# X-Ray Imaging Physics for Nuclear Medicine Technologists. Part 1: Basic Principles of X-Ray Production*

J. Anthony Seibert, PhD

*Department of Radiology, Imaging Research Center, University of California Davis, Sacramento, California*

The purpose is to review in a 4-part series: (i) the basic principles of x-ray production, (ii) x-ray interactions and data capture/conversion, (iii) acquisition/creation of the CT image, and (iv) operational details of a modern multislice CT scanner integrated with a PET scanner. Advances in PET technology have lead to widespread applications in diagnostic imaging and oncologic staging of disease. Combined PET/CT scanners provide the high-resolution anatomic imaging capability of CT with the metabolic and physiologic information by PET, to offer a significant increase in information content useful for the diagnostician and radiation oncologist, neurosurgeon, or other physician needing both anatomic detail and knowledge of disease extent. Nuclear medicine technologists at the forefront of PET should therefore have a good understanding of x-ray imaging physics and basic CT scanner operation, as covered by this 4-part series. After reading the first article on x-ray production, the nuclear medicine technologist will be familiar with (a) the physical characteristics of x-rays relative to other electromagnetic radiations, including γ-rays in terms of energy, wavelength, and frequency; (b) methods of x-ray production and the characteristics of the output x-ray spectrum; (c) components necessary to produce x-rays, including the x-ray tube/x-ray generator and the parameters that control x-ray quality (energy) and quantity; (d) x-ray production limitations caused by heating and the impact on image acquisition and clinical throughput; and (e) a glossary of terms to assist in the understanding of this information.

***Key Words:*** PET/CT; medical imaging; electromagnetic spectrum; x-ray production; x-ray tube; x-ray generator; CT

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For correspondence or reprints contact: J. Anthony Seibert, PhD, Department of Radiology, Imaging Research Center, University of California Davis, 4701 X St., Sacramento, CA 95817.

E-mail: jaseibert@ucdavis.edu

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Nuclear medicine imaging has been an integral component of the diagnostic radiology armamentarium for several decades and is undergoing a renaissance of importance as the world of molecular imaging and genomics becomes the current research topic of interest. As those involved in nuclear diagnostic medicine have always known, the nature of nuclear medicine has been and will continue to be molecular and provide metabolic and physiologic information as well. However, a shortcoming of nuclear medicine procedures is a lack of spatial resolution and anatomic detail, in which x-ray projection imaging, CT, and MRI excel. To get the best of structural and functional imaging requires dual-modality imaging, which allows accurate registration of anatomy and physiology, the ability to visualize biologic differences between diseased and normal tissues, and quantification of the functional status of organs and tissues. This is particularly relevant for oncologic diagnosis and staging, therapy planning, and outcomes assessment. Not surprisingly, tremendous interest in dual-modality imaging, particularly with respect to combined PET and CT, has resulted in the implementation and adoption of dedicated PET/CT and SPECT/CT imagers in clinics throughout the nation, not only in diagnostic radiology but also in oncology and nuclear medicine (*1*). For the nuclear medicine technologist, therefore, added expectations, responsibilities, education requirements, and opportunities will be part of an expanding future for those involved and willing to participate in dual-modality imaging. A basic understanding of x-ray imaging physics is important for the nuclear medicine technologist; the goal of this series of papers is to provide this information.

This article reviews the topic of x-ray production and control of the x-ray beam quality and quantity through the use of x-ray tubes, x-ray generators, and beam-shaping devices. Part 2 of this series investigates the characteristics of x-ray interactions, the formation of the projection image, image contrast, signal-to-noise ratio, and radiation dose. Part 3 explores the CT scanner design; acquisition and reconstruction of the projection data into a tomographic slice; characteristics of the CT image in terms of resolution,

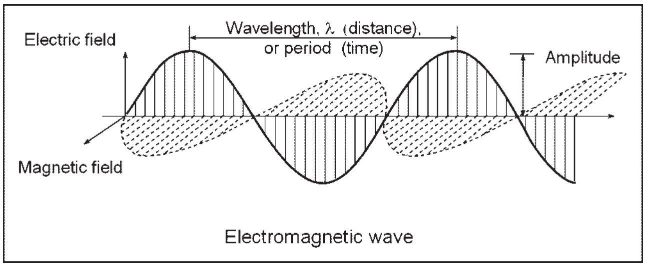


**FIGURE 1.** Electromagnetic radiation is described as a cyclic repeating wave having electrical and magnetic fields with amplitude (peak value from the average) and period (time between repeating portions of the wave). Frequency equals the number of cycles per second, and the wavelength is the distance between repeating points as determined from the frequency and velocity (see text for relationship between velocity, wavelength, and frequency).

contrast and noise, quantitative data extraction, and introduction of multislice CT scanners and their impact on image acquisition and speed. The final article, part 4, covers the physics and technical acquisition issues relevant to image fusion of nuclear medicine images acquired with SPECT and PET to those acquired with CT. Operational details of a combined PET/CT scanner are reviewed.

Medical x-rays for diagnostic imaging have been used for over a century, soon after the published discovery by Roentgen in 1896. Then, as now, the underlying basis for medical applications of x-rays depends on the differential attenuation of x-rays when interacting with the human body. A uniform x-ray beam incident on the patient interacts with the tissues of the body, producing a variable transmitted x-ray flux that is dependent on the attenuation along the beam paths. This produces a superimposed "shadow" of the internal anatomy. An x-ray–sensitive detector captures the transmitted fraction and converts the x-rays into a visible projection image to provide anatomic information to a medical practitioner. More recently, in the early 1970s, engineers and physicists introduced the ability to provide a true 3-dimensional representation of the anatomy by the acquisition of multiple, angular-dependent projections synthesized into tomographic images with computer algorithms in the computer. CT revolutionized the use of x-rays in diagnostic medical imaging and propelled the use of computerized image acquisition in diagnostic radiology for medical diagnosis. For all x-ray imaging, the common entity is the controlled x-ray beam of known energy and quantity. X-rays are electromagnetic radiation of high energy.

## ELECTROMAGNETIC RADIATION

The use of electromagnetic radiation for CT and SPECT/PET is intrinsic as the carrier of image content, when interacting with (x-ray attenuation/transmission) or emitting (metabolic agents labeled with γ-ray emitters) within the patient. Electromagnetic radiation is characterized as periodic cyclic waves that contain both electrical and magnetic fields and can be described in both time and space, using period (time) and wavelength (distance) between repeating points of the wave (Fig. 1). The *cycle* represents the repeating unit of the sinusoidal electromagnetic wave. The length of time that 1 cycle occupies is the *period*, expressed in seconds, and the number of cycles per second is the *frequency*, $f$, usually expressed in inverse seconds ($s^{-1}$) or Hertz (Hz). Frequency and period are inversely equal: $f$ = 1/period. Electromagnetic radiation travels at a *velocity*, C, of $3.0 \times 10^8$ m/s in a vacuum. The velocity in a vacuum is constant but will vary slightly in other materials. The distance traveled by 1 cycle is the *wavelength*, λ, typically described in units of nanometers ($10^{-9}$ m). Wavelength is the product of velocity and period and, therefore, is inversely related to the frequency. The relationship between velocity, frequency, and wavelength is given as $C = \lambda f$. Details of the electromagnetic spectrum in terms of wavelength, frequency, energy, and description are diagrammed in Figure 2.

At higher energies and extremely short wavelengths (e.g., x-rays and γ-rays), electromagnetic radiation manifests particle-like traits, whereby interactions are collisional in nature. This means, for instance, that an x-ray photon with sufficient energy can interact with and remove electrons bound to an atom (the process of ionization). This is why x-rays and γ-rays are referred to as ionizing radiation. Photon energy is directly proportional to the frequency of the wave, and higher frequency is higher energy, as given by $E = h f$, where E is energy and $h$ is Planck's constant =

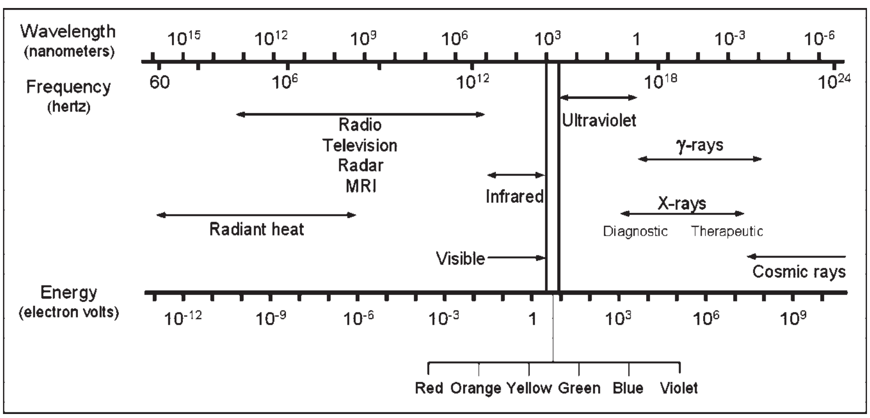


**FIGURE 2.** The electromagnetic spectrum, presented as a function of wavelength, frequency, and energy. X-rays and γ-rays comprise the high-energy portion of the electromagnetic spectrum.

$6.62 \times 10^{-34}$ J-s $= 4.13 \times 10^{-21}$ eV-s. The joule (J) and the electron volt (eV) are common units of energy. In diagnostic imaging the important unit is the electron volt, where 1 eV is equal to the kinetic energy gained by an electron in a vacuum accelerated by a potential difference of 1 V.

Einstein's theory of relativity, $E = mc^2$ (energy is equal to the product of mass and the square of the speed of electromagnetic radiation), states that mass and energy are interchangeable and that mass and energy are conserved in an interaction. This is clearly exemplified in nuclear medicine PET studies, when a positive electron emitted from a proton-rich unstable atom interacts with a negative electron, resulting in an antimatter/matter conversion of the total mass of the positron and the electron into energy, with the emission of two 511-keV $\gamma$-ray photons (the rest mass of an electron is equivalent in energy to 511 keV). In radioactive elements with unstable nuclei, the energy released during the decay of the nucleus to a more stable (lower energy) form is often in the form of a $\gamma$-ray photon of discrete energy. X-rays and $\gamma$-rays are forms of high-energy electromagnetic radiation that are indistinguishable, except for their origin. $\gamma$-rays, by definition, originate from excess energy released by the atomic nucleus when decaying to a more stable daughter, whereas x-rays originate via interactions external to the nucleus. Like $\gamma$-rays, x-rays are created as a result of energy conversion and conservation.

## X-RAY TUBES AND X-RAY PRODUCTION

X-rays result from the conversion of the kinetic energy attained by electrons accelerated under a potential difference—the magnitude of which is termed voltage with units of volts (V)—into electromagnetic radiation, as a result of collisional and radiative interactions. An x-ray tube and x-ray generator are the necessary components for x-ray production and control. The x-ray tube provides the proper environment and components to produce x-rays, whereas the x-ray generator provides the source of electrical voltage and user controls to energize the x-ray tube. Basic components of an x-ray system are illustrated in Figure 3. In the x-ray tube, 2 electrodes, the cathode and anode, are situated a small distance apart (about 1–2 cm) in a vacuum enclosure called the insert, made of either glass or metal. Connected to the cathode and the anode are negative and positive high-voltage cables, respectively, from the x-ray generator. A separate, isolated circuit connects the cathode filament (a coiled wire structure similar to a coiled light-bulb filament) to a low-voltage power source. To produce x-rays, a specific sequence of events is required.

The first step for x-ray production requires free electrons to be available in the evacuated environment of the x-ray tube insert to allow electrical conduction between the electrodes. The electron beam emitter consists of the cathode filament set centrally in a slot machined in a metal focusing cup (cathode cup). Activating the filament circuit causes intense heating of the filament due to its electrical resistance and releases electrons by a process known as thermionic emission. A larger filament current produces more heat and releases a greater number of electrons. Electron accumulation occurs at the filament surface, creating a buildup of negative charge that prohibits further electron release because of repulsion forces. The electron cloud distribution is maintained at equilibrium by the surrounding negatively charged focusing cup.

The second step involves the application of a high voltage, typically ranging from 50,000 to 150,000 V (50–150 kV) supplied by the x-ray generator to the cathode and anode. Upon activation, electrons are immediately accelerated to the electrically positive anode along a path determined by the filament and focusing cup geometry. Continuous electron emission continues from the filament surface at a rate dependent on the filament temperature (i.e., the filament current) during the exposure. Tube current, defined as the number of electrons traveling between the electrodes, is expressed in milliampere (mA) units, where 1 A is equal to $6.24 \times 10^{18}$ electrons/s and 1 mA $= 6.24 \times 10^{15}$ electrons/s. Typical tube currents for CT operation have a

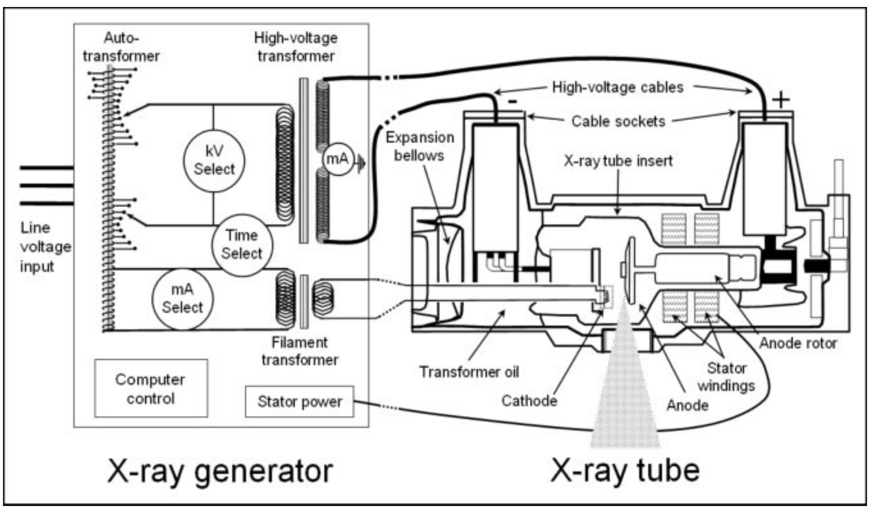


**FIGURE 3.** X-ray generator and x-ray tube components are illustrated. The x-ray generator provides operator control of the radiographic techniques, including tube voltage (kVp), tube current (mA), and exposure duration, and delivers power to the x-ray tube. The x-ray tube provides the environment (evacuated x-ray tube insert and high-voltage cable sockets), source of electrons (cathode), source of x-rays (anode), induction motor to rotate the anode (rotor/stator), transformer oil and expansion bellows to provide electrical and heat build-up protection, and the tube housing to support the insert and provide protection from leakage radiation.

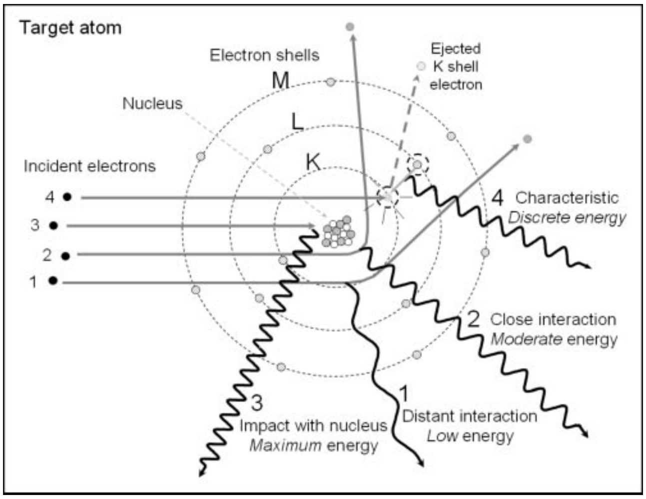


**FIGURE 4.** X-ray production by energy conversion. Events 1, 2, and 3 depict incident electrons interacting in the vicinity of the target nucleus, resulting in bremsstrahlung production caused by the deceleration and change of momentum, with the emission of a continuous energy spectrum of x-ray photons. Event 4 demonstrates characteristic radiation emission, where an incident electron with energy greater than the K-shell binding energy collides with and ejects the inner electron creating an unstable vacancy. An outer shell electron transitions to the inner shell and emits an x-ray with energy equal to the difference in binding energies of the outer electron shell and K shell that are "characteristic" of tungsten.

selectable range of 50–300 mA, depending on the type of examination and required image quality. Each electron attains a kinetic energy (in keV) equal to the applied tube voltage, which typically is set to a single value that ranges from 50 to 150 kV depending on the examination. (In CT operation, 120–130 kV is most often used, but 80-, 100-, and 140-kV settings are also available on some CT scanners.) Thus, the tube voltage (kV), tube current (mA), and exposure duration (s) are user-selectable parameters for x-ray production. Often, the combination of tube current and exposure time in milliampere-seconds (mAs) is provided as part of the technique or protocol. For instance, with CT scanner operation, typical acquisition techniques are quoted in kVp and mAs/slice.

Step 3, x-ray production, occurs when the highly energetic electrons interact with the x-ray tube anode (also known as the target). Targets used in x-ray tubes are generally made of tungsten, which has 74 protons in the nucleus. In rare (0.5% or 5/1,000) events, an electron comes in close proximity to the nucleus of a target atom and experiences attractive forces due to the positive charge of the protons in the nucleus. This combined positive charge decelerates and changes direction of the electron, the magnitude of which strongly depends on the impact parameter distance, as 1/distance (*2*). As to the magnitude of this distance, the tungsten atom has a spheric shape with a radius of about $10^{-8}$ cm, and its nucleus has a radius of $<10^{-12}$ cm, indicating that the atom is comprised of mostly empty space. The interaction distance for a deceleration event by the incident electron is thus on the order of $10^{-12}$ cm. Kinetic energy lost is converted to electromagnetic radiation with equivalent energy in a process known as bremsstrahlung (a German term meaning "braking radiation"), as shown in Figure 4. Closer interactions with the nucleus cause a greater deceleration and result in higher x-ray photon energy, but the probability decreases as the interaction distance decreases. In extremely rare instances, the incident electron gives up all of its kinetic energy when stopped by the nucleus, producing the maximum x-ray energy possible. The output is a continuous spectrum of x-ray energies with maximum x-ray energy (in keV) determined by the peak potential difference (in kVp). A larger number of low-energy x-rays are produced in the output spectrum, simply due to the lower probability of interaction closer to the nucleus. A dartboard analogy can help explain this phenomenon, as randomly thrown darts (the electrons) have equal probability to land anywhere on the board. The lowest probability of interaction is a bulls eye (e.g., incident electron totally decelerated, producing the highest x-ray energy), and moving outward, the annular rings emanating from the center become larger and accommodate more darts at greater distance (e.g., electrons are decelerated less and produce greater numbers of lower x-ray energies). Thus, a spectrum of x-rays is produced with a minimum number at the peak energy, and linearly increasing in number with decreasing energy (the unfiltered bremsstrahlung spectrum) (*2–4*). However, lower energy x-rays are more easily attenuated (filtered) from the beam exiting the x-ray tube port, and the measured spectrum peaks at intermediate energy and goes back to zero at the lowest x-ray energies, as shown in Figure 5 for several spectra produced with different acceleration voltages. The average x-ray energy in a typical x-ray spectrum is about one-third to one-half peak energy, dependent on the amount of filtration placed in the beam.

Another possible interaction of incident electrons with the target is the removal of inner shell electrons from the tungsten atom. All elements have atoms with the number of protons equal to the atomic number, and an equal number electrons residing in electron shells. The innermost shell, designated K, has a high binding energy to maintain stability of the 2 occupying electrons. Outer electron shells (L, M, N. . . . ) have successively lower binding energies and

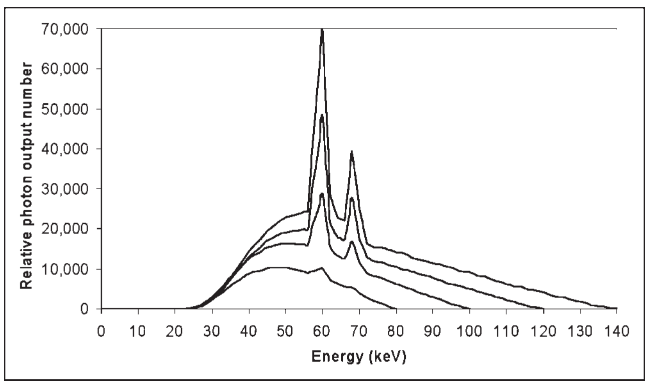


**FIGURE 5.** Bremsstrahlung and characteristic radiation spectra are shown for a tungsten anode with x-ray tube operation at 80, 100, 120, and 140 kVp and equal tube current.

greater number of electrons. For tungsten, binding energies of the K, L, and M shells are 69.5, 11.5, and 2.5 keV, respectively. A highly energetic incident electron can interact with and remove a K-shell electron if it has at least 69.5-keV kinetic energy and will leave an electron vacancy. Because the atom is now energetically unstable, electrons from adjacent (the L shell) or nonadjacent (M, N, O shells) will readily transition and fill the K-shell vacancy, as shown in Figure 4, event number 4, depicting the creation of characteristic radiation. As a result, a discrete energy x-ray photon is created with energy equal to the difference in binding energies. For instance, an L-to-K electron transition produces a characteristic x-ray of 69.5–11.5 = 57.0 keV. Since each element has different electron binding energies, the emitted x-ray energies are characteristic of the element (tungsten, in this example). These characteristic x-rays generate the monoenergetic spikes added to the continuous spectrum, as seen in Figure 5. Note that K-characteristic radiation from a tungsten target will occur only when the x-ray tube is operated at a voltage of >69.5 kVp. As the tube voltage is increased above the minimum value, characteristic x-ray production becomes a greater fraction of the x-ray spectrum.

The anode in most diagnostic x-ray tubes is comprised of tungsten, chiefly because of its high atomic number (Z = 74) and extremely high melting point—necessary for efficient x-ray production and tolerance of high power deposition, respectively. Two major anode designs include a simple, fixed geometry or a more elaborate, rotating configuration as shown in the x-ray tube diagram in Figure 3. Most prevalent is the rotating anode, comprised of a tungsten disk attached to a bearing-mounted rotor within the x-ray tube insert and stator windings outside of the insert. The rotor and stator comprise the induction motor that rotates the anode disk at angular frequency of 3,000 or 10,000 revolutions/min. When the x-ray tube is energized, a delay of about 1–2 s allows the anode to reach operating speed before high voltage is applied. Rotating the anode allows a large surface area over which heat is spread, providing an ability to tolerate greater heat deposition and to produce more x-ray photons per unit time compared with a fixed anode.

The focal spot is the area of electron interaction and emanation of x-rays from the target surface. Typical dimensions are nominal sizes of 1.0- to 1.2-mm (large) and 0.3- to 0.6-mm (small) focal spots, where nominal encompasses a range of focal spot sizes that are specified as acceptable according to manufacturer standards (*5*). Ideally, the use of small focal spots is preferred to minimize geometric blurring of patient anatomy with magnification. However, the small focal area constrains x-ray tube output and heat loading factors, mainly due to heat concentrated in a small area. Larger focal spots have higher instantaneous x-ray production capacity and are preferred, as long as blurring does not adversely affect resolution. CT scanners usually have larger focal spots (e.g., 1.2-mm nominal size), which still provide good geometric resolution that is compatible with the sampling resolution of the discrete CT detector array. Focal spots vary in size with projected image location ("line-focus principle") and, because of the reflection geometry of x-ray production, radiation intensity across the projected x-ray field in the cathode-to-anode direction varies from high to low intensity ("heel-effect") and is most severe on the anode side of the field to the point that the x-ray beam intensity is reduced to zero (field cutoff). These 2 phenomena are consequences of the anode surface angle made with respect to the central axis of the emitted x-ray beam. The anode angle is a fixed value ranging from 7° to 20°, the choice of which depends on the required field coverage at a particular focal spot to detector distance. Orientation of the x-ray tube cathode–anode axis in fixed x-ray tube/detector systems considers the field intensity variations caused by the heel effect. In CT scanners, for instance, the tube is mounted so that the heel effect is minimized by orienting the cathode–anode axis in the slice acquisition direction (perpendicular to the tube travel direction). Subsequent articles in this series will point out clinically pertinent issues related to focal spot size and heel effect.

A collimator assembly, constructed with movable lead shutters, is situated adjacent to the x-ray tube output port to define the x-ray beam shape incident on the patient. For CT, the collimator shutters determine the slice thickness setting for a specific examination. A collimator light or laser beam positioned at a "virtual" focal spot location provides a visible indication of the x-ray beam. Important for CT operation is the coincidence of the slice thickness defined by the collimators to the light beam and the x-ray profile transmitted to the detector array, which must be periodically verified for accuracy during regular quality control checks.

X-rays are emitted in all directions from the anode structure, but only a small fraction of the reflected x-rays that emerge through the collimator-defined area are used for image formation, and all other x-rays must be attenuated. Protection from leakage radiation is provided by a lead-shielded x-ray tube housing, which absorbs essentially all but those x-rays emerging from the x-ray tube port and collimator assembly. Manufacturers are required to design x-ray tubes and housing assemblies to meet the federal regulations for x-ray systems described in the Code of Federal Regulations (*6*).

In comparison with the $\gamma$-ray fluence per curie of activity (number photons per unit area per unit time) emitted by a radioactive material, the x-ray fluence from an x-ray tube is many orders of magnitude larger. For example, a typical CT scanner operating at 120 kVp produces an output exposure in air of ~25 mR (milliroentgen) in 1 s per mA of tube current at 50 cm. This exposure is equivalent to an x-ray fluence of $6.7 \times 10^6$ x-ray photons/mm$^2$. A radioactive point source at a distance of 50 cm and an activity of $3.7 \times 10^{10}$ Bq (1 Ci) has a fluence of $\sim 1.2 \times 10^4$ photons/mm$^2$ over the same time interval. Therefore, from this simple analysis, at 50 cm, 1 mA of tube current (CT tube at 120 kVp) is equal to about 2.072 $\times$ $10^{13}$ Bq (560 Ci) of activity!! When a typical tube current of 200 mA is used for scanning, the equivalent activity is increased by 200 times to 4.14 $\times$ $10^{15}$ Bq (112,000 Ci), demonstrating the extremely high photon fluence of the x-ray tube compared with a radioactive source. There are obvious differences as well, in terms of examination time, detector characteristics, half-life of a radioactive material, and radiation dose among others.

## X-RAY GENERATOR

X-ray generators supply the electrical power to the x-ray tube and provide selection of the technique parameters. Control of x-ray energy and quantity is attained through adjustments of the voltage potential in kilovolts (kV), the x-ray tube current in milliamperes (mA), and the exposure time in seconds (s), which are user-adjusted at the x-ray generator console. Several electrical circuits and voltage transformers within the x-ray generator assembly provide this capability. Figure 3 illustrates the major x-ray generator components. High voltage is obtained with a "step-up" transformer, comprised of input (primary) and output (secondary) wires wound about a common iron core. Transformers operate on the principle of mutual induction, which means that a moving charge (e.g., an electron in a wire moving in a given direction and velocity) generates a magnetic field and, correspondingly, a changing magnetic field induces electrons to flow in a wire immersed in the field. Simple transformers are comprised of 2 electrically insulated primary and secondary wire windings, each wrapped around opposite sides of an iron core with a known number of turns. As alternating voltage and current are applied to the primary windings, the associated changing magnetic field permeates the iron core and induces an electromotive force (voltage) on the secondary windings. Secondary voltage amplitude is proportional to the secondary-to-primary turns ratio; for 1,000 secondary turns to 1 primary turn of a step-up transformer, the input voltage is amplified by a factor of 1,000 (and the current is reduced by the same factor to maintain overall power). Similarly, a "step-down" transformer has a greater number of primary-to- secondary turns, and the output voltage is reduced. High-voltage step-up transformers are used for supplying the x-ray tube voltage, and low-voltage step-down transformers are used for supplying power to the filament circuit in the x-ray generator. A third type of transformer, the autotransformer, is also used in the x-ray generator, which permits voltage selection on an incrementally smaller scale, usually on the order of volts, and is part of the circuit that controls the input voltage to the high-voltage transformer.

X-ray generator configurations include single-phase, 3-phase, high-frequency, and constant-potential designs. Differences in internal electrical components and transformer circuitry result in voltage waveforms that vary significantly with time as in the single- phase system or with a nearly constant voltage as characterized by the constant-potential system. High-frequency generators have been the most widely used over the past decade, chiefly for their superb accuracy, self-calibration, near constant-potential waveform, small size, reliability, and modular design. All modern CT scanners use generators based on the high-frequency inverter generator. Specific issues related to generator design and function can be found in separate references (*2–4*).

Exposure timing circuitry starts and ends the application of high voltage across the x-ray tube electrodes. Exposure duration, defined as the time x-rays are being produced for image formation, varies depending on the diagnostic imaging procedure and modality being used. For radiography, exposure durations are extremely short, typically $\leq$100 ms, combined with large tube current (200–1,000 mA) to achieve high photon fluence (directly proportional to the mAs). High-power electronic switches (called triodes or tetrodes) are placed in the high-voltage circuit and can turn on and off the power rapidly ($<$1 ms is typical). When coupled with an electronic timer, accurate control of exposure is provided, which can reduce voluntary and involuntary patient motion artifacts and preserve image quality. For fluoroscopy, the exposure duration is continuous and is usually initiated and terminated by foot pedals that signal mechanical contactor switches in the low-voltage section of the x-ray generator. Timer accuracy is somewhat poor (no greater than about 8 ms or 1/120 s), but for fluoroscopy applications (1- to 5-mA tube current) this is more than acceptable. For CT operation, exposure durations of 0.4 to tens of seconds with tube currents of 50–400 mA are required for typical operation, depending on the type of examination and acquisition protocol. Electronic timing is computer controlled to synchronize the x-ray exposure with the data collection subsystem.

## TUBE HEAT LOADING, COOLING, AND PROTECTION

Buildup of heat energy is the major limit to instantaneous x-ray production and x-ray tube longevity (the latter by focal track scarring or rotor-bearing failure). Continuous x-ray production depends on heat dissipation by the anode assembly and tube housing. The energy deposited into the x-ray tube is a product of the tube current (amperes), tube voltage (volts), and exposure time (seconds), defined in joules (the SI unit of energy). Often, the tube heat loading will be expressed in heat units (HU), an historical term defined as the product of tube current (mA), peak tube voltage (kVp), and exposure time (s). Voltage waveform variability for single-phase x-ray generators have an average voltage of about 70% of the peak energy, and the HU underestimates the energy deposited for 3-phase and high-frequency generators by ~1.35 or 35%, requiring a "fudge-factor" multiplier of 1.35.

The power rating of x-ray generators and x-ray tubes is specified in kilowatts (kW), where the watt is defined as the amount of energy per unit time that the generator can deliver or the x-ray tube can receive to the focal spot area.

Necessary generator power is based on the intended imaging applications. In general, examinations that require high instantaneous power and photon fluence, such as interventional radiology examinations, must have matching power delivery capabilities, and 80- to 120-kW generators are specified. CT scanners must deliver a significant power load over an extended time (e.g., 300 mA for 30–50 s), and typical power ratings of 30–80 kW are therefore necessary. On the receiving end is the x-ray tube, where power deposition capabilities depend on the size of the focal spot, the diameter and angle of the anode, and the anode rotation speed. During extended x-ray tube operation, heat energy accumulates, and ultimately limits x-ray production if the cooling rates of the anode and the tube housing are insufficient. These often result in examination delays until sufficient latent heat has been dissipated, which is a particular problem for busy CT operations and large patient volume scanning procedures. Modern CT tubes designed for high-throughput scanners have a large anode volume to serve as a heat reservoir and efficient heat exchanger systems to provide rapid cooling (e.g., 6–8 MHU capacity and dissipation of >800 kHU/min). From a clinical perspective, the ability to scan a patient without stopping the procedure to allow for tube cooling is a high priority. Thus, it is important to understand the potential limitations of CT scanner operation and to ensure that a tube capable of performing the most demanding acquisition protocol is installed (part of the purchasing specification requirements). Fortunately, with most modern CT scanners, software algorithms assist in the determination of the tube heating/cooling limitations and, if necessary, will require the system to wait an appropriate length of time before initiation of the acquisition sequence.

Dramatic improvements in CT multirow detector acquisition arrays, data acquisition speed, simultaneous table travel, and volumetric imaging have pushed x-ray tube technology to be able to withstand extreme centrifugal forces in addition to high heat. Currently, the fastest CT systems can rotate the x-ray tube fully around the patient (360°) in as little as 0.4 s. Based on the typical radius of rotation within the CT gantry, rotation speed, and mass of the x-ray tube, the forces acting on the x-ray tube components (mainly the anode) are on the order of 13 gravitational forces (1 G represents the gravitational force on the earth). A 45-kg (100 pound) object at rest would weigh 585 kg (1,300 pounds!) when rotating at 2.5 revolutions per second (0.4 s per revolution). To operate in this environment, CT x-ray tubes have anodes that are supported on both sides by a single shaft which runs through the center of the anode disk and firmly supports the high-speed rotation of the anode, even when extremely hot. CT tubes also have enhanced heat exchange efficiency and rapid cooling rates. From a clinical perspective, this overcomes many of the shortcomings caused by slow acquisition times and heat loading limitations of older CT systems.

## CONCLUSION

X-rays are the basic radiologic information carrier for projection radiography and CT. X-ray production is a process that involves conversion of highly energetic electrons into a polyenergetic x-ray energy spectrum and monoenergetic characteristic x-rays (the latter when energetically feasible) in a controlled process within the x-ray tube. Selection of x-ray energy and number is controlled by user adjustment of x-ray tube voltage, tube current, and exposure duration at the x-ray generator. A uniform output beam emanating from the focal spot of the tube is shaped and controlled by collimator shutters to expose the area of interest on the patient, and the transmitted x-ray flux is subsequently recorded to form the projection image. For CT scanners, the x-ray spectra are usually produced with tube voltages ranging from 120 to 140 kVp, giving average energies of 50–70 keV. A major limitation is the inefficiency of x-ray production and the excessive heat produced for a typical CT procedure, which often requires a waiting time during the examinations to allow the tube to cool. Multirow detectors and helical scanning (simultaneous tube rotation and table translation) improve x-ray utilization and reduce the overall requirements for x-ray tube heat loading needed to cover an anatomic area.

Part 2 in this series reviews x-ray interactions with the patient, generation of "subject" contrast, and detection of the transmitted x-rays that ultimately render the anatomic detail in the x-ray projection image and in the CT image.

## GLOSSARY

*Anode*.

In an x-ray tube, the anode (also known as the target) is the positively charged electrode that attracts free electrons to subsequently produce x-rays. Fixed anodes and rotating disk designs are common.

*Bremsstrahlung*.

A word of German origin meaning "braking radiation," which describes the process of x-ray production resulting from the deceleration of highly energetic electrons in the vicinity of an atomic nucleus of the target (anode) of the x-ray tube.

*Bremsstrahlung spectrum*.

A recording of the continuous x-ray energies produced as a result of the bremsstrahlung interaction. This is also known as a polychromatic x-ray spectrum.

*Cathode*.

In an x-ray tube, the cathode (also known as the source) is the negatively charged electrode, typically comprised of a filamentous structure that produces free electrons as a result of heating by electrical resistance.

*Characteristic x-rays*.

Monoenergetic x-rays produced by the ejection of a K-shell electron from an element (e.g., the target of an x-ray tube) and the subsequent filling of the vacant electron by electrons from shells with lower binding energies, with energy equal to the difference in the binding energies of the shells.

*Electromagnetic spectrum*.

The electromagnetic radiation spectrum is the complete range of the wavelengths of electromagnetic radiation, beginning with the longest radio waves (including those in the audio range) and extending through visible light (a very small part of the spectrum) all the way to the extremely short γ-rays that are a product of radioactive atoms.

*Energy*.

The capacity of a physical system to do work. The common symbol is the uppercase letter $E$, and the standard unit is the joule, J. The common unit used for x-rays and γ-rays is the electron-volt, eV. Two main forms of energy are potential energy and kinetic energy. Potential energy is the energy stored in a system—for example, a stationary object in a gravitational field—or a stationary charged particle in an electric field has potential energy. Kinetic energy is observable as motion of an object, particle, or set of particles—for example, the motion of an electron in an electric field.

*Fluence*.

The number of photons per unit area (e.g., $mm^2$) specified at a given distance. For a point source with equal probability of emission at any angle, the fluence is calculated as the number of photons released from the source to the surface area of a sphere ($4\pi r^2$), where r is the radius at the specified distance.

*Fluence rate*.

The fluence per unit time (e.g., per second).

*Focal spot*.

The area on the x-ray tube anode where electrons from the cathode interact and produce x-rays.

*Focal track*.

The annular area on a rotating anode over which the stationary electron beam interacts during the production of x-rays.

*Frequency*.

The number of complete cycles per second, measured in hertz (1 cycle per second), kilohertz, megahertz, gigahertz, or terahertz.

*γ-Rays*.

Ionizing electromagnetic radiation of very short wavelength and energy $>10$ eV; by definition, γ-rays are the result of nuclear decay or nuclear energy release and are thus said to be of "nuclear" origin.

*Heat unit (HU)*.

The unit, equal to the product of the kVp and mA, is used to describe the accumulation or dissipation of heat energy in an x-ray tube anode or x-ray tube housing. Heat unit ratings are typically in the range of kHU to MHU. An alternate (and preferred) unit is the joule, which takes into account the characteristics of the high- voltage waveform.

*Joule*.

One joule is defined as the amount of energy exerted when a force of 1 newton is applied over a displacement of 1 m. One joule is the equivalent of 1 watt of power radiated or dissipated for 1 s.

*kVp*.

Kilovolt peak: the peak voltage applied to the x-ray tube, equal to 1,000 times the voltage; voltage varies with time with x-ray generator-produced waveforms. The average voltage ($kV_{avg}$) is less than the peak voltage for all but constant-potential waveforms.

*keV*.

Kiloelectron volt: 1,000 times the base unit of energy, the electron-volt. This is a common unit of energy used for describing the energy of x-ray photons. One eV is equal to the kinetic energy attained by an electron accelerated by a potential difference of 1 volt.

*Leakage radiation*.

Results from the production of x-rays in directions other than the tube output port and is reduced by attenuating materials placed in the x-ray tube housing such that the total radiation is kept below a minimum value required by federal laws.

*mA*.

Milliampere: 1/1,000 of an A, unit of electrical current, which describes the transport of charge (positive or negative) per second as a result of an applied potential difference; 1 A of current represents 1 coulomb of charge/second, where 1 coulomb is equivalent to $6.24 \times 10^{18}$ electrons. X-ray tube operation uses tube currents ranging from $<1$ mA (for fluoroscopy) up to 1,000 mA (for radiography).

*Magnification*.

Ratio of the image size to object size in an x-ray projection from a point source. X-ray systems use a distributed source (focal spot), which causes geometric blurring of magnified objects (not in contact with the detector) and loss of detail.

*Period*.

The time duration of 1 cycle of a sinusoidal (periodic) wave. The period and the frequency are inversely equal.

*Potential difference*.

The voltage placed across 2 electrodes, as in an x-ray tube, measured in volts.

*Power*.

Electrical power is the rate at which electrical energy is converted to another form, such as motion, heat, or an electromagnetic field. The common symbol for power is the uppercase letter P. The standard unit is the watt, symbolized by W. In x-ray circuits, the kilowatt (kW) is often specified, where 1 kW $=$ 1,000 W.

*Rotor*.

The cylindric component of an induction motor mounted on bearings inside the evacuated x-ray tube, comprised of alternating iron/copper core. The rotor, attached to the anode disk, turns in response to a rotating magnetic field produced by the stator outside of the x-ray tube envelope.

*Stator*.

A wire-wrapped "donut" placed around the thin neck of the x-ray tube insert and adjacent to the rotor that produces a rotating magnetic field when energized with an electric current. The rotating magnetic field, permeable to the x-ray tube insert materials, drags the ferromagnetic iron structures of rotor assembly, causing the attached anode disk to rotate.

*Thermionic emission*.

The process by which electrons are liberated from an intensely heated element, as in the cathode filament of an x-ray tube, by the filament circuit.

*Tube rating charts*.

A graphic or tabular description of the allowable combinations of kVp, mA, and exposure time for an x-ray tube with specific power tolerance and dissipation capabilities dependent on several factors, including the size of the focal spot/anode angle, the anode rotation speed, and the anode diameter. Also included in this are anode and housing cooling charts, which describe the rate of heat dissipation as a function of accumulated heat energy.

*Voltage*.

Voltage, also called *electromotive force*, is a quantitative expression of the potential difference in charge between 2 points in an electrical field. The standard unit is the volt, symbolized by an uppercase letter V. Voltage can be direct or alternating. A direct voltage maintains the same polarity at all times. In an alternating voltage, the polarity reverses direction periodically.

*Watt (W)*.

The standard unit of power (or energy per unit time) that is the equivalent of 1 J/s. The watt is used to specify the rate at which electrical energy is dissipated or the rate at which electromagnetic energy is radiated, absorbed, or dissipated.

*Wavelength*.

The distance between identical points in the adjacent cycles of a waveform signal propagated in space or along a wire.

*X-rays*.

Electromagnetic radiation with energy ranging from >10 eV through extremely high-energy, short wavelengths; by definition, x-rays are created by energy conversion by interactions outside of the atomic nucleus ("extranuclear" origin).

*X-ray tube*.

An evacuated container (typically known as the tube insert) comprised of 2 electrodes (cathode and anode) in which x-rays are produced and surrounded by the tube housing, which provides electrical safety and radiation protection from leakage x-rays.

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