The AAPM/RSNA Physics Tutorial for Residents

X-ray Tubes

Frank E. Zink, PhD

The x-ray tube serves the function of creating x-ray photons from electric energy supplied by the x-ray generator. The process of creating the x-ray beam is very inefficient, with only 1% of the electric energy converted to x-ray photons and the remaining 99% converted to heat in the x-ray tube assembly. Thus, to produce sufficient x-ray output for diagnostic imaging, the x-ray tube must withstand and dissipate a substantial heat load, a requirement that affects the design and composition of the x-ray tube. The major x-ray tube components are the cathode and anode assemblies, the tube envelope, the rotor and stator (for rotating anode systems), and the tube housing. The design of the x-ray tube determines the basic characteristics of the x-ray beam such as focal spot size, x-ray field uniformity, and the x-ray energy spectrum. These x-ray beam characteristics are important because they affect radiologic parameters such as spatial resolution, image contrast, and patient dose.

INTRODUCTION

A basic understanding of the x-ray tube is important because x-ray beam characteristics substantially affect spatial resolution, image contrast, and patient dose. The objectives of this article are to review the process of x-ray generation for diagnostic imaging, identify the basic components of the x-ray tube, and describe how x-ray tube design determines the characteristics of the x-ray beam. In addition, basic x-ray beam characteristics as well as some common variations in x-ray tube design are described.

GENERATION OF X RAYS

X rays are generated when accelerating electrons interact with matter. In a radiographic x-ray tube, energetic electrons interact with a target material and a portion of the kinetic energy of the electrons is converted to electromagnetic radiation, or x rays (most of the kinetic energy of the electrons is dissipated as heat).

The simplified x-ray tube electric circuit shown in Figure 1 illustrates the basic process of generating x rays with a radiographic tube. An x-ray generator provides a large potential difference (20–150 kV) between the cathode and anode components of the x-ray tube. A separate low-voltage circuit (also provided by the generator) produces a current through a wire filament at the cathode of the tube. The current in the wire...
filament causes it to heat up and emit electrons, a process called thermionic emission. The emitted electrons are accelerated by the large potential difference from the cathode to the anode. The flow of electrons in the wire filament within the cathode is called the filament current. The flow of electrons between the cathode and anode components of the tube is called the tube current.

At the anode target within the anode, the energetic electrons are converted to x rays in two ways: the bremsstrahlung process and characteristic x-ray production. X rays exit the tube in all directions but are limited to a desired beam size by lead housing and collimators and finally interact with the patient and detector to form the useful image.

- **X-ray Generator**
  The function of the x-ray generator is to provide the power necessary for producing x rays. Ideally, a constant high voltage is provided to the x-ray tube during the x-ray exposure to produce the tube current. In practice, some variation in voltage occurs during an x-ray exposure, a characteristic referred to as "voltage ripple." The voltage as a function of time is called the voltage waveform.

  The high voltage (20-150 kV) provided by the generator is selected by the user, and it determines the maximum energy of the resulting x rays. This peak voltage value is called the kilovolt peak (kVP). The generator also provides a fixed low voltage (approximately 10 V) to produce the filament current. The user selects the filament current, which determines the number of electrons emitted by the filament and thus the number of electrons available to be accelerated to the anode, which effectively determines the tube current (1-1,000 mA). The tube current is very sensitive to changes in the filament current, with a 1% change in filament current resulting in more than a 10% change in tube current. The user also may select the exposure time, that is, the time during which the high voltage is applied to the tube to produce x rays. Exposure times may vary from 25 msec to 2 seconds for general radiographic applications.

- **Bremsstrahlung Process**
  The process of x-ray generation in an x-ray tube is very inefficient, with a large majority (99%) of the kinetic energy from the accelerated electrons being converted to heat, while the remaining portion (1%) is converted to x rays. The substantial heat produced in an x-ray tube limits its design and performance.

  Of the x rays that are produced, most are bremsstrahlung ("braking") radiation. Bremsstrahlung x rays are produced when an energetic electron passes close to an atomic nucleus of the target material. The positive charge of the nucleus decelerates the negatively charged electron, causing the electron to change its path and give up energy in the form of an x ray.

  The amount of energy given up by the electron and thus the energy of the resulting x-ray is determined by how close the electron path is to the nucleus. As electrons are accelerated toward the target material, electrons pass by nuclei at a variety of distances, and thus a spectrum of x-ray energies is produced. Because the space between target nuclei is relatively large compared with the diameter of the nuclei, low-energy x rays are more likely to be produced than are high-energy x rays, which are produced only when electrons travel close to the target nucleus.
to the nuclei. The maximum possible x-ray energy is produced when an electron collides directly with a nucleus and gives up all its kinetic energy in the form of an x-ray. Figure 2 shows a bremsstrahlung energy spectrum. For an unfiltered spectrum, the energy of the bremsstrahlung x-rays produced ranges from zero to a maximum value determined by the kilovolt peak setting of the generator.

To maximize bremsstrahlung x-ray output, it is desirable to use a target material with a high atomic number and therefore a nucleus with a larger positive charge; this characteristic makes electrostatic deflection of passing accelerated electrons more likely. Tungsten is a common target material, chosen both for its high atomic number \((Z = 74)\) and its high melting point.

**Characteristic X-ray Production**

If the energy of the electrons accelerated toward the target is high enough, a characteristic x-ray may be produced as a result of the accelerated electron interacting with an inner shell orbital electron in the target material. Orbital electrons of the target material are bound to the nucleus at discrete binding energies. If the energy of the accelerated electron exceeds the discrete binding energy, the inner shell electron may be ejected, leaving an unfilled inner shell. An outer shell electron will subsequently fill the vacancy, emitting an x-ray of energy equal to the difference in binding energy between the outer and inner shells. The emitted x-rays have discrete energy values “characteristic” of the atomic number of the target material (which determines the discrete orbital energy levels and possible energy transitions).

Although many energy transitions are possible, most characteristic x-rays in diagnostic radiography arise from vacancies within the innermost K shell that are filled by electrons from the adjacent L, M, and N shells. Figure 2 shows the x-ray intensity contributed at discrete energy levels to the emission spectrum of an x-ray tube. For a general radiographic x-ray tube, characteristic radiation may contribute as much as 20% of the overall x-ray intensity, with the remaining 80% contributed by the bremsstrahlung process. Table 1 shows characteristic x-ray energies for two common target materials, tungsten (used in general radiography) and molybdenum (used in mammography). For characteristic x rays to be produced, the energy of the accelerated electrons must exceed the K-shell binding energy of the target material (69.4 keV for tungsten, 20 keV for molybdenum).

**Table 1**

<table>
<thead>
<tr>
<th>Target Material</th>
<th>Orbital Transition</th>
<th>X-ray Energy (keV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tungsten</td>
<td>L to K</td>
<td>59.3</td>
</tr>
<tr>
<td>Tungsten</td>
<td>L to K</td>
<td>60.0</td>
</tr>
<tr>
<td>Tungsten</td>
<td>M to K</td>
<td>67.2</td>
</tr>
<tr>
<td>Molybdenum</td>
<td>L to K</td>
<td>17.4</td>
</tr>
<tr>
<td>Molybdenum</td>
<td>L to K</td>
<td>17.5</td>
</tr>
<tr>
<td>Molybdenum</td>
<td>M to K</td>
<td>19.6</td>
</tr>
</tbody>
</table>

**Figure 2.** Emission spectrum for a tungsten target. The dotted line represents the unfiltered, bremsstrahlung portion of the x-ray spectrum emitted from the target. The solid line represents the total spectrum of x rays after they have exited the x-ray tube and housing. The peaks in the solid line (vertical solid lines) are characteristic x rays emitted by the x-ray tube. The total emission spectrum includes both bremsstrahlung and characteristic radiation.


**Components of X-ray Tubes**

The basic components of the x-ray tube include the cathode assembly, anode assembly, tube envelope, rotor and stator, and the tube housing (Fig 3). The tube envelope and the components within are referred to as the tube insert. Typically, when an x-ray tube fails, only the insert needs to be replaced. The tube envelope is evacuated (i.e., it contains no gas molecules), and the space between the envelope and housing is filled with oil to aid in tube cooling and to provide electric insulation.

- **Cathode Assembly**

  The cathode assembly is the source of electrons in the x-ray tube (Fig 4). A low-voltage circuit from the x-ray generator provides current through the tube filament. The filament is made of tungsten and consists of a helical wire coil 10–20 mm in length and 2–5 mm in width within a focusing cup. The filament current "boils" electrons off the tungsten wire (thermionic emission). The high voltage between the cathode and anode creates an electric field. The shape of the electric field and the shape of the filament form an electrostatic lens, which determines the trajectories of the electrons and their impact area on the target, an area called the focal spot.

  To resolve fine detail in a radiograph, it is desirable to have as small a focal spot as possible. The size of the focal spot is directly related to the size of the filament. As the size of the focal spot is made smaller, the local heat loading on the target becomes relatively greater. Thus, the maximum allowed power setting (tube current kilovolt peak) must be lowered. As a result, the maximum x-ray intensity that the tube can provide is also lowered.

  A practical design minimum for the filament size is determined by the need to produce sufficient x-ray intensity in a short exposure time. If exposure times are too long, motion blurring may occur in the image. Typically, a cathode assembly has two filaments: (a) a larger filament to accommodate imaging applications that require high x-ray tube output in a short exposure time and (b) a smaller filament for use in applications in which resolving capability is crucial or in which motion blurring is not a concern.

- **Anode Assembly**

  Two anode types are commonly used: fixed and rotating. Originally, a fixed anode was used, with tungsten target material imbedded...
in a fixed copper block. The low heat capacity of the fixed anode tube design limited its x-ray intensity (ie, x-ray output). A substantial increase in x-ray intensity became possible with the development of a rotating anode (target) design. With a rotating anode, the instantaneous heat load produced at the focal point of the accelerated electrons can be spread over a much larger area.

The target material used in an x-ray anode, whether fixed or rotating, must have a high atomic number to maximize bremsstrahlung output. The target material must also tolerate a tremendous heat load. Tungsten is chosen as the target material for general radiography because of its high atomic number, high melting point, and low vapor pressure. The tungsten is typically alloyed with 10% rhenium to improve its resistance to thermal damage. Tungsten is not a good conductor and is usually used as a thin overlay on a molybdenum substrate (for a rotating anode) or as a thin inlay within a copper substrate (for a fixed anode).

Figure 5 illustrates a fixed anode x-ray tube. The copper block in which the target is imbedded is designed to conduct heat away from the anode to the oil bath surrounding the tube envelope. The heat load on this type of tube limits its use to low-output applications, such as dental radiography and portable fluoroscopy.

The limited x-ray output of the fixed anode design led to the development of the rotating anode (Fig 6). By rotating the target rapidly during the exposure time, the effective surface area of the target is substantially increased (by more than 100 times), allowing the instantaneous heat load on the target to be increased. In turn, higher heat load allows higher x-ray output.

A particular challenge for the designers of the rotating anode was how to achieve anode rotation while maintaining the vacuum within the tube envelope. The problem was solved by using an electric induction motor, the elements of which are shown in Figure 6. An alternating current in the stator windings (which are outside the tube envelope) induces a changing magnetic field within the envelope. The rotor (which is within the envelope) rotates in response to the changing magnetic field, causing the attached target disk to rotate.
In both fixed and rotating anode tube designs, the target is canted from the perpendicular anode-cathode axis by a small angle (Fig 7). The x rays emanate from the target over a wide angle, which is limited on the target side by re-absorption in the target itself and which is limited on the cathode side by the collimation of the exit port on the x-ray tube. The effective focal spot size is the projection of the electron path on the target as viewed from the detector. A smaller effective focal spot size is desirable for better spatial resolution. Tube designs vary in target angle, with a tradeoff occurring between effective focal spot size and maximum field size. A larger target angle allows a larger x-ray field size but also produces a larger effective focal spot size.

If a tube is designed with a small anode angle, the electrons can strike a relatively large actual area on the target, whereas the projection of the focal spot as viewed from the detector appears relatively small. This "projection" effect is a result of the line focus principal and is illustrated in Figure 7. The effective focal spot size is the product of the actual size and the sine of the anode angle. This principal applies only in the anode-cathode direction. In the direction perpendicular to the anode-cathode axis, no projection effect occurs. The effective size is the same as the actual size.

Ideally, the distribution of x-ray intensity would be equal across the x-ray field. However, x-ray intensity varies substantially because of the heel effect, which is defined as the reduction in x-ray beam intensity along the anode-cathode axis caused by self-absorption by the anode target material (Fig 8). Self-absorption occurs because most of the x rays are produced at a finite depth within the target. X rays exiting the target toward the anode side of the x-ray field must traverse a longer path of target material, which makes self-absorption by the target more likely. Thus, the beam intensity is lower on the anode side of the field. However, there is no heel effect in the direction perpendicular to the anode-cathode axis.

Because the cathode side of the x-ray field has a higher x-ray intensity due to the heel effect, the x-ray tube is sometimes positioned such that the cathode side of the field irradiates the most attenuating portion of the patient anatomy. For example, in mammography, the x-ray tube is positioned such that the cathode side of the tube corresponds to the chest wall side of the patient.

- X-ray Tube Envelope
The envelope is an important component of the x-ray tube because it maintains the required evacuated environment. If the internal tube elements emit gas (outgas) or the vacuum fails, the tube becomes gassy and the gas molecules will impede electron flow between the cathode and anode. In this situation, the filament on the cathode is also likely to oxidize, causing the tube to fail.

The tube envelope has an exit port or window, a region through which the x rays exit the tube. The exit window is typically a thinner region of glass, but it may also consist of an entirely different material. Mammographic x-ray tubes are typically made of glass, with exit windows composed of beryllium. Beryllium has a lower atomic number than glass and absorbs less of the low-energy x rays used in mammography.

- Rotor and Stator
The electric induction motor that turns the rotating anode consists of a rotor, which rotates on a set of bearings within the glass x-ray tube envelope, and the stator, which consists of wire windings external to the envelope (Fig 6).
An alternating current is applied to the stator windings, inducing a changing magnetic field within the region of the rotor. The metal rotor turns in response to the changing magnetic field, causing the attached anode to turn. Anode rotation speeds range from 3,000 rpm (revolutions per minute) to 10,000 rpm for a high-speed anode.

The rotor is connected to the anode disk by the anode stem, which is typically made of molybdenum or stainless steel. The stem is designed to protect the bearings from heat damage, a common cause of x-ray tube failure.

**X-ray Tube Housing**

The tube housing (Fig 9) is the external structure of the x-ray tube. The housing provides structural support, supplies electric insulation, and shields the patient and personnel from radiation outside the intended primary beam defined by the exit port of the housing. Stray radiation that exits the tube housing from areas other than the exit port is called leakage radiation. The maximum acceptable amount of leakage radiation is limited by regulation to protect both patients and personnel. Within the housing, an oil bath surrounds the tube envelope to carry away heat radiated by the rotating anode (infrared radiation). The tube may contain an expansion bellows to allow expansion and contraction of the oil volume as the tube heats and cools.

The exit port of the tube housing serves to restrict the primary x-ray beam exiting from the tube. As the beam passes through the exit port, it has already been "hardened" by inherent filtration. Hardening is the preferential attenuation of lower-energy x rays, resulting in a beam of higher effective energy. The beam is further hardened by added filtration, typically consisting of aluminum. Hardening of the beam by both types of filtration is desirable because it eliminates most of the very low-energy x rays (<10 keV) that otherwise would contribute only to patient dose.

External to the x-ray tube housing is mounted a set of adjustable lead collimators that allow the operator to collimate the x-ray beam.
beam to the region of interest. The external collimator also houses the optical light source that provides (via a mirror) a light field congruent with the x-ray field to aid the technologist in positioning the patient.

- **X-ray Tube Heat Capacity**

Because the heat capacity of different x-ray tubes varies and because the heat load on a given tube must be monitored during use, we quantify heat loading by using the heat unit. The heat unit (HU) is the product of the kilovolt peak, tube current, exposure time, and a constant multiplier that depends on the voltage waveform.

A typical anode has a heat capacity of 250,000 HU. A heavy-duty anode may have a heat capacity of more than 1,000,000 HU and is much thicker and heavier than a typical anode. Because the anode transfers most of its heat energy by radiation to the tube housing, the size of the housing, the amount of oil, and its material design determine how much heat the x-ray tube can dissipate without material failure. Fans or oil circulators can be used to cool the tube housing more rapidly in imaging applications that have a high heat load. Heat loading varies considerably depending on the application; for example, a single chest radiograph produces only 5,000 HU and a series of cine exposures produces as much as 200,000 HU.

X-ray tube rating charts are used to express the operating limits of tubes due to heat loading and are provided by the x-ray manufacturer. Separate charts are available that express instantaneous heat capacity, continuous-load heat capacity (fluoroscopy), as well as cooling rates for anodes and tube housing. These operating limits depend on various tube characteristics, including filament and focal spot size, anode rotation speed, and a cooling mechanism for the oil bath. In modern x-ray systems, heat loading is monitored by a system computer that can warn the operator if x-ray exposure should be delayed to prevent damage to the x-ray tube.

- **X-RAY BEAM CHARACTERISTICS**

In addition to recognizing the x-ray tube components, it is important to understand how they affect x-ray beam characteristics, such as the energy spectrum and the effective focal spot size. These beam characteristics are important because they affect spatial resolution, image contrast, and patient dose.

- **Energy Spectrum**

The energy spectrum is the distribution of x-ray intensity as a function of energy. The emission spectrum shown in Figure 2 is for a tungsten target, and the individual peaks correspond to characteristic x-ray energies. The energy distribution of x rays is important because subject contrast—that is, the relative difference in x-ray attenuation between a structure of interest and its background—decreases as the effective energy of the x-ray beam increases. A decrease in subject contrast results in decreased image contrast. In the emission spectrum of Figure 2, the inherent and added filtration of the x-ray tube has eliminated the very low-energy x-ray photons that otherwise would contribute only dose to the patient.

The x-ray tube components that affect the shape of the energy spectrum include the anode target material (which determines the amount of bremsstrahlung x rays and position of characteristic peaks) and the tube envelope and housing (which determines inherent and added filtration). The amount of filtration is chosen such that very low-energy photons are filtered to reduce patient dose and excessive filtration of the x-ray beam (which would result in reduced image contrast) is avoided. For a given x-ray tube design, the shape of the energy spectrum is substantially affected by the voltage waveform provided by the generator.

- **Effective Focal Spot Size**

Another x-ray beam characteristic significantly affected by x-ray tube design is the effective size of the focal spot. A larger effective focal size...
The focal spot size is specified in two ways: the nominal size (that specified by the manufacturer) and the measured focal spot size (the effective size actually measured). The measured value may exceed the nominal value and still be within acceptable tolerance. Table 2 shows these tolerances for three common nominal focal spot sizes.

The focal spot size of the x-ray tube should be measured at initial acceptance testing of the imaging equipment and later as part of routine monitoring of system performance. Several devices are available for focal spot measurement, including the pinhole camera and slit camera (which directly measure the size of the focal spot) and star and bar resolution patterns. These three measurement methods are illustrated in Figure 11.

The pinhole camera allows the focal spot distribution to be measured in both dimensions as the x-ray beam is projected through a very small hole in an x-ray opaque material. The slit camera allows measurement of one dimension of the focal spot distribution, with the second dimension being measured in a second exposure after the camera is rotated 90°. The star and bar patterns show the ability of the focal spot to resolve closely lead spaced bars. In the star pattern, the spacing varies with radial distance. In the bar pattern, different bar spacings are used in different groupings of parallel bars. The measurements of the effective focal spot size should be within specified tolerances, depending on the nominal size specified by the x-ray tube manufacturer.

Table 2

<table>
<thead>
<tr>
<th>Nominal Size (mm)</th>
<th>Maximum Acceptable Measured Size Width</th>
<th>Length</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.3</td>
<td>0.45</td>
<td>0.65</td>
</tr>
<tr>
<td>0.6</td>
<td>0.9</td>
<td>1.3</td>
</tr>
<tr>
<td>1.2</td>
<td>1.7</td>
<td>2.4</td>
</tr>
</tbody>
</table>

*As specified by the National Electrical Manufacturers’ Association.

spot causes greater focal spot blurring, which results in diminished image spatial resolution. The diagram in Figure 10 shows an ideal focal spot (point source) and a larger focal spot employed at two different magnifications. Both of the focal spots that are not point sources cause blurred object edges, with the problem being more pronounced at higher magnifications. As discussed, the effective size of the focal spot is determined by the actual focal spot size (determined by the filament size and focusing cup design) and the tube angle. It also may be larger at higher filament currents, a phenomenon known as blooming.

The effective focal spot is smaller in the case of a smaller actual focal spot size or a smaller anode angle. However, there is a tradeoff. A smaller actual focal spot size means there will be a higher, instantaneous, local heat load on the target and thus a lower limit for tube exposure. A smaller anode angle also reduces the maximum available field size. Thus, minimizing the effective focal spot size may improve spatial resolution, but it may sacrifice effective field size and tube power rating (limits tube current and exposure time).
Table 3
X-ray Tube Design Parameters: Conventional Radiography versus Mammography

<table>
<thead>
<tr>
<th>X-ray Tube Design Parameter</th>
<th>Conventional Radiography</th>
<th>Mammography</th>
</tr>
</thead>
<tbody>
<tr>
<td>Operating peak voltage (kVp)</td>
<td>40-120</td>
<td>25-30</td>
</tr>
<tr>
<td>Anode material</td>
<td>Tungsten</td>
<td>Molybdenum or rhodium</td>
</tr>
<tr>
<td>Tube exit window</td>
<td>Glass</td>
<td>Beryllium</td>
</tr>
<tr>
<td>Added filtration</td>
<td>Aluminum</td>
<td>Molybdenum or rhodium</td>
</tr>
</tbody>
</table>

VARIATIONS IN X-RAY TUBE DESIGN

X-ray tubes have some common design variations that are chosen for suitability to a specific application. In cardiac imaging, a series of very short exposures (milliseconds in duration) is desirable; thus, a grid-biased tube is often used. In a conventional tube, the capacitance of the tube cables makes it impossible to turn the tube anode voltage on and off quickly. In a grid-biased tube, the filament focusing cup is used as a control grid, which allows very short exposures. In such a system, the focusing cup is at a variable negative potential relative to the filament. The resulting electric force of repulsion can be used to "pinch" the electron flow off and on very rapidly. Another common design variation is the use of a heavy-duty anode and additional cooling mechanisms (fans, oil circulators, radiators) for applications in which high heat loading of the tubes is common.

Significant design changes are seen in x-ray tubes developed for mammography. Table 3 compares key design parameters for x-ray tubes used in conventional radiography versus mammography. In mammographic x-ray tubes, a different target material is chosen for its lower K edge. The molybdenum target commonly used in mammographic tubes produces characteristic energy peaks at approximately 17.5 and 19.5 keV, and those characteristic peaks contribute as much as 40% to the total x-ray intensity (bremsstrahlung process contributes the remainder). Beryllium (which has a low atomic number) is used for the exit window in mammographic tubes to minimize attenuation that otherwise would occur with a glass exit window at the lower mammographic energies. Molybdenum (rather than aluminum) is commonly chosen as the filtration material in mammographic x-ray tubes to accentuate the characteristic portion of the mammographic energy spectrum. Rhodium is also used as both a target and filtration material in mammography. Rhodium has slightly higher-energy characteristic peaks (approximately 20.0 and 22.7 keV), thereby producing a penetrating x-ray beam that is more suitable for imaging dense breasts.

SUMMARY

The task of the x-ray tube is to convert accelerated electrons into a useful x-ray beam. Both bremsstrahlung and characteristic x rays are produced. X-ray generation is a very inefficient process, and the heat that is produced limits the design and components of the tube. The major x-ray tube components are the cathode and anode assemblies, the tube envelope, the rotor and stator (for rotating anode systems), and the tube housing. These components affect x-ray beam characteristics such as the energy spectrum and the effective focal spot size, which in turn affect image quality and patient dose. Common variations in x-ray tube design include grid-biased tubes for rapid exposures in cardiac or vascular imaging, as well as tubes composed of different target and filtration materials for use in mammography.

SUGGESTED READINGS


This article meets the criteria for 1.0 credit hour in Category 1 of the AMA Physician's Recognition Award. To obtain credit, see the questionnaire on pp 1253-1258.