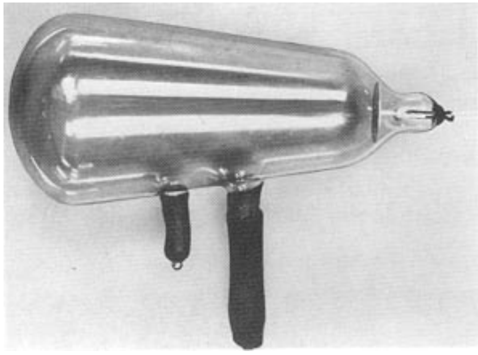
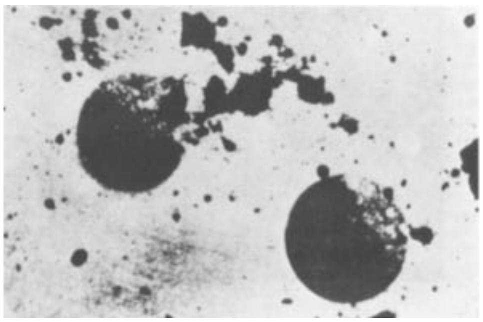
X rays were discovered on November 8, 1895 by Wilhelm C. Röntgen, a physicist at the University of Würzburg in Germany.[1] He named his discovery "x rays" because "x" stands for an unknown quantity. For this work, Röntgen received the first Nobel Prize in Physics in 1901.



MARGIN FIGURE 5-1
Crookes tube, an early example of a cathode ray tube.

Röntgen was not the first to acquire an x-ray photograph. In 1890 Alexander Goodspeed of the University of Pennsylvania, with the photographer William Jennings, accidentally exposed some photographic plates to x rays. They were able to explain the images on the developed plates only after Röntgen announced his discovery of x rays.



MARGIN FIGURE 5-2
An x-ray picture obtained accidentally in February 1890 by Arthur Goodspeed of the University of Pennsylvania. The significance of this picture, acquired more than 5 years before Röntgen's discovery, was not recognized by Professor Goodspeed.

Heating a filament to release electrons is called thermionic emission or the Edison effect.

Coolidge's contributions to x-ray science, in addition to the heated filament, included the focusing cup, imbedded x-ray target, and various anode cooling devices.

OBJECTIVES

By studying this chapter, the reader should be able to:

- Identify each component of an x-ray tube and explain its function.
- Describe single- and three-phase voltage, and various modes of voltage rectification.
- Explain the shape of the x-ray spectrum, and identify factors that influence it.
- Discuss concepts of x-ray production, including the focal spot, line-focus principle, space charge, power deposition and emission, and x-ray beam hardening.
- Define and apply x-ray tube rating limits and charts.

INTRODUCTION

To produce medical images with x rays, a source is required that:

1. Produces enough x rays in a short time
2. Allows the user to vary the x-ray energy
3. Provides x rays in a reproducible fashion
4. Meets standards of safety and economy of operation

Currently, the only practical sources of x rays are radioactive isotopes, nuclear reactions such as fission and fusion, and particle accelerators. Only special-purpose particle accelerators known as x-ray tubes meet all the requirements mentioned above. In x-ray tubes, bremsstrahlung and characteristic x rays are produced as high-speed electrons interact in a target. While the physical design of x-ray tubes has been altered significantly over a century, the basic principles of operation have not changed.

Early x-ray studies were performed with a cathode ray tube in which electrons liberated from residual gas atoms in the tube were accelerated toward a positive electrode (anode). These electrons produced x rays as they interacted with components of the tube. The cathode ray tube was an unreliable and inefficient method of producing x rays. In 1913, Coolidge[2] improved the x-ray tube by heating a wire filament with an electric current to release electrons. The liberated electrons were repelled by the negative charge of the filament (cathode) and accelerated toward a positive target (the anode). X rays were produced as the electrons struck the target. The Coolidge tube was the prototype for "hot cathode" x-ray tubes in wide use today.

CONVENTIONAL X-RAY TUBES

Figure 5-1 shows the main components of a modern x-ray tube. A heated filament releases electrons that are accelerated across a high voltage onto a target. The stream of accelerated electrons is referred to as the *tube current*. X rays are produced as the electrons interact in the target. The x rays emerge from the target in all directions but are restricted by collimators to form a useful beam of x rays. A vacuum is maintained inside the glass envelope of the x-ray tube to prevent the electrons from interacting with gas molecules.

ELECTRON SOURCE

A metal with a high melting point is required for the filament of an x-ray tube. Tungsten filaments (melting point of tungsten 3370°C) are used in most x-ray tubes. A current of a few amperes heats the filament, and electrons are liberated at a rate that increases with the filament current. The filament is mounted within a negatively charged focusing cup. Collectively, these elements are termed the *cathode assembly*.

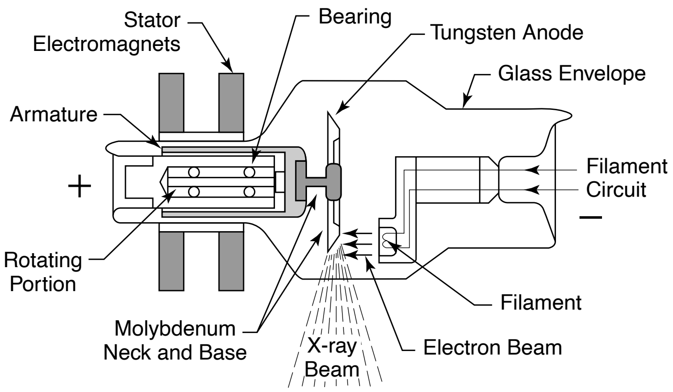


FIGURE 5-1
Simplified x-ray tube with a rotating anode and a heated filament.

Vaporized tungsten from both the filament and anode deposits on the glass envelope of the x-ray tube, giving older tubes a mirrored appearance.

An x-ray tube with two filaments is called a *dual-focus* tube.

MARGIN FIGURE 5-3
A dual-focus x-ray tube with a rotating anode.

The focal spot is the volume of target within which electrons are absorbed and x rays are produced. For radiographs of highest clarity, electrons should be absorbed within a small focal spot. To achieve a small focal spot, the electrons should be emitted from a small or "fine" filament. Radiographic clarity is often reduced by voluntary or involuntary motion of the patient. This effect can be decreased by using x-ray exposures of high intensity and short duration. However, these high-intensity exposures may require an electron emission rate that exceeds the capacity of a small filament. Consequently many x-ray tubes have two filaments. The smaller, fine filament is used when radiographs with high detail are desired and short, high-intensity exposures are not necessary. If high-intensity exposures are needed to limit the blurring effects of motion, the larger, coarse filament is used. The cathode assembly of a dual-focus x-ray tube is illustrated in Margin Figure 5-4.

TUBE VOLTAGE AND VOLTAGE WAVEFORMS

The intensity and energy distribution of x rays emerging from an x-ray tube are influenced by the potential difference (voltage) between the filament and target of the tube. The source of electrical power for radiographic equipment is usually alternating current (ac). This type of electricity is by far the most common form available for general use, because it can be transmitted with little energy loss through power lines that span large distances. Figure 5-2 shows a graph of voltage and current in an ac power line. X-ray tubes are designed to operate at a single polarity, with a positive

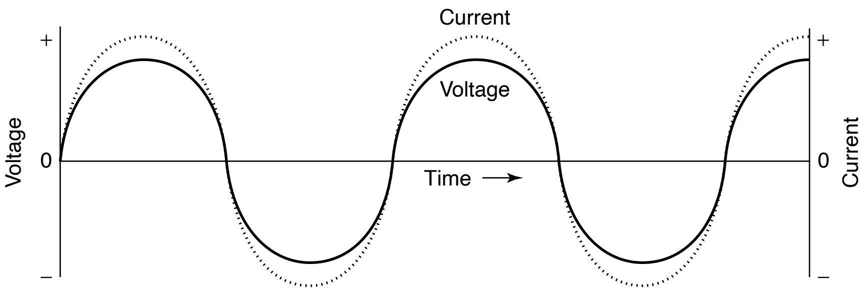
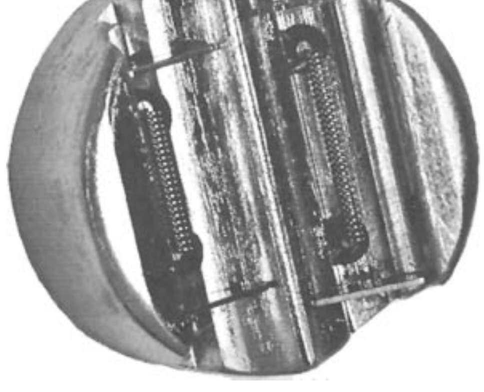


FIGURE 5-2
Voltage and current in an ac power line. Both voltage and current change from positive to negative over time. The relationship between voltage and current (i.e., the relative strength and the times at which they reach their peaks) depends upon a complex quantity called reactance. The positive and negative on the voltage scale refer to polarity, while the positive and negative on the current scale refer to the direction of flow of electrons that constitute an electric current.



MARGIN FIGURE 5-4
Cathode assembly of a dual-focus x-ray tube. The small filament provides a smaller focal spot and a radiograph with greater detail, provided that the patient does not move. The larger filament is used for high-intensity exposures of short duration.

A positive electrode is termed an *anode* because negative ions (anions) are attracted to it. A negative electrode is referred to as a cathode because positive ions (cations) are attracted to it.

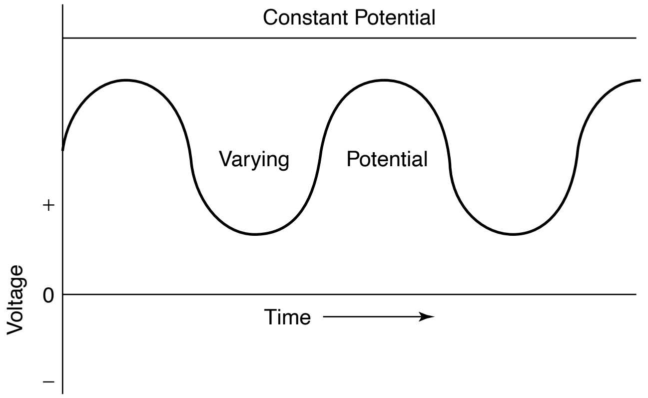
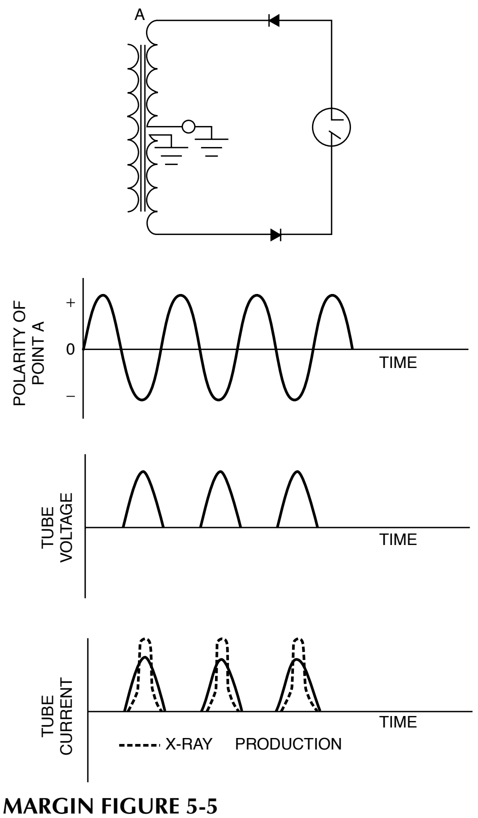


FIGURE 5-3
Voltage in a dc power line. The term *direct current* means that the voltage (and current not shown here) never reverse (change from positive to negative), although they may vary in intensity. X-ray tubes are most efficient when operated at constant potential.

The term *alternating* means that the voltage reverses polarity at some frequency (in the United States, 120 reversals per second (60 Hz) alternating current is standard).

target (anode) and a negative filament (cathode). X-ray production is most efficient (more x rays are produced per unit time) if the potential of the target is always positive and if the voltage between the filament and target is kept at its maximum value. In most x-ray equipment, ac is converted to direct current (dc), and the voltage between filament and target is kept at or near its maximum value (Figure 5-3). The conversion of ac to dc is called *rectification*.

One of the simplest ways to operate an x-ray tube is to use ac power and rely upon the x-ray tube to permit electrons to flow only from the cathode to the anode. The configuration of the filament (a thin wire) is ideal for producing the heat necessary to release electrons when current flows through it. Under normal circumstances the target (a flat disk) is not an efficient source of electrons. When the polarity is reversed (i.e., the filament is positive and the target is negative), current cannot flow in the x-ray tube, because there is no source of electrons. In this condition, the x-ray tube "self-rectifies" the ac power, and the process is referred to as *self-rectification*. At high tube currents, however, the heat generated in the target can be great enough to release electrons from the target surface. In this case, electrons flow across the x-ray tube when the target is negative and the filament is positive. This reverse flow of electrons can destroy the x-ray tube.



MARGIN FIGURE 5-5
A circuit for half-wave rectification (**top**), with resulting tube voltage, tube current, and efficiency for production of x rays. Rectifiers indicate the direction of conventional current flow, which is opposite to the actual flow of electrons.

A rectified voltage waveform can also be attained by use of circuit components called diodes. Diodes are devices that, like x-ray tubes, allow current to flow in only one direction. A simple circuit containing diodes that produces the same waveform as *self-rectification* is shown in the margin. Rectification in which polarity reversal across the x-ray tube is eliminated is called *half-wave rectification*.

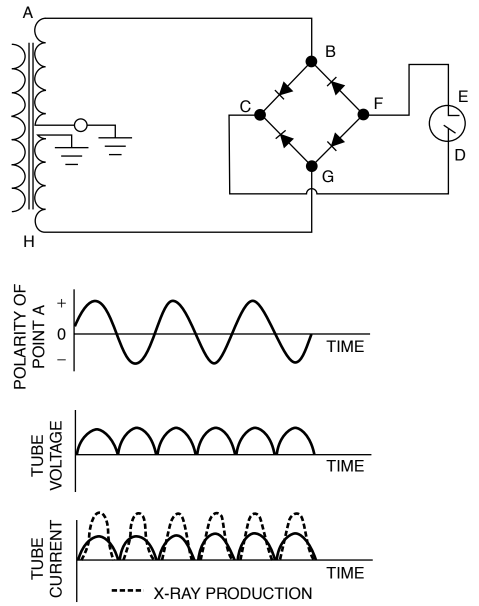
A half-wave rectifier converts ac to a dc waveform with 1 pulse per cycle. X-ray production could be made more efficient if the negative half-cycle of the voltage waveform could be used. A more complex circuit called a full-wave rectifier utilizes both half-cycles. In both the positive and negative phases of the voltage waveform, the voltage is impressed across the x-ray tube with the filament (or cathode) at a negative potential and the target (or anode) at a positive potential. This method of rectifying the ac waveform is referred to as *full-wave rectification*. In full-wave rectification the negative pulses in the voltage waveform are in effect "flipped over" so that they can be used by the x-ray tube to produce x rays. Thus a full-wave rectifier converts an ac waveform into a dc waveform having 2 pulses per cycle.

The efficiency of x-ray production could be increased further if the voltage waveform were at high potential most of the time, rather than decreasing to zero at least twice per cycle as it does in full-wave rectification. This goal can be achieved by use of three-phase (3φ) power. Three-phase power is provided through three separate voltage lines connected to the x-ray tube.

The term *phase* refers to the fact that all three voltage lines carry the same voltage waveform, but the voltage peaks at different times in each line. Each phase (line) is rectified separately so that three distinct (but overlapping) full-wave-rectified waveforms are presented to the x-ray tube. The effect of this composite waveform is to supply voltage to the x-ray tube that is always at or near maximum. In a three-phase full-wave-rectified x-ray circuit the voltage across the x-ray tube never drops to zero. With three separate phases of ac, six rectified pulses are provided during each voltage cycle.

A refinement of 3φ circuitry provides a slight phase shift for the waveform presented to the anode compared with that presented to the cathode. This refinement yields 12 pulses per cycle and provides a slight increase in the fraction of time that the x-ray tube operates near peak potential.

Modern solid-state voltage-switching devices are capable of producing "high-frequency" waveforms yielding thousands of x-ray pulses per second. These voltage waveforms are essentially constant potential and provide further improvements in the efficiency of x-ray production.



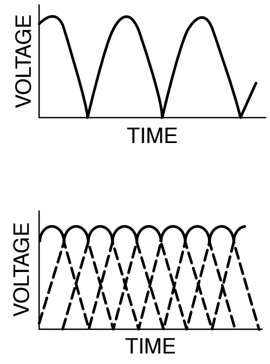
MARGIN FIGURE 5-6
A circuit for full-wave rectification (**top**), with resulting tube voltage, tube current, and efficiency for production of x rays. Electrons follow the path ABFEDCGH when end A of the secondary of the high-voltage transformer is negative. When the voltage across the secondary reverses polarity, the electron path is HGFEDCBA.

RELATIONSHIP BETWEEN FILAMENT CURRENT AND TUBE CURRENT

Two electrical currents flow in an x-ray tube. The *filament current* is the flow of electrons through the filament to raise its temperature and release electrons. The second electrical current is the flow of released electrons from the filament to the anode across the x-ray tube. This current, referred to as the *tube current,* varies from a few to several hundred milliamperes.

The two currents are separate but interrelated. One of the factors that relates them is the concept of "space charge." At low tube voltages, electrons are released from the filament more rapidly than they are accelerated toward the target. A cloud of electrons, termed the *space charge,* accumulates around the filament. This cloud opposes the release of additional electrons from the filament.

The curves in Figure 5-4 illustrate the influence of tube voltage and filament current upon tube current. At low filament currents, a saturation voltage is reached above which the current through the x-ray tube does not vary with increasing voltage. At the saturation voltage, tube current is limited by the rate at which electrons are released from the filament. Above the saturation voltage, tube current can be increased only by raising the filament's temperature in order to increase the rate of electron emission. In this situation, the tube current is said to be *temperature or filament-emission limited*. To obtain high tube currents and x-ray energies useful for diagnosis, high filament currents and voltages between 40 and 140 kV must be used. With high filament currents and lower tube voltages, the space charge limits the tube current, and hence the x-ray tube is said to be *space-charge limited.*

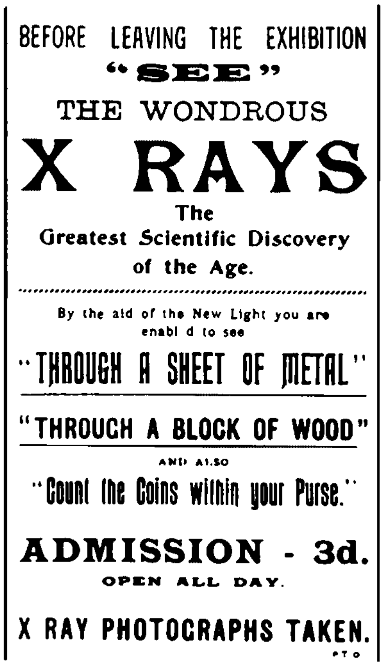


MARGIN FIGURE 5-7
Single-phase (**top**) and three-phase (**bottom**) voltages across an x-ray tube. Both voltages are full-wave rectified. The three-phase voltage is furnished by a six-pulse circuit.

EMISSION SPECTRA

The useful beam of an x-ray tube is composed of photons with an energy distribution that depends on four factors:

- Bremsstrahlung x rays are produced with a range of energies even if electrons of a single energy bombard the target.
- X rays released as characteristic radiation have energies independent of that of the bombarding electrons so long as the energy of the bombarding electrons exceeds the threshold energy for characteristic x ray emission.
- The energy of the bombarding electrons varies with tube voltage, which fluctuates rapidly in some x-ray tubes.



MARGIN FIGURE 5-8
Poster for a public demonstration of x rays, 1896, Crystal Place Exhibition, London.

X-ray generators that yield several thousand voltage (and hence x ray) pulses per second are known as *constant potential generators*.

X-ray tubes can operate in one of two modes:

- filament-emission limited
- space-charge limited

As a rough approximation, the rate of production of x rays is proportional to $Z_{\text{target}} \times (\text{kVp})^2 \times \text{mA}$.

Inherent filtration is also referred to as *intrinsic filtration.*

Mammography x-ray tubes often employ exit windows made of beryllium to allow low-energy x rays to escape from the tube.

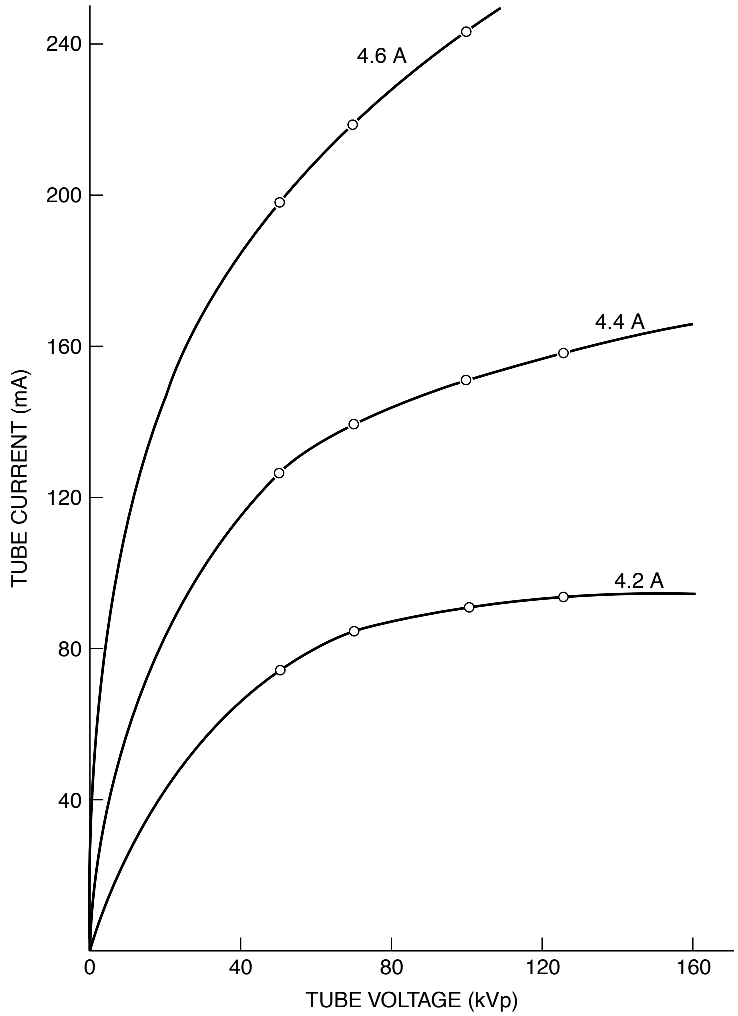


FIGURE 5-4
Influence of tube voltage and filament current upon electron flow in a Machlett Dynamax x-ray tube with a rotating anode, 1-mm apparent focal spot, and full-wave-rectified voltage.

- X rays are produced at a range of depths in the target of the x-ray tube. These x rays travel through different thicknesses of target and may lose energy through one or more interactions.

Changes in other variables such as filtration, target material, peak tube voltage, current, and exposure time all may affect the range and intensity of x-ray energies in the useful beam. The distribution of photon energies produced by a typical x-ray tube, referred to as an *emission spectrum,* is shown in Figure 5-5.

FILTRATION

An x-ray beam traverses several attenuating materials before it reaches the patient, including the glass envelope of the x-ray tube, the oil surrounding the tube, and the exit window in the tube housing. These attenuators are referred to collectively as the inherent filtration of the x-ray tube (Table 5-1). The aluminum equivalent for each component of inherent filtration is the thickness of aluminum that would reduce the exposure rate by an amount equal to that provided by the component. The inherent filtration is approximately 0.9 mm Al equivalent for the tube described in Table 5-1, with most of the inherent filtration contributed by the glass envelope. The *inherent filtration* of most x-rays tubes is about 1 mm Al.

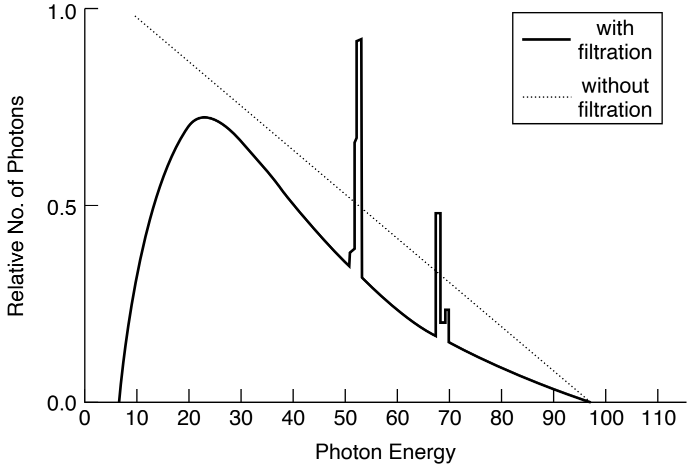


FIGURE 5-5
Emission spectrum for a tungsten target x-ray tube operated at 100 kVp. K-characteristic x-ray emission occurs for tungsten whenever the tube voltage exceeds 69 keV, the K-shell binding energy for tungsten. The *dotted line* represents the theoretical bremsstrahlung emission from a tungsten target. The *solid line* represents the spectrum after self-, inherent, and added filtration. The area under the spectrum represents the total number of x rays.

An x-ray beam of higher average energy is said to be "harder" because it is able to penetrate more dense (i.e., harder) substances such as bone. An x-ray beam of lower average energy is said to be "softer" because it can penetrate only less dense (i.e., softer) substances such as fat and muscle.

In any medium, the probability that incident x rays interact photoelectrically varies roughly as $1/E^3$, where E is the energy of the incident photons (see Chapter 4). That is, low-energy x rays are attenuated to a greater extent than those of high energy. After passing through a material, an x-ray beam has a higher average energy per photon (that is, it is "harder") even though the total number of photons in the beam has been reduced, because more low-energy photons than high-energy photons have been removed from the beam.

The inherent filtration of an x-ray tube "hardens" the x-ray beam. Additional hardening may be achieved by purposefully adding filters of various composition to the beam. The total filtration in the x-ray beam is the sum of the inherent and added filtration as shown in Table 5-1. Usually, additional hardening is desirable because the filter removes low-energy x rays that, if left in the beam, would increase the radiation dose to the patient without contributing substantially to image formation.

Emission spectra for a tungsten-target x-ray tube are shown in Figure 5-6 for various thicknesses of added aluminum filtration. The effect of the added aluminum is to decrease the total number of photons but increase the average energy of photons in the beam. These changes are reflected in a decrease in the overall height of the emission spectrum and a shift of the peak of the spectrum toward higher energy.

Equalization filters are sometimes used in chest and spine imaging to compensate for the large differences in x-ray transmission between the mediastinum and lungs.

Tube Voltage

As the energy of the electrons bombarding the target increases, the high-energy limit of the x-ray spectrum increases correspondingly. The height of the spectrum also

TABLE 5-1 Contributions to Inherent Filtration in Typical Diagnostic X-Ray Tube

Component	*Thickness (mm)*	*Aluminum-Equivalent Thickness (mm)*
Glass envelope	1.4	0.78
Insulating oil	2.36	0.07
Bakelite window	1.02	0.05

[a] Data from Trout, E. *Radiol. Technol.* 1963; **35**:161.

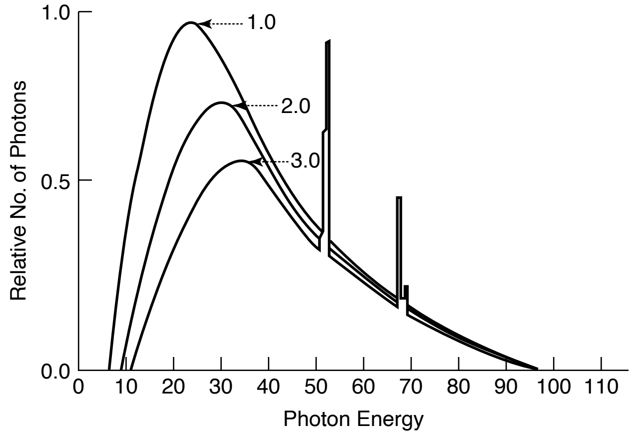
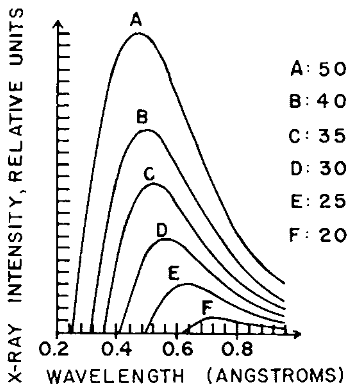


FIGURE 5-6
X-ray emission spectra for a 100-kVp tungsten target x-ray tube with total filtration values of 1.0, 2.0, and 3.0 mm aluminum. kVp and mAs are the same for the three spectra. (Computer simulation courtesy of Todd Steinberg, Colorado Springs.)



MARGIN FIGURE 5-9
Tungsten-target x-ray spectra generated at different tube voltages with a constant current through the x-ray tube. (From Ulrey, C., *Phys. Rev.* 1918; **1**:401.)

increases with increasing tube voltage because the efficiency of bremsstrahlung production increases with electron energy (see margin).

Tube Current and Time

The product of tube current in milliamperes and exposure time in seconds (mA · sec) describes the total number of electrons bombarding the target.

Example 5-1

Calculate the total number of electrons bombarding the target of an x-ray tube operated at 200 mA for 0.1 sec.

The ampere, the unit of electrical current, equals 1 coulomb/sec. The product of current and time equals the total charge in coulombs. X-ray tube current is measured in milliamperes, where 1 mA $= 10^{-3}$ amp. The charge of the electron is 1.6×10^{-19} coulombs, so

$$1 \text{ mA} \cdot \text{s} = \frac{(10^{-3}\text{coulomb/ sec})(\text{sec})}{1.6 \times 10^{-19}\text{coulomb/electron}} = 6.25 \times 10^{15}\text{electrons}$$

For 200 mA and 0.1 sec,

$$\begin{aligned}\text{No. of electrons} &= (200 \text{ mA})(0.1 \text{ sec})(6.25 \times 10^{15}\text{electrons/mA} \cdot \text{sec}) \\ &= 1.25 \times 10^{17}\text{electrons}\end{aligned}$$

Other factors being equal, more x rays are produced if more electrons bombard the target of an x-ray tube. Hence the number of x rays produced is directly proportional to the product (mA · sec) of tube current in milliamperes and exposure time in seconds. Spectra from the same x-ray tube operated at different values of mA · sec are shown in Figure 5-7. The overall shape of the spectrum (specifically the upper and lower limits of energy and the position of characteristic peaks) remains unchanged. However, the height of the spectrum and the area under it increase with increasing mA · sec. These increases reflect the greater number of x rays produced at higher values of mA · sec.

Target Material

The choice of target material in an x-ray tube affects the efficiency of x-ray production and the energy at which characteristic x rays appear. If technique factors (tube voltage, milliamperage, and time) are fixed, a target material with a higher atomic

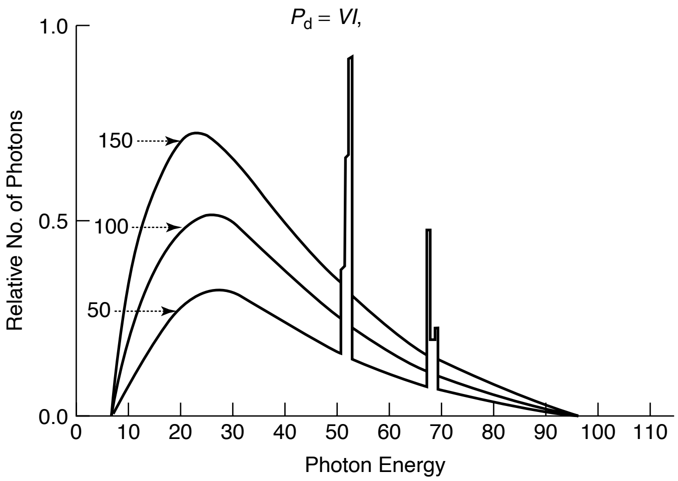


FIGURE 5-7
X-ray emission spectra for a 100-kVp tungsten target x-ray tube operated at 50, 100, and 150 mA. kVp and exposure time are the same for the three spectra.

number (Z) will produce more x rays per unit time by the process of bremsstrahlung.

The efficiency of x-ray production is the ratio of energy emerging as x radiation from the x-ray target divided by the energy deposited by electrons impinging on the target. The rate at which electrons deposit energy in a target is termed the *power deposition* P_d (in watts) and is given by

$$P_d = VI$$

In the equation $P_d = VI$, the voltage may also be expressed in kilovolts if the current is described in milliamperes.

where V is the tube voltage in volts and I is the tube current in amperes. The rate at which energy is released as x radiation,[3] termed the *radiated power* P_r, is

$$P_r = 0.9 \times 10^{-9} ZV^2 I \tag{5-1}$$

where P_r is the radiated power in watts (W) and Z is the atomic number of the target. Hence, the efficiency of x-ray production is

$$\text{Efficiency} = \frac{P_r}{P_d} = \frac{0.9 \times 10^{-9} ZV^2 I}{VI} \tag{5-2}$$
$$= 0.9 \times 10^{-9} ZV$$

Equation (5-2) shows that the efficiency of x-ray production increases with the atomic number of the target and the voltage across the x-ray tube.

Example 5-2

In 1 sec, 6.25×10^{17} electrons (100 mA) are accelerated through a constant potential difference of 100 kV. At what rate is energy deposited in the target?

$$P = (10^5 \text{V})(0.1\text{A})$$
$$= 10^4 \text{W}$$

X-ray production is a very inefficient process, even in targets with high atomic number. For x-ray tubes operated at conventional voltages, less than 1% of the energy deposited in the target appears as x radiation. Almost all of the energy delivered by impinging electrons is degraded to heat within the target.

Efficiency of converting electron energy into x rays as a function of tube voltage.[4]

kV	Heat (%)	X Rays (%)
60	99.5	0.5
200	99	1.0
4000	60	40

The characteristic radiation produced by a target is governed by the binding energies of the K, L, and M shells of the target atoms. Theoretically, any shell could contribute to characteristic radiation. In practice, however, transitions of electrons among shells beyond the M shell produce only low-energy x rays, ultraviolet light,

TABLE 5-2 Electron Shell Binding Energies (keV)

Shell	*Molybdenum* ($Z = 42$)	*Tungsten* ($Z = 74$)
K	20	69
L	2.9, 2.6, 2.5[a]	12, 11, 10
M	0.50, 0.41, 0.39, 0.23, 0.22	2.8, 2.6, 2.3, 1.9, 1.8

[a] Multiple binding energies exist within the L and M shells because of the range of discrete values of the quantum number *l*, discussed in Chapter 2.

A characteristic x ray released during transition of an electron between adjacent shells is known as an α x ray. For example, a K_α x ray is one produced during transition of an electron from the L to the K shell. A β x ray is an x ray produced by an electron transition among nonadjacent shells. For example, a K_β x ray reflects a transition of an electron from the M to the K shell.

and visible light. Low-energy x rays are removed by inherent filtration and do not become part of the useful beam. The characteristic peak for a particular shell occurs only when the tube voltage exceeds the binding energy of that shell. Binding energies in tungsten and molybdenum are shown in Table 5-2.

The characteristic radiation produced by an x-ray target is usually dominated by one or two peaks with specific energies slightly less than the binding energy of the K-shell electrons. The most likely transition involves an L-shell electron dropping to the K shell to fill a vacancy in that shell. This transition yields a photon of energy equal to the difference in electron binding energies of the K and L shells. A characteristic photon with an energy equal to the binding energy of the K shell alone is produced only when a free electron from outside the atom fills the vacancy. The probability of such an occurrence is vanishingly small.

During interaction of an electron with a target nucleus, a bremsstrahlung photon may emerge with energy equal to the total kinetic energy of the bombarding electron. Such a bremsstrahlung photon would have the maximum energy of all photons produced at a given tube voltage. The maximum energy of the photons depicted in Figure 5-5 therefore reflects the peak voltage applied across the x-ray tube, described in units of kilovolts peak (kVp). Photons of maximum energy E_{max} in an x-ray beam possess the maximum frequency and the minimum wavelength.

$$E_{max} = h\nu_{max} = \frac{hc}{\lambda_{min}}$$

The minimum wavelength in nanometers for an x-ray beam may be computed as

$$\lambda_{min} = \frac{hc}{E_{max}} = \frac{hc}{\text{kVp}}$$

where E_{max} is expressed in keV.

In the past, x-ray wavelengths were described in units of angstroms (Å), where $Å = 10^{-10}$ m. The minimum wavelength in angstroms would be λ_{min} (Å) $= 12.4/\text{kVp}$. The angstrom is no longer used as a measure of wavelength.

$$\lambda_{min} = \frac{(6.62 \times 10^{-34}\ \text{J-sec})(3 \times 10^{8}\ \text{m/sec})(10^{9}\ \text{nm/m})}{\text{kVp}(1.6 \times 10^{-16}\ \text{J/keV})}$$

$$\lambda_{min} = \frac{1.24}{\text{kVp}} \tag{5-3}$$

with λ_{min} expressed in units of nanometers (nm).

Example 5-3

Calculate the maximum energy and minimum wavelength for an x-ray beam generated at 100 kVp.

The maximum energy (keV) numerically equals the maximum tube voltage (kVp). Because the maximum tube voltage is 100 kVp, the maximum energy of the photons is 100 keV:

Compared with visible light, x rays have much shorter wavelengths. Wavelengths of visible light range from 400 nm (blue) to 700 nm (red).

$$\lambda_{min} = \frac{1.24}{100\ \text{kVp}}$$
$$= 0.0124\ \text{nm}$$

TUBE VACUUM

To prevent collisions between air molecules as electrons accelerate between the filament and target, x-ray tubes are evacuated to pressures less than 10^{-5} Hg. Removal of air also reduces deterioration of the hot filament by oxidation. During the manufacture of x-ray tubes, evacuation is accomplished by "outgassing" procedures that employ repeated heating cycles to remove gas occluded in components of the x-ray tube. Tubes still occasionally become "gassy," either after prolonged use or because the vacuum seal is not perfect. Filaments are destroyed rapidly in gassy tubes.

Many x-ray tubes include a "getter circuit" (active ion trap) to remove gas molecules that otherwise might accumulate in the x-ray tube over time.

ENVELOPE AND HOUSING

A vacuum-tight glass envelope (the "x-ray tube") surrounds other components required for the efficient production of x rays. The tube is mounted inside a metal housing that is grounded electrically. Oil surrounds the x-ray tube to (a) insulate the housing from the high voltage applied to the tube and (b) absorb heat radiated from the anode. Shockproof cables that deliver high voltage to the x-ray tube enter the housing through insulated openings. A bellows in the housing permits heated oil to expand when the tube is used. Often the bellows is connected to a switch that interrupts the operation of the x-ray tube if the oil reaches a temperature exceeding the heat storage capacity of the tube housing. A lead sheath inside the metal housing attenuates radiation emerging from the x-ray tube in undesired directions. A cross section of an x-ray tube and its housing is shown in Figure 5-8.

Secondary electrons can be ejected from a target bombarded by high-speed electrons. As these electrons strike the glass envelope or metallic components of the x-ray tube, they interact to produce x rays. These x rays are referred to as off-focus x rays because they are produced away from the target.

Off-focus radiation can also be reduced by using small auxiliary collimators placed near the output window of the x-ray tube.

The quality of an x-ray image is reduced by off-focus x rays. For example, off-focus radiation contributes as much as 25% of the total amount of radiation emerging from some x-ray tubes with rotating anodes.[5] The effects of off-focus radiation on images may be reduced by placing the beam collimators as close as possible to the x-ray target.

For x-ray images of highest quality, the volume of the target from which x rays emerge should be as small as possible. To reduce the "apparent size" of the focal spot,

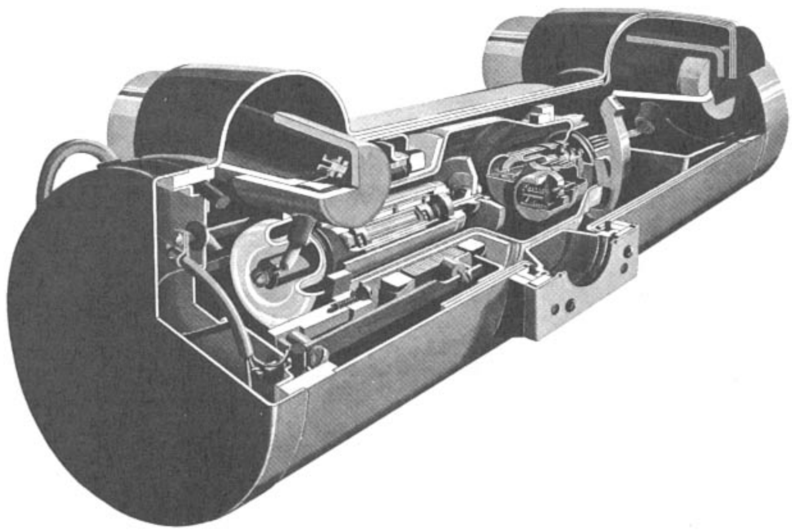
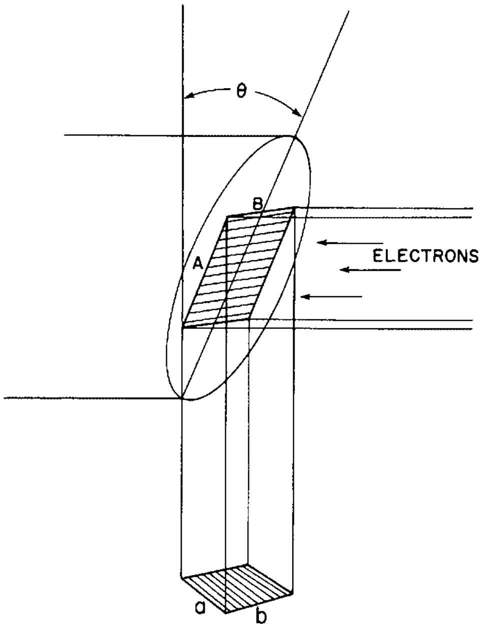
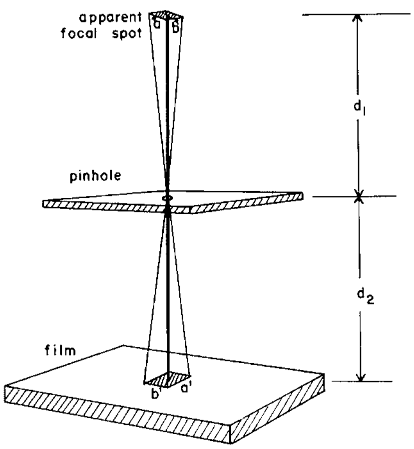


FIGURE 5-8
Cutaway of a rotating-anode x-ray tube positioned in its housing. (Courtesy of Machlett Laboratories, Inc.)

Electrons accelerated from the filament to the target are absorbed in the first 0.5-mm thickness of target. In this distance, a typical electron will experience 1000 or more interactions with target atoms.



MARGIN FIGURE 5-10
Illustration of the line-focus principle, which reduces the apparent size of the focal spot.



MARGIN FIGURE 5-11
Pinhole method for determining the size *ab* of the apparent focal spot.

the target of an x-ray tube is mounted at a steep angle with respect to the direction of impinging electrons (see Margin Figure 5-10). With the target at this angle, x rays appear to originate within a focal spot much smaller than the volume of the target absorbing energy from the impinging electrons. This apparent reduction in size of the focal spot is termed the *line-focus principle.* Most diagnostic x-ray tubes use a target angle between 6 and 17 degrees. In the illustration, side a of the projected or apparent focal spot may be calculated by

$$a = A \sin\theta \tag{5-4}$$

where A is the corresponding dimension of the true focal spot and θ is the target angle.

Example 5-4

By using the illustration in the margin and Eq. (5-4), calculate a if $A = 7$ mm and $\theta = 17$ degrees.

$$\begin{aligned} a &= \sin\theta \\ &= (7 \text{ mm})(\sin 17 \text{ degrees}) \\ &= (7 \text{ mm})(0.29) \\ &= 2 \text{ mm} \end{aligned}$$

Side b of the apparent focal spot equals side B of the true focal spot because side B is perpendicular to the electron beam. However, side B is shorter than side A of the true focal spot because the width of a filament is always less than its length. When viewed in the center of the field of view, the apparent focal spot usually is approximately square.

As mentioned earlier, dual-focus diagnostic x-ray tubes furnish two apparent focal spots, one for fine-focus radiography (e.g., 0.6 mm^2 or less) produced with a smaller filament, and another for course-focus radiography (e.g., 1.5 mm^2) produced with a larger filament. The apparent focal spot to be used is determined by the tube current desired. The small filament is used when a low tube current (e.g., 100 mA) is satisfactory. The coarse filament is used when a larger tube current (e.g., 200 mA or greater) is required to reduce exposure time. Apparent focal spots of very small dimensions (e.g., $\leq$0.1 mm) are available with certain x-ray tubes used for magnification radiography.

The apparent size of the focal spot of an x-ray tube may be measured with a pinhole x-ray camera.[5,6] A hole with a diameter of a few hundredths of a millimeter is drilled in a plate that is opaque to x rays. The plate is positioned between the x-ray tube and film. The size of the image of the hole is measured on the exposed film. From the dimensions of the image and the position of the pinhole, the size of the apparent focal spot may be computed. For example, the dimension (a) of the apparent focal spot shown in the margin may be computed from the corresponding dimension (a') in the image by

$$a = a'\left(\frac{d_1}{d_2}\right) \tag{5-5}$$

where d_1 is the distance from the target to the pinhole and d_2 is the distance from the pinhole in the film.

Focal spot size also can be measured with a resolution test object such as the star pattern shown in Figure 5-9. The x-ray image of the pattern on the right reveals a blur zone where the spokes of the test pattern are indistinct. From the diameter of the blur zone, the effective size of the focal spot can be computed in any dimension. This effective focal spot size may differ from pinhole camera measurements of the focal spot along the same dimension because the diameter of the blur zone is influenced not only by the actual focal spot dimensions, but also by the distribution of x-ray

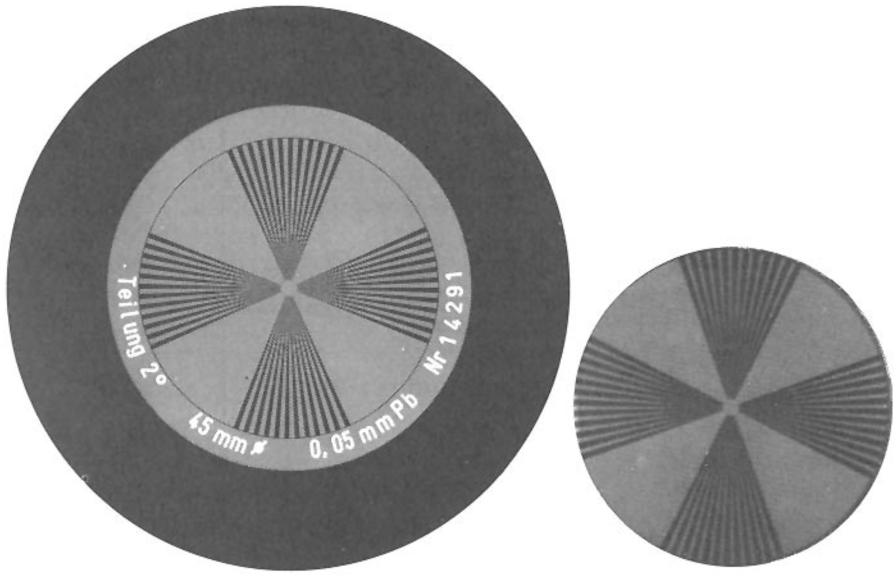
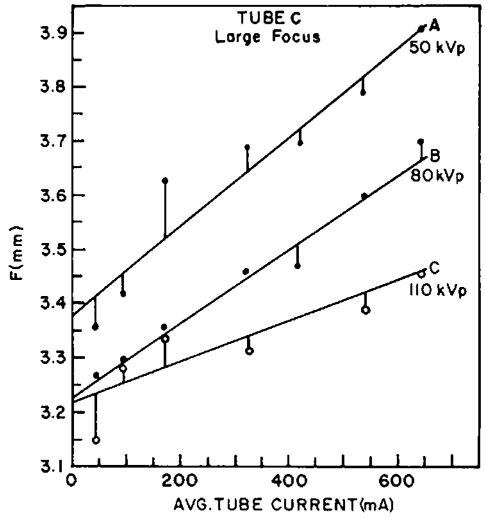
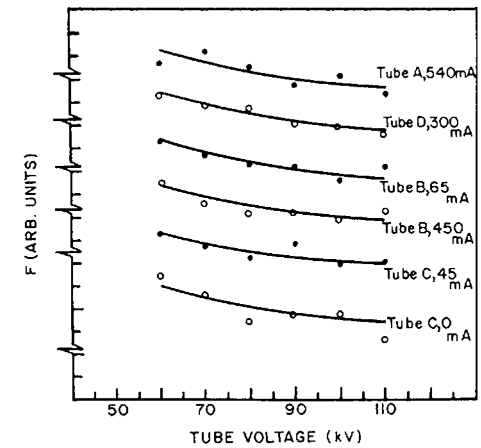


FIGURE 5-9
Contact radiograph (**left**) and x-ray image (**right**) of a star test pattern. The effective size of the focal spot may be computed from the diameter of the blur zone in the x-ray image.





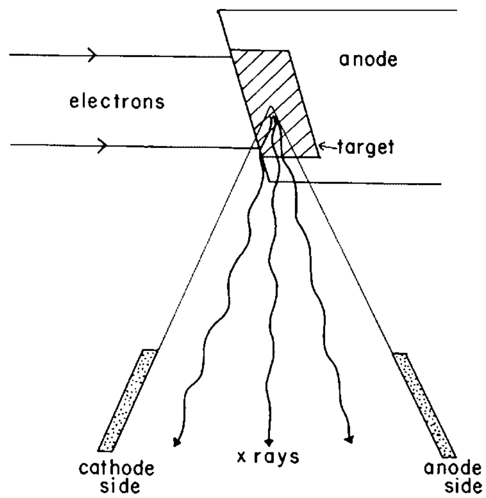
MARGIN FIGURE 5-12
Influence of tube current (**top**) and tube voltage (**bottom**) on the focal spot in a direction parallel to the motion of impinging electrons. (From Chaney, E., and Hendee, W. *Med Phys* 1974; **1**:131.)

intensity across the focal spot. In most diagnostic x-ray tubes, this distribution is not uniform. Instead, the intensity tends to be concentrated at the edges of the focal spot in a direction perpendicular to the electron beam.

For most x ray tubes, the size of the focal spot is not constant. Instead, it varies with both the tube current and the voltage applied to the x-ray tube.[7,8] This influence is shown in the margin for the dimension of the focal spot parallel to the motion of impinging electrons. On the top, the growth or "blooming" of the focal spot with tube current is illustrated. The gradual reduction of the same focal spot dimension with increasing levels of peak voltage is shown on the bottom.

Low-energy x rays generated in a tungsten target are attenuated severely during their escape from the target. For targets mounted at a small angle, the attenuation is greater for x rays emerging along the anode side of the x-ray beam than for those emerging along the side of the beam nearest the cathode. Consequently, the x-ray intensity decreases from the cathode to the anode side of the beam. This variation in intensity across an x-ray beam is termed the *heel effect*. The heel effect is noticeable for x-ray beams used in diagnostic radiology, particularly for x-ray beams generated at low kVp, because the x-ray energy is relatively low and the target angles are steep. To compensate for the heel effect, a filter may be installed in the tube housing near the exit portal of the x-ray beam. The thickness of such a filter increases from the anode to the cathode side of the x-ray beam. Positioning thicker portions of a patient near the cathode side of the x-ray beam also helps to compensate for the heel effect.

The heel effect increases with the steepness of the target angle. This increase limits the maximum useful field size obtainable with a particular target angle. For example, a target angle no steeper than 12 degrees is recommended for x-ray examinations using 14- ×17-in. film at a 40-in. distance from the x-ray tube, whereas targets as steep as 7 degrees may be used if field sizes no larger than 10 × 10 in. are required at the same distance.



MARGIN FIGURE 5-13
The heel effect is produced by increased attenuation of x rays in the sloping target near the anode side of the x-ray beam.

SPECIAL-PURPOSE X-RAY TUBES

Many x-ray tubes have been designed for special applications. A few of these special-purpose tubes are discussed in this chapter.

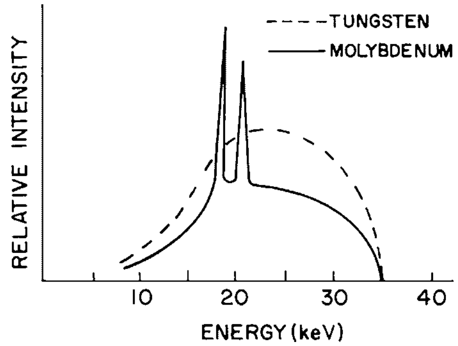


FIGURE 5-10
X-ray spectrum from molybdenum (solid line) and tungsten (dashed line) target x-ray tubes.

Grid-Controlled X-Ray Tubes

Grid-controlled x-ray tubes are also referred to as grid-pulsed or grid-biased x-ray tubes.

Grid-controlled x-ray tubes cost significantly more than non biased tubes.

In a grid-controlled x-ray tube, the focusing cup in the cathode assembly is maintained a few hundred volts negative with respect to the filament. In this condition, the negative potential of the focusing cap prevents the flow of electrons from filament to target. Only when the negative potential is removed can electrons flow across the x-ray tube. That is, applying and then momentarily removing the potential difference between the focusing cup and the filament provides an off-on switch for the production of x rays. Grid-controlled x-ray tubes are used for very short ("pulsed") exposures such as those required during digital radiography and angiography.

Field-Emission X-Ray Tubes

In the field-emission x-ray tube, the cathode is a metal needle with a tip about 1 μm in diameter. Electrons are extracted from the cathode by an intense electrical field rather than by thermionic emission. At diagnostic tube voltages, the rate of electron extraction is too low to provide tube currents adequate for most examinations, and efforts to market field-emission x-ray tubes in clinical radiology have been limited primarily to two applications: (1) pediatric radiography where lower tube currents can be tolerated and (2) high-voltage chest radiography where higher tube voltages can be used (as much as 300 kVp) to enhance the extraction of electrons from the cathode. Neither application has received much acceptance in clinical radiology.

Molybdenum-Target X-Ray Tubes

For low-voltage studies of soft-tissue structures (e.g., mammography), x-ray tubes with molybdenum and rhodium targets are preferred over tubes with tungsten targets. In the voltage range of 25 to 45 kVp, K-characteristic x rays can be produced in molybdenum but not in tungsten. These characteristic molybdenum photons yield a concentration of x rays on the low-energy side of the x-ray spectrum (Figure 5-10), which enhances the visualization of soft-tissue structures. Properties of molybdenum-target x-ray tubes are discussed in greater detail in Chapter 7.

RATINGS FOR X-RAY TUBES

The high rate of energy deposition in the small volume of an x-ray target heats the target to a very high temperature. Hence a target should have high thermal conductivity to transfer heat rapidly to its surroundings. Because of the high energy deposition in the target, rotating anodes are used in almost all diagnostic x-ray tubes. A rotating anode increases the volume of target material that absorbs energy from impinging electrons, thereby reducing the temperature attained by any portion of the anode. The anode is attached to the rotor of a small induction motor by a stem that usually is molybdenum. Anodes are 3 to 5 inches in diameter and rotate at speeds up to 10,000 rpm. The induction motor is energized for about 1 second before high voltage is applied to the x-ray tube. This delay ensures that electrons do not strike the target before the anode reaches its maximum speed of rotation. Energy deposited in the rotating anode is radiated to the oil bath surrounding the glass envelope of the x-ray tube.

Larger diameters and higher rotational speeds are required for applications such as angiography and helical-scan computed tomography that deliver greater amounts of energy to the x-ray target.

The rotating anode, together within the stator and rotor of the induction motor, are known collectively as the *anode assembly*.

Maximum Tube Voltage, Filament Current, and Filament Voltage

The maximum voltage to be applied between filament and target is specified for every x-ray tube. This "voltage rating" depends on the characteristics of the applied voltage (e.g., single phase, three phase, or constant potential) and on the properties of the x-ray tube (e.g., distance between filament and target, shape of the cathode assembly and target, and shape of the glass envelope). Occasional transient surges in voltage may be tolerated by an x-ray tube, provided that they exceed the voltage rating by no more than a few percent.

Limits are placed on the current and voltage delivered to coarse and fine filaments of an x-ray tube. The current rating for the filament is significantly lower for continuous compared with pulsed operation of the x-ray tube, because the temperature of the filament rises steadily as current flows through it.

Maximum Energy

Maximum-energy ratings are provided for the target, anode, and housing of an x-ray tube.[10,11] These ratings are expressed in heat units, where for single-phase electrical power

$$\begin{aligned}\text{Number of heat units (HU)} &= \text{(Tube voltage) (Tube current) (Time)}\\ &= \text{(kVp) (mA) (sec)} \qquad \textbf{(5-6)}\end{aligned}$$

If the tube voltage and current are constant, then 1 HU = 1J of energy. For three-phase power, the number of heat units is computed as

$$\begin{aligned}\text{Number of heat units (HU)} &= \text{(Tube voltage) (Tube current) (Time) (1.35)}\\ &= \text{(kVp) (mA) (sec) (1.35)} \qquad \textbf{(5-7)}\end{aligned}$$

For x-ray tubes supplied with single-phase (1φ), full-wave rectified voltage, the peak current through the x-ray tube is about 1.4 times the average current. The average current nearly equals the peak current in x-ray generators supplied with three-phase (3φ) voltage. For this reason, the number of heat units for an exposure from a 3φ generator is computed with the factor 1.35 in Eq. (5-7). For long exposures or a series of exposures with an x-ray tube supplied with 3φ voltage, more energy is delivered to the target, and the number of exposures in a given interval of time must be reduced. Separate rating charts are usually provided for 1φ and 3φ operation of an x-ray tube.

X-ray tubes are also rated in terms of their maximum power loading in kilowatts. Representative maximum power loadings are shown below[9]:

Focal Spot (mm)	Power Rating (kW)
1.2–1.5	80–125
0.8–1.0	50–80
0.5–0.8	40–60
0.3–0.5	10–30
$\leq$ 0.3	1–10
$\leq$ 0.1	<1

Energy ratings for the anode and the tube housing are expressed in terms of heat storage capacities. The heat storage capacity of a tube component is the total number of heat units that may be absorbed without damage to the component. Anode heat storage capacities for diagnostic x-ray tubes range from several hundred thousand to over a million heat units.

The heat storage capacity of the x-ray tube housing is also important because heat is transferred from the anode to the tube housing. The housing heat storage capacity exceeds the anode capacity and is usually on the order of 1.5 million HU.

To determine whether the target of an x-ray tube might be damaged by a particular combination of tube voltage, tube current, and exposure time, *energy rating charts* furnished with the x-ray tube should be consulted. To use the sample chart shown in Figure 5-11, a horizontal line is drawn from the desired tube current on the *y* axis to the curve for the desired tube voltage. From the intersection of the line and the curve, a vertical line dropped to the *x* axis reveals the maximum exposure time

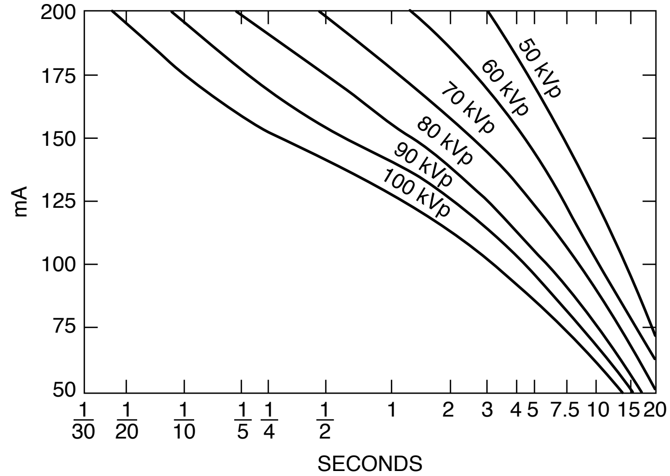


FIGURE 5-11
Energy rating chart for a Machlett Dynamax "25" x-ray tube with a 1-mm focal spot and single-phase, fully rectified voltage.(Courtesy of Machlett Laboratories, Inc.)

that can be used for a single exposure without possible damage to the x-ray target. The area under each voltage curve encompasses combinations of tube current and exposure time that do not exceed the target-loading capacity when the x-ray tube is operated at that voltage. The area above each curve reflects combinations of tube current and exposure time that overload the x-ray tube and might damage the target. Often switches are incorporated into an x-ray circuit to prevent the operator from exceeding the energy rating for the x-ray tube. Shown in Figure 5-12 are a few targets damaged by excess load or improper rotation of the target.

An *anode thermal-characteristics chart* describes the rate at which energy may be delivered to an anode without exceeding its capacity for storing heat (Figure 5-13). Also shown in the chart is the rate at which heat is radiated from the anode to the insulating oil and housing. For example, the delivery of 425 HU per second to the anode of the tube exceeds the anode heat storage capacity after 5.5 minutes. The delivery of 340 HU per second could be continued indefinitely. The cooling curve in Figure 5-13 shows the rate at which the anode cools after storing a certain amount of heat.

The housing-cooling chart in Figure 5-14 depicts the rate at which the tube housing cools after storing a certain amount of heat. The rate of cooling with and without forced circulation of air is shown. Charts similar to those in Figures 5-13 and 5-14 are used to ensure that multiple exposures in rapid succession do not damage an x-ray tube or its housing.

When several exposures are made in rapid succession, a target-heating problem is created that is not directly addressed in any of the charts described above. This target-heating problem is caused by heat deposition that exceeds the rate of heat dissipation in the focal track of the rotating anode. To prevent this buildup of heat from damaging the target, an additional tube-rating chart should be consulted. This chart, termed an angiographic rating chart because the problem of rapid successive exposures occurs frequently in angiography, is illustrated in Figure 5-15. Use of this chart is depicted in Example 5-9.

Example 5-5

From the energy-rating chart in Figure 5-11, is a radiographic technique of 150 mA, 1 second at 100 kVp permissible?

The maximum exposure time is slightly longer than 0.25 seconds for 150 mA at 100 kVp. Therefore, the proposed technique is unacceptable.

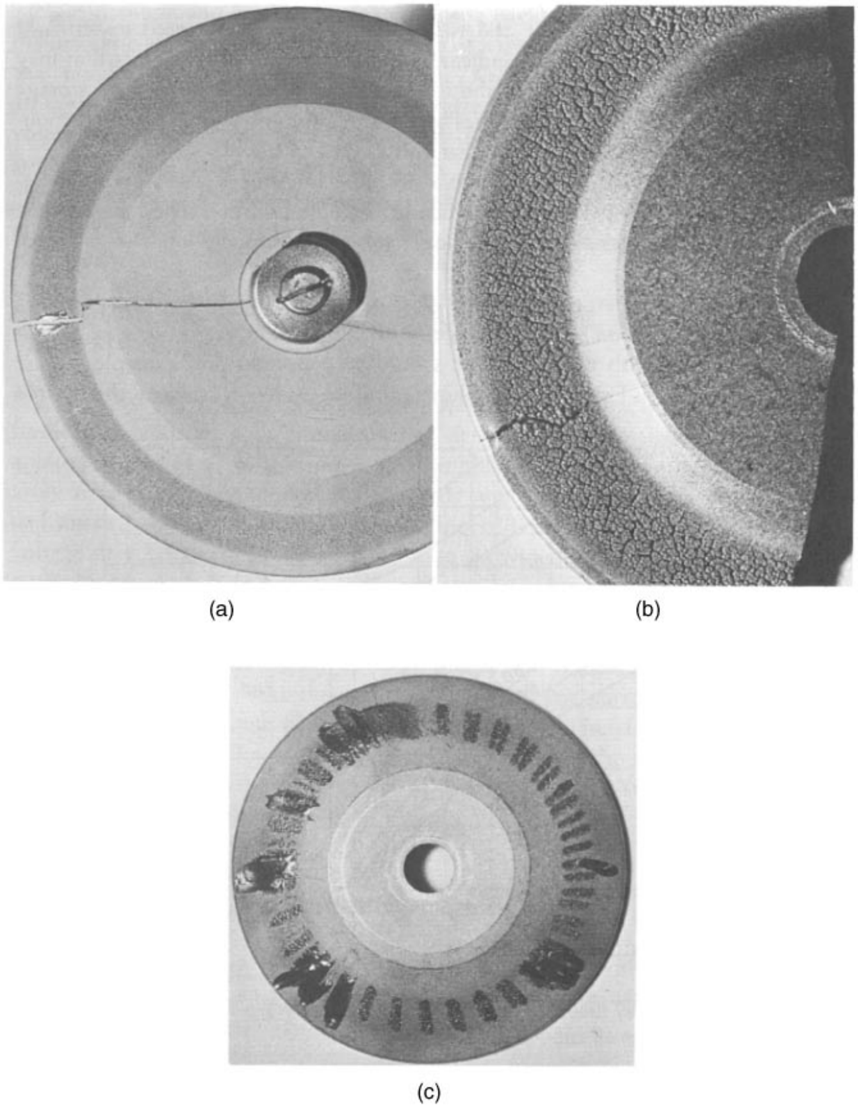


FIGURE 5-12
Rotating targets damaged by excessive loading or improper rotation of the target. **A:** Target cracked by lack of rotation. **B:** Target damaged by slow rotation and excessive loading. **C:** Target damaged by slow rotation.

Is 100 mA at 100 kVp for 1.5 seconds permissible?

The maximum exposure time is 3 seconds for 100 mA at 100 kVp. Therefore the proposed technique is acceptable.

Example 5-6

Five minutes of fluoroscopy at 4 mA and 100 kVp are to be combined with eight 0.5-second spot films at 100 kVp and 100 mA. Is the technique permissible according to Figures 5-11 and 5-13?

The technique is acceptable according to the energy-rating chart in Figure 5-11. By rearranging Eq. (5-6) we calculate that the rate of delivery of energy to the anode during fluoroscopy is (100 kVp) (4 mA) = 400 HU per second. From Figure 5-13, after 5 minutes approximately 60,000 HU have been accumulated by the anode. The eight spot films contribute an additional 40,000 HU [(100 kVp) (100 mA) (0.5 sec) 8 = 40,000 HU]. After all exposures have been made, the total heat stored in the

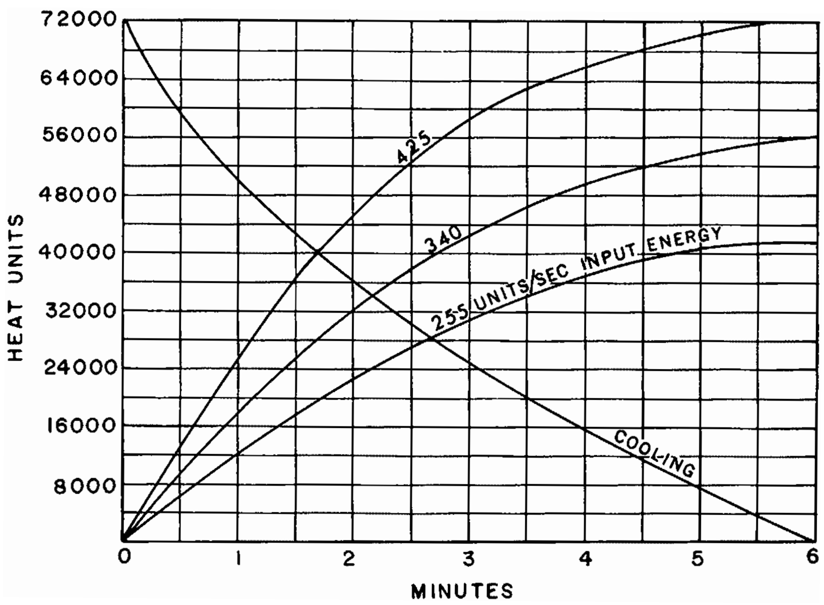


FIGURE 5-13
Anode thermal characteristics chart for a Machlett Dynamax "25" rotating anode x-ray tube. The anode heat-storage capacity is 72,000 HU. (Courtesy of Machlett Laboratories, Inc.)

anode is 40,000 + 60,000 = 100,000 HU. This amount of heat exceeds the anode heat storage capacity of 72,000 HU. Consequently the proposed technique is unacceptable.

Example 5-7

Three minutes of fluoroscopy at 3 mA and 85 kVp are combined with four 0.25-second spot films at 85 kVp and 150 mA. From Figure 5-13, what time must elapse before the procedure may be repeated?

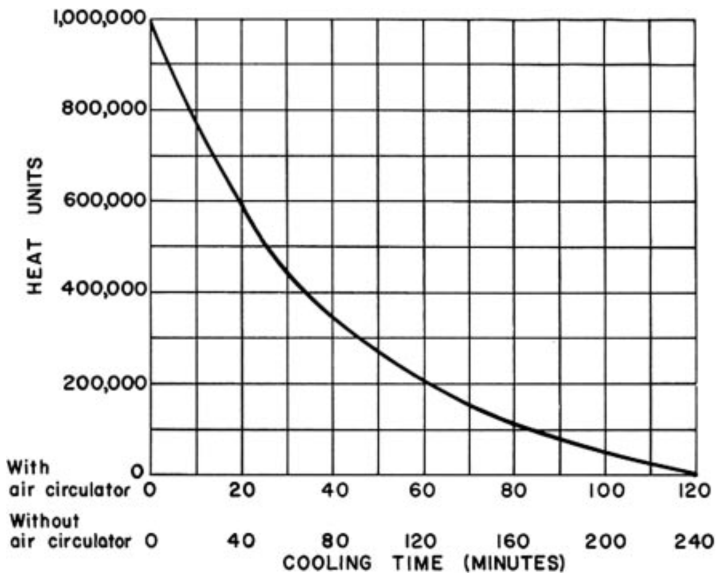


FIGURE 5-14
Housing-cooling chart for a Machlett Dynamax "25" x-ray tube. (Courtesy of Machlett Laboratories, Inc.)

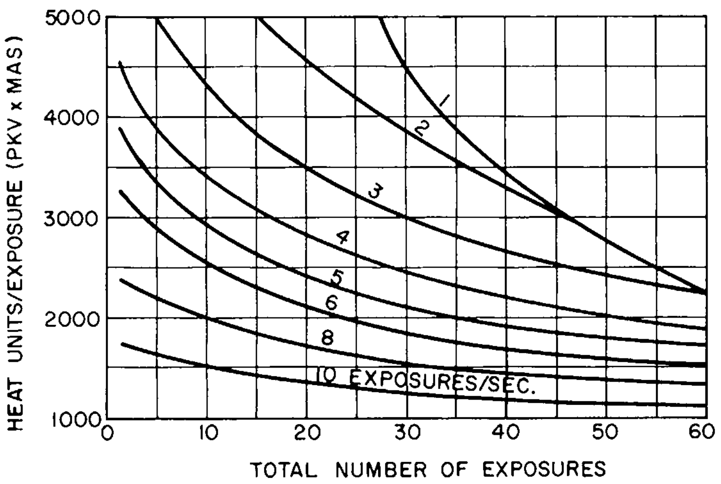


FIGURE 5-15
Angiographic rating chart for a Machlett Super Dynamax x-ray tube, 1.0-mm focal spot, full-wave rectification, single phase.

The rate of delivery of energy to the anode is (85 kVp) (3 mA) = 255 HU per second, resulting in a heat load of 31,000 HU after 3 minutes. To this heat load is added (85 kVp) (150 mA) (0.25 sec) (4) = 12,750 HU for the four spot films to yield a total heat load of 43,750 HU. From the position of the vertical axis corresponding to this heat load, a horizontal line is extended to intersect the anode-cooling curve at 1.4 minutes. From this intersection, the anode must cool until its residual heat load is 72,000 – 43,750 = 28,250 HU, so that when the 43,750 HU from the next procedure is added to the residual heat load, the total heat load does not exceed the 72,000 HU anode heat storage capacity. The time corresponding to a residual heat load of 28,250 HU is 2.6 minutes. Hence the cooling time required between procedures is 2.6 – 1.4 = 1.2 minutes.

Example 5-8

From Figures 5-11 and 5-13, it is apparent that three exposures per minute are acceptable if each 1φ exposure is taken at 0.5 second, 125 mA, and 100 kVp. Could this procedure be repeated each minute for 1 hour? The rate of energy transfer to the housing is

$$= (100 \text{ kVp})\ (125 \text{ mA})\ (0.5\,\text{sec})\ (3\,\text{exposures/min})$$

$$= 18,750 \text{ HU/min}$$

At the end of 1 hour, (18,750 HU/min) (60 min/hr) = 1,125,000 HU will have been delivered to the housing. The heat storage capacity of the housing is only 1 million HU. Without forced circulation of air, the maximum rate of energy dissipation from the housing is estimated to be approximately 12,500 HU per minute (Figure 5-14). With air circulation, the rate of energy dissipation is 25,000 HU per minute. Therefore the procedure is unacceptable if the housing is not air-cooled, but is acceptable if the housing is cooled by forced circulation of air.

Example 5-9

From Figure 5-15, how many consecutive exposures can be made at a rate of six exposures per second if each exposure is taken at 85 kVp, 500 mA, and 0.05 seconds?

Each exposure produced (85 kVp) (500 mA) ().05 sec) = 2125 HU. A horizontal line from this position on the y axis intersects the 6 exposures per second curve at a position corresponding to 20 exposures. Hence, no more than 20 exposures should be made at a rate of 6 exposures per second.