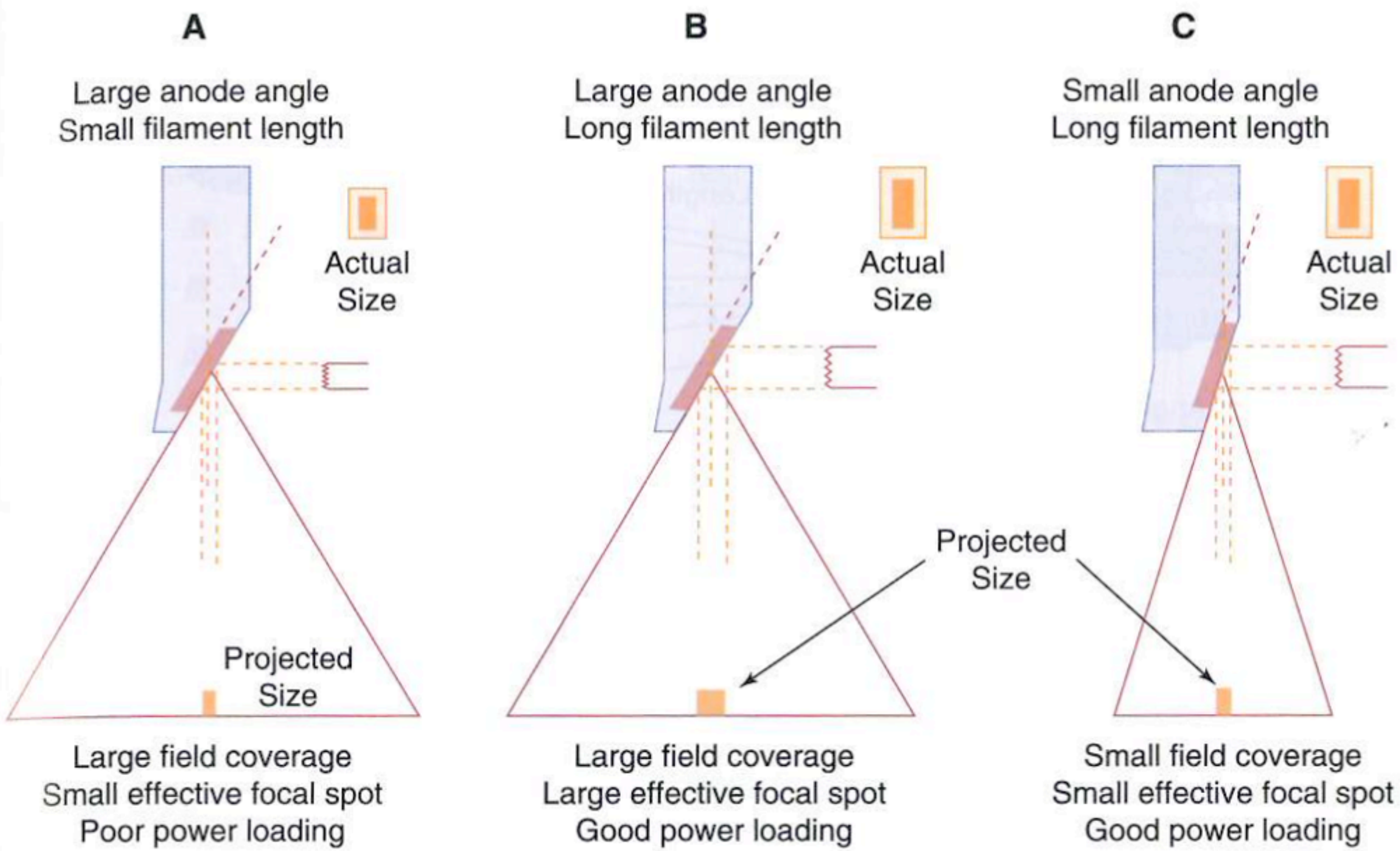
EXAMPLE 2: If the anode angle in Example 1 is reduced to 10 degrees and the actual focal spot size remains the same, what is the projected focal spot size at the central axis position?

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.



anode angle – focal spot size

As the anode angle decreases the effective focal spot becomes smaller for the same actual focal area.
Typical anode angles range from 7 to 20 degrees (most common from 12 to 15 degrees)



■ 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.

Geometry of projection radiography

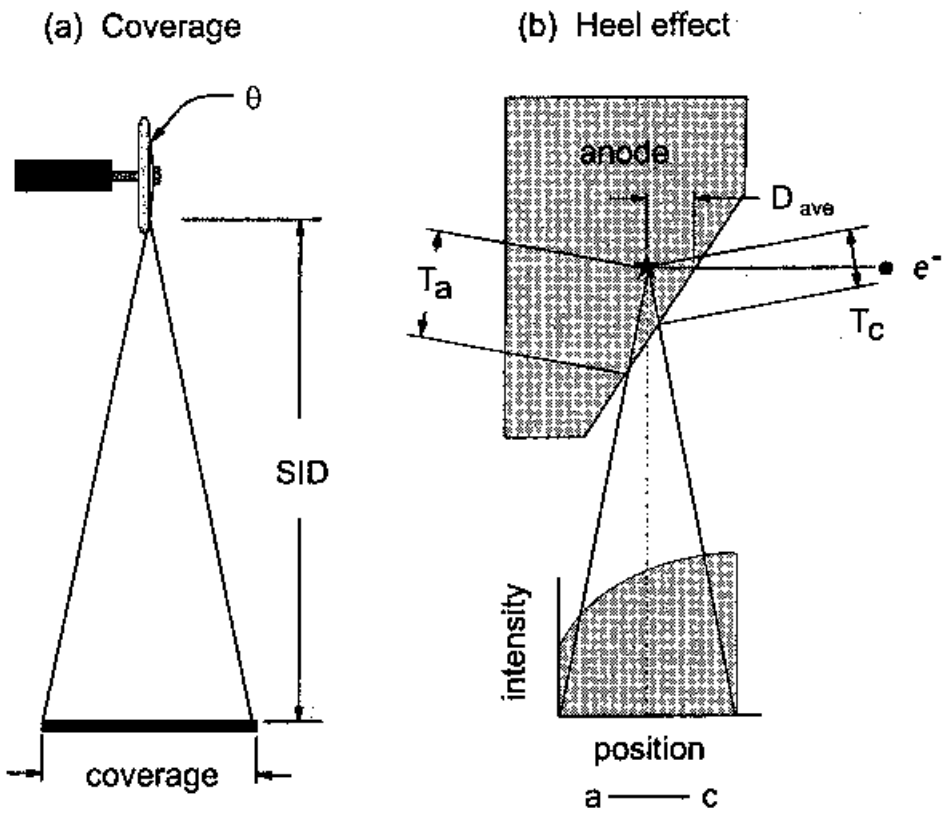


Figure 1.8: (a) Because of the line focus principle (angled anode), the field of view or *coverage* is restricted at a given source-to-image distance (SID). For instance, a 7-degree anode angle is too restrictive if 35×43 cm ($14'' \times 17''$) radiographs are to be acquired at an SID of 100 cm. (b) Another consequence of the line-focus principle is the *heel effect*. X rays are emitted at an average depth (D_{ave}) within the anode; the path length through the tungsten anode that x rays transit is different on the anode side (T_{a}) than on the cathode side (T_{c}) of the x-ray field. This difference in tungsten filtration across the field of view causes a reduction in x-ray intensity on the anode side of the x-ray field.

Geometry of projection radiography: magnification

The geometry of projection transmission imaging is described in Figure 7-2. Magnification can be defined simply as

$$M = \frac{L_{image}}{L_{object}} \qquad [7\text{-}1]$$

where L_{image} is the length of the object as seen on the image and L_{object} is the length of the actual object. Due to similar triangles, the object magnification can be computed using the source to object distance (a) and the object to detector distance (b)

$$M = \frac{a+b}{a} \qquad [7\text{-}2]$$

The magnification will always be greater than 1.0 but approaches 1.0 when a relatively flat object (such as a hand in radiography) is positioned in contact with the detector, where $b \approx 0$. The magnification factor changes slightly for each plane perpendicular to the x-ray beam axis in the patient.

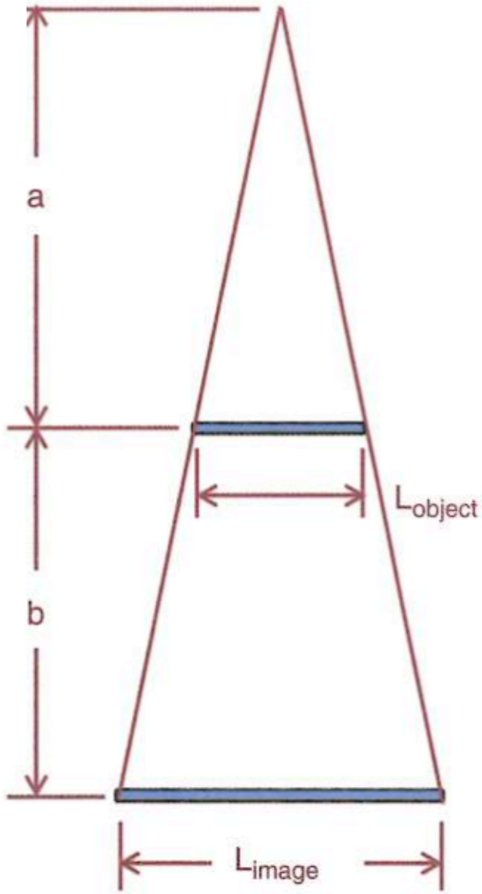


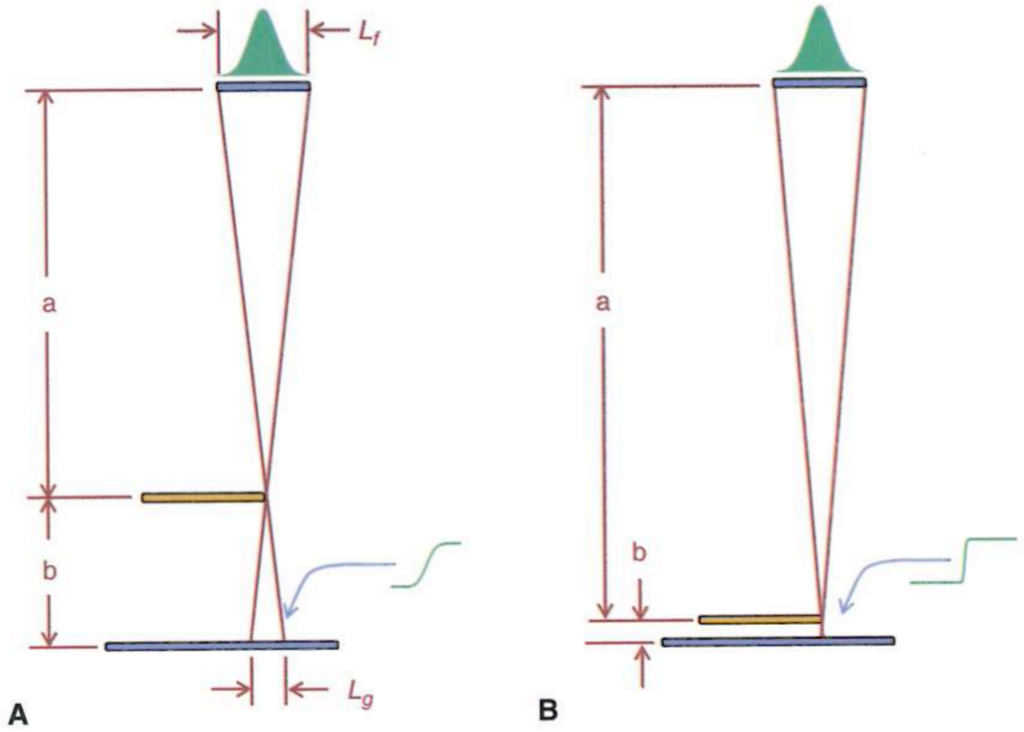
FIGURE 7-2 The geometry of beam divergence is shown. The object is positioned a distance *a* from the x-ray source, and the detector is positioned a distance (*a* + *b*) from the source. From the principle of similar triangles, $m = \frac{L_{image}}{L_{object}} = \frac{a+b}{a}$

Geometry of projection radiography: penumbra

The focal spot in the x-ray tube is very small but is not truly a point source, resulting in magnification-dependent resolution loss. The blurring from a finite x-ray source is dependent upon the geometry of the exam, as shown in Figure 7-3. The length of the edge gradient (L_g) is related to the length of the focal spot (L_f) by

$$L_g = L_f \frac{b}{a},$$

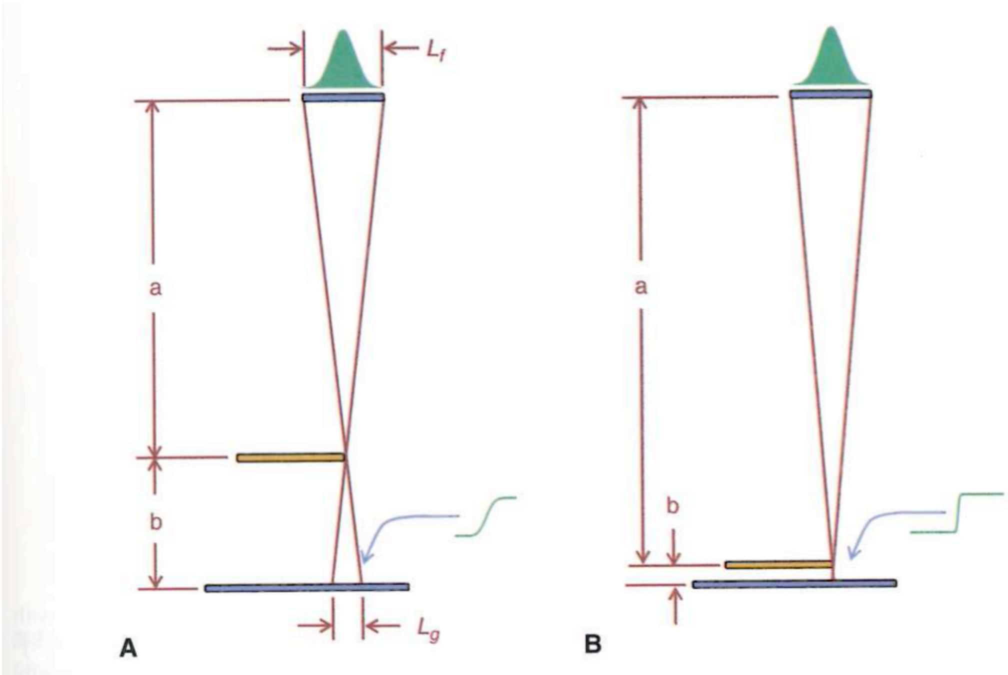
b/a is the magnification of the focal spot



■ FIGURE 7-3 The magnification of a sharp edge in the patient will result in some blurring of that structure because of the use of a focal spot that is not truly a point source. For a focal spot with width L_f, the intensity distribution across this distributed (nonpoint) source will result in the edge being projected onto the image plane. The edge will no longer be perfectly sharp, but rather its shadow will reflect the source distribution. Most sources are approximately gaussian in shape, and thus the blurred profile of the edge will also appear gaussian. The length of the blur in the image is related to the width of the focal spot by $L_g = \frac{b}{a} L_f$.

Geometry of projection radiography: penumbra

The "edge gradient" is measured using a highly magnified metal foil with a sharp edge. In most circumstances, higher object magnification increases the width of the edge gradient and reduces the spatial resolution of the image. Consequently, in most cases, the patient should be positioned as close as possible to the detector to reduce magnification. For thin objects that are placed in contact with the detector, the magnification is approximately 1.0, and there will be negligible blurring caused by the finite dimensions of the focal spot. In some settings (e.g., mammography), a very small focal spot is used intentionally with magnification.



Factors affecting x-ray emission

The output of the x-ray beam is described in terms of quality (penetrability of the x-ray beam with higher x-ray photons) and quantity (number of photons comprising the beam)

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 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.

$$\text{Exposure} \propto \text{kV}^2$$

Factors affecting x-ray emission

For example 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.*

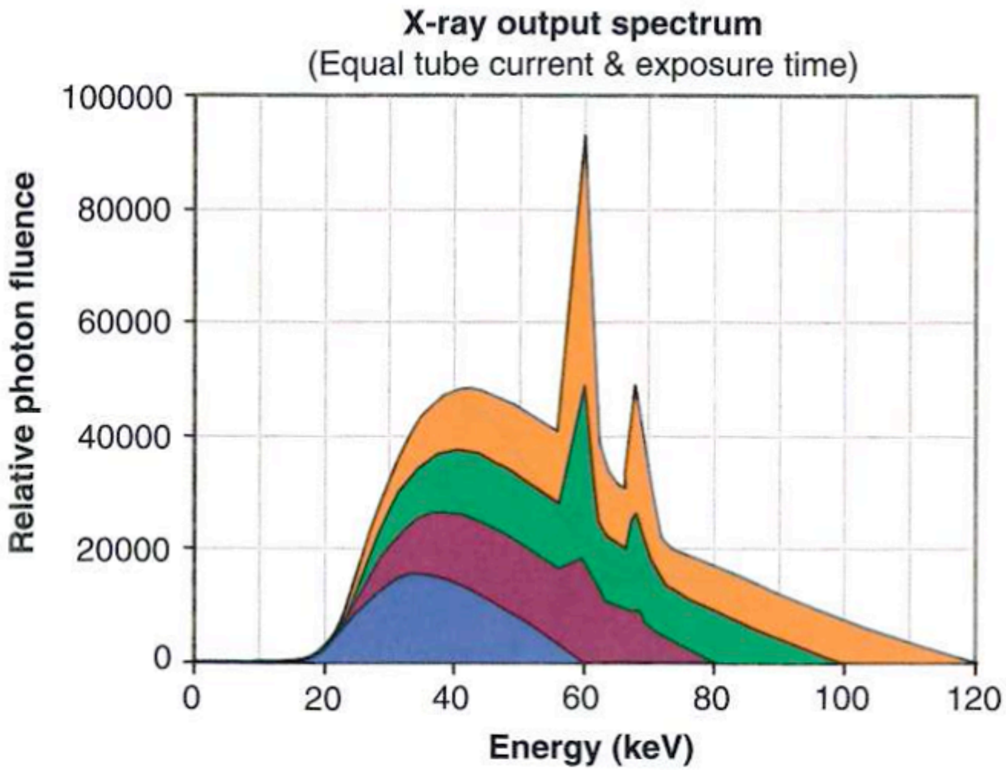


FIGURE 6-30 x-Ray tube output intensity varies as the square of tube voltage (kV). In this example, the same tube current and exposure times (mAs) are compared for 60 to 120 kV. The relative area under each spectrum roughly follows a squared dependence (characteristic radiation is ignored).

Factors affecting x-ray emission

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.

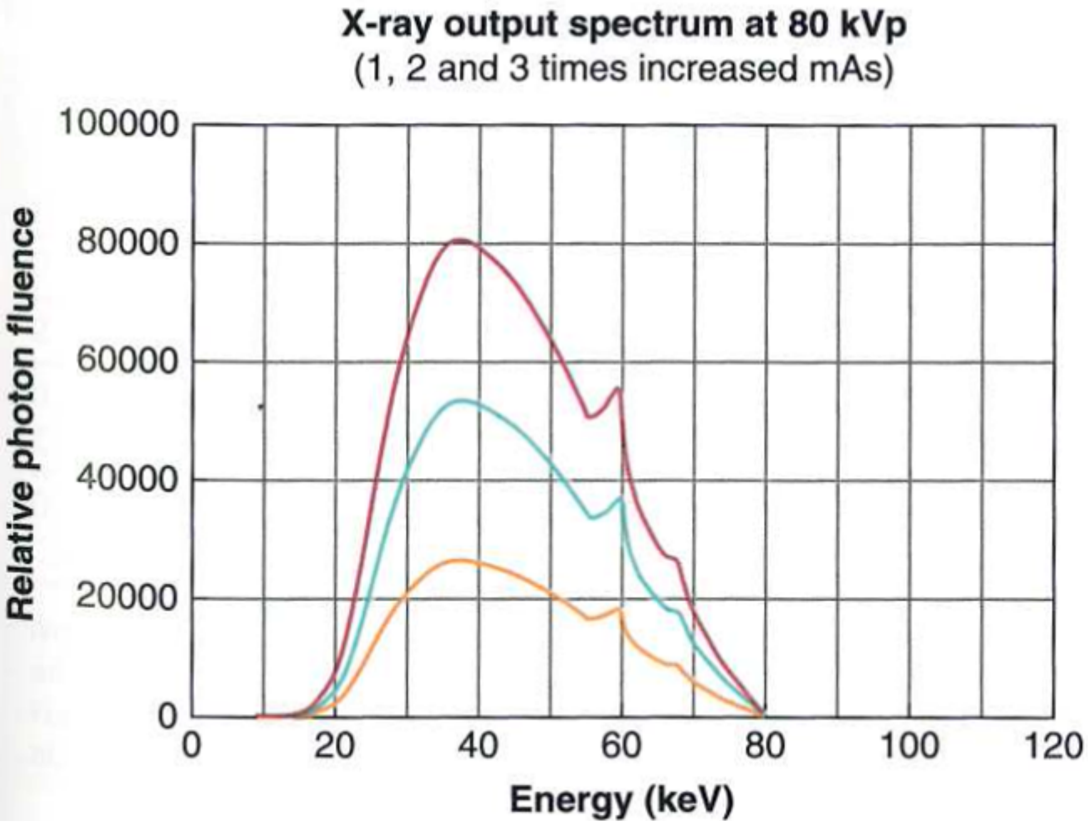
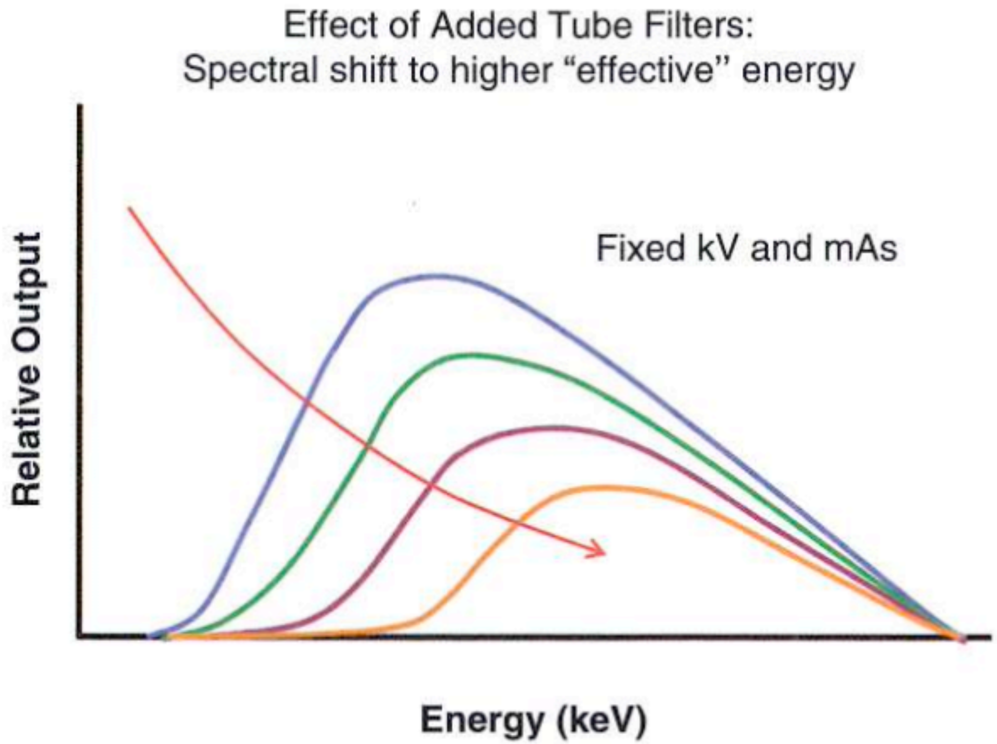


FIGURE 6-31 X-ray tube output intensity is proportional to the mAs (tube current and exposure time). Shown is the result of increasing the mAs by a factor of two and three, with a proportional change in the number of x-rays produced.

Factors affecting x-ray emission

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).

■ 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.

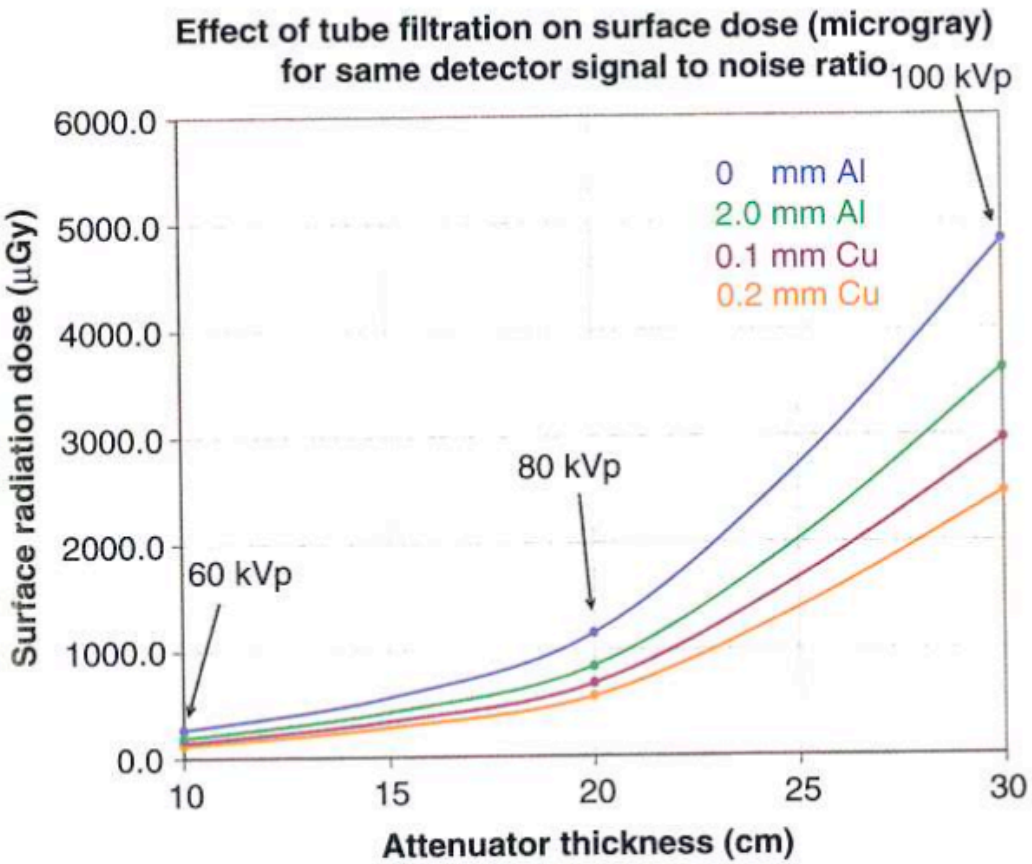


Filtration

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).

Filtration



■ 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.