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Biomedical Equipment Journeyman

Volume 4. Diagnostic Imaging and Related Support Equipment

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THIS IS THE fourth and final volume of CDC 4A251B. The material contained within this volume is extremely important and should garner your full attention. The journey through this volume will take you through three units. This volume begins with the fundamentals of X-radiation (X-ray), and concludes with special equipment and procedures.

Unit 1 covers radiation principles, including how X-rays are generated, ways radiation is detected and measured, and means of personal protection. The unit also covers material on the principles of X-ray generation, which we'll go over many X-ray circuits and components to familiarize you with the basic X-ray system. This unit will also introduce you to concepts of digital radiography.

Unit 2 moves on to X-ray systems and will familiarize you with more specific pieces of diagnostic imaging equipment, including fixed and mobile X-ray systems, mammographic systems, and dental X-ray. This unit also contains information on various support equipment related to diagnostic imaging equipment, including contrast injectors, computerized radiography (CR) plate readers, laser imagers, diagnostic imaging monitors, and picture archiving and communication systems. The unit concludes with some information related to the maintenance of X-ray equipment, a lesson on non-invasive X-ray test equipment, and a breakdown of how to perform a post calibration radiation inspection.

The volume concludes with Unit 3, which covers specialized diagnostic imaging equipment and procedures. While your current facility may not have all of the equipment contained within this unit, it is important that you are familiar with the equipment, as you will certainly see it at some time in the future. This unit includes various tomographic systems including conventional tomography, computed tomography, and cone beam computed tomography systems. The remainder of the unit focuses on nuclear medicine, magnetic resonance imaging systems, and diagnostic ultrasound equipment.

A glossary is included for your use.

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NOTE:

In this volume, the subject matter is divided into self-contained units. A unit menu begins each unit, identifying the lesson headings and numbers. After reading the unit menu page and unit introduction, study the section, answer the self-test questions, and compare your answers with those given at the end of the unit. Then complete the unit review exercises.

Acknowledgment

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Delmar Learning, a part of Cengage, Inc	ISBN 9781439058725, Principles of Radiographic Imaging, Carlton/Adler, 5e	1–2, 1–4, 1–5, 1–8, 1–25, 1–27 to 1–38, 1–40, 1–46, 1–48, 1–49, 2–2, 2–4, 2–7, 2–9, 2–10, 2–12, 2–13, 2–17 to 2–20, 2–23 to 2–25, 3–3, 3–5 to 3–8, 3–10, 3–14, 3–16, 3–26, 3–27, 3–30, and 3–41
Gammex Inc.	The QA Cookbook for Ultrasound and Doppler 403 Flow Phantom Manual (Model 1425B)	3–47 to 3–54
Fluke Biomedical Corporation/Raysafe Inc.	Raysafe X2 X-ray Measurement System Raysafe/Xi Prestige X-ray Measurement System/Raysafe DXR+ Direct X-ray Ruler	2–28, 2–31, 2–32, 2–41, 2–44, and 2–45
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Page

Please read the menu for Unit 1 and begin ➔

Unit 1. X-ray Fundamentals

AS a Biomedical Equipment Journeyman, you should understand the theory, regulations, and the operational characteristics of the specific equipment that you are working on. In this case, we are addressing the specific operation of an X-radiation (X-ray) machine. Dealing with radiation requires much attention to detail and adherence to safety principles. You should know the concepts behind X-ray generation, the specific Federal and Air Force standards that govern radiation protections measures, and how to assess equipment operation and the environment for compliance. For instance, what if you make several high-load exposures in rapid succession during an X-ray unit calibration? Would you be able to determine whether the exposures exceed the maximum limits of your X-ray tube? The ability to answer these types of questions, along with many others, should spur your interest to be as informed as possible in the operation and safety hazards associated with X-ray equipment. You are not expected to learn all the complexities of X-ray production or of machine operation in this volume; however, a good working knowledge will surely assist you in your interaction with the X-ray department, personnel, and equipment.

In this volume, we will discuss various components and principles commonly encountered in typical and some special-purpose X-ray departments. The first section covers basic radioactivity, ionization, detection and measurement, and shielding.

1–1. Radiation Principles

When you work on various types of X-ray devices, you should be knowledgeable of certain fundamental concepts regarding radiation energy. Knowledge of the basic relationships of radioactivity helps you understand how X-rays form, behave, and affect the human body.

601. Radiation

You may recall from previous courses in physics, or the basic course, that radioactivity is the spontaneous disintegration of radioactive substances, in which the atomic nuclei undergo partial breakdown and give off penetrating radiation at the same time. Radioactivity is a natural property of all existing elements with atomic numbers (Z) above 83. Radioactivity can also be induced in all known periodic table elements.

A factor contributing to nuclear disintegration in radioactive elements is the instability of their atomic nuclei. For these elements to reach a more stable or less energetic state, they release excess energy in the form of alpha (α), beta (β), or gamma (γ) radiation. This nuclear instability may occur from the natural configuration of the atoms or be artificially created.

Natural radioactivity, in which nuclear disintegration is spontaneous, is exhibited in certain naturally occurring elements with unstable atomic nuclei, such as radium. Artificial radioactivity may be produced in elements with stable atomic nuclei by bombarding the nuclei with various subatomic particles, such as α and β particles, protons, neutrons, and γ rays. However, the relatively small size of

an atomic nucleus makes bombardment extremely difficult. To produce artificial radioactivity, the bombarding subatomic particle must deliver a direct hit to the atomic nucleus.

Now that we've seen the two sources of radiation, let's look at the three different types of radiation: α, β, and γ particles.

Alpha

The α particle is a stable combination of two protons and two neutrons, the same composition as the nucleus of the helium (He) atom, and holds a positive charge. When an element emits an α particle during nuclear disintegration, the particle is traveling at speeds of 9,000–20,000 miles per second, but is slowed down rapidly in its passage through matter.

The energy of an α particle is transferred over a very short range in matter. In air, alpha particles can travel about 5 centimeters (cm). As a result, α particles from external sources are essentially harmless. An α particle is unable to penetrate the outer layer of dead skin cells. If an alpha emitting substance is ingested through food or air however, it can cause serious cell damage internally.

Beta

A β particle is an extremely high-speed single electron (negative β) or positron (positive β), which ejects from the nucleus of a disintegrating atom. Electrons emitted from the nucleus probably result from spontaneous conversion of a neutron into a proton and an electron, as the atomic nucleus ordinarily does not contain free electrons. Except for its speed and origin, the electron is identical to the electrons that orbit about the nucleus of atoms.

In relation to the α particle, the β particle has a smaller mass and travels at a much higher speed, almost the speed of light, from 100,000–186,000 miles per second. Beta radiation can travel through air approximately 10–100 cm, and have a penetration ability of about 1 cm in tissue. Beta particles can be stopped by a thick piece of plastic, or even a stack of paper. Since the β particle can penetrate skin and damage living cells, the β emitter can create an external hazard.

Gamma

Both γ radiation and X-rays are bundles of pure energy, which we call photons. Gamma radiation results when protons and neutrons are rearranged in the nucleus of the unstable atom without emission of a particle. Because the components have different energies in different nuclear configurations, when rearranged, they release energy in the form of electromagnetic waves called γ rays. Sometimes γ radiation accompanies β and α emission.

Since the emission energy of γ radiation is usually measured in megaelectronvolts (MeV), the wavelengths are extremely short, and the γ radiation is the most highly penetrating. The emission from γ radiation affects only the involved nucleus, which is left with less energy. The velocity of γ radiation is 186,000 miles per second—the speed of light. Having no mass or charge, gamma radiation can travel much farther through the air than alpha or beta, losing (on average) half its energy for every 500 feet. To stop γ waves, a material must be thick or dense enough with a high atomic number, such as lead or depleted uranium.

X-ray

X-rays and γ radiation are identical, except for their origin and method of production. X-rays are the result of the conversion of the kinetic or potential energy of electrons into another form of energy, while γ radiation emits from radioactive nuclei. X-ray production stems from radioactive interactions of electrons with matter and collisional interactions of electrons with matter. These reactions happen within an X-ray tube. The fact that X-rays are man-made is the primary difference between X-rays and gamma rays. See figure 1–1 for a reference of radiation penetration mediums.

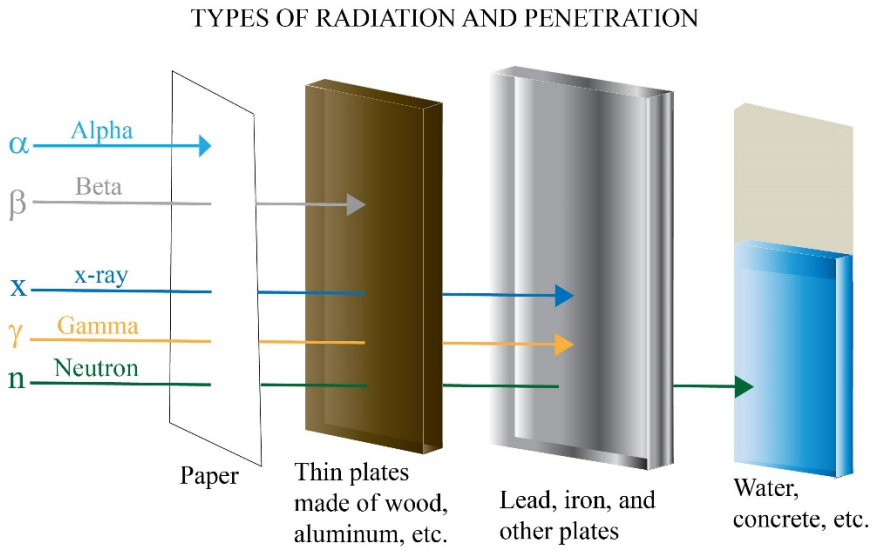

Figure 1–1. Radiation penetration mediums.

Bremsstrahlung

This term is used when a negatively charged electron approaches the positively charged nucleus, it may be deflected from its original direction by the attractive force of the nucleus. The change of direction causes a deceleration of the electron or a loss of some of its kinetic energy. The energy lost by the electron emits as an X-ray photon. This process is referred to as radioactive interaction of an electron with matter. The radiation produced by this type of interaction is called Bremsstrahlung (German for braking radiation), general radiation, or white radiation. Bremsstrahlung interactions may occur only when the incident electron interacts with the force field of the nucleus. The incident electron must have enough energy to pass through the orbital shells and approach the nucleus of the atom. Because atomic nuclei have a positive charge and the incident electron has a negative charge, there is a mutual attraction between them. When the incident electron gets close to the nucleus, the powerful nuclear force field is much too great for the electron to penetrate. Instead, the force field causes the incident electron to slow down (or brake) and then it diverts the electron's course. These bremsstrahlung photons and their energy is exactly the difference between the entering and exiting kinetic energy of the electron.

At longer distances, very little kinetic energy is lost, resulting in low-energy bremsstrahlung radiation. At shorter distances, more energy is lost, resulting in higher-energy bremsstrahlung radiation. The incident electron can also have a direct impact with the nucleus, resulting in the loss of all of the electron's kinetic energy. Because of the relatively small size of the nucleus, the chance of a direct impact is very low. Figure 1–2 shows the energy variations of bremsstrahlung interactions based on the speed and impact point of the incident electron.

Notice that when the incident electron strikes the nucleus directly, it expels all of its energy. The energy of the emitted photon would be equal to that of the electron. When the interaction emits a low energy photon however, the energy continues on to the next interaction. The energy of the resultant low-level energy photon depends upon:

- The original kinetic energy of the electron.
- How close the electron comes to the nucleus.
- The charge of the nucleus.

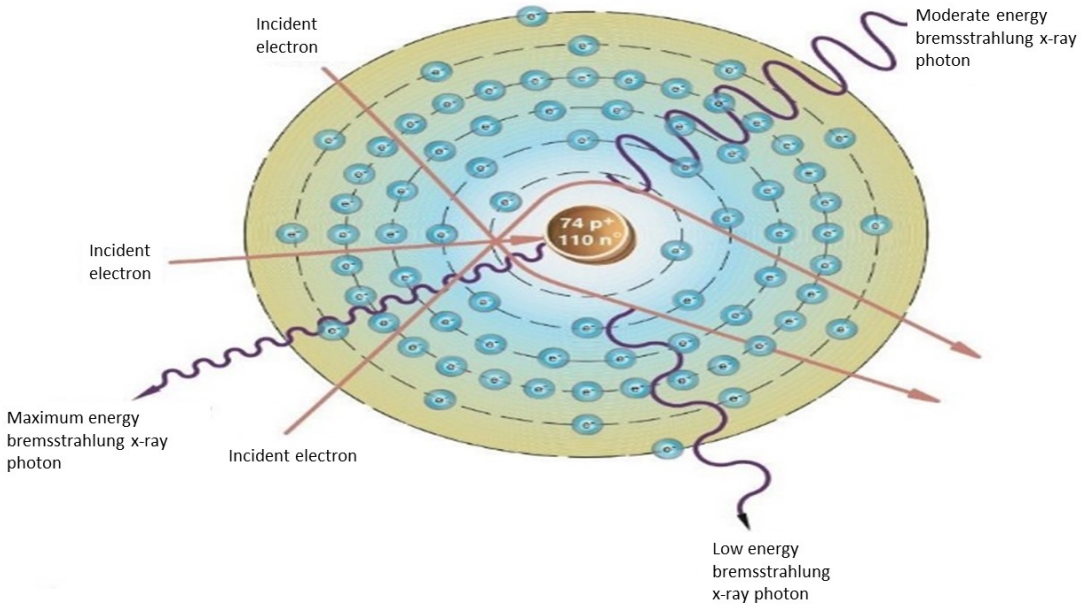

Figure 1–2. Bremsstrahlung interactions in a tungsten atom. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

In this type of interaction, the electron loses only a portion of its kinetic energy; it may have one or more interactions with other atoms before expending all of its energy. In this manner, it could produce several photons with various energies. Figure 1–3 illustrates how two electrons might interact with more than one atom to produce photons with a wide range of energies.

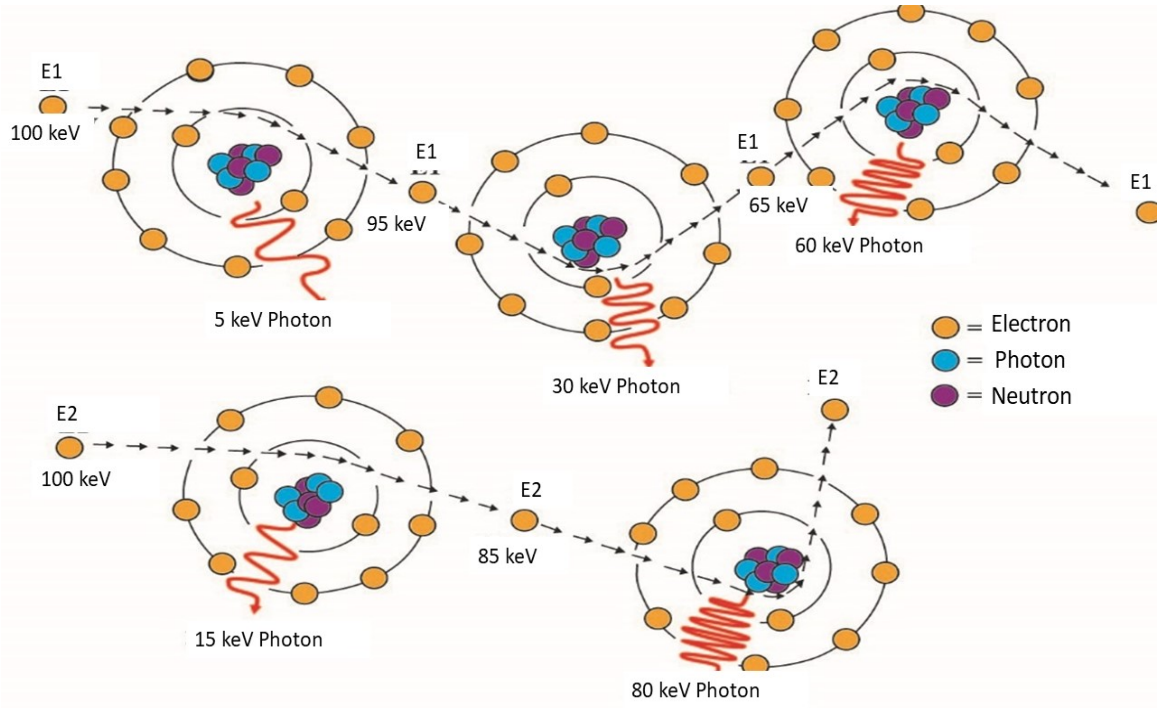

Figure 1–3. Bremsstrahlung interactions with many atoms.

Assume the two electrons, E1 and E2, entering the illustration from the left, each possess 100 kiloelectronvolts (keV) of kinetic energy. The first interaction of E1 results in it losing 5 keV of energy, which is given off as a 5–keV photon. E1 continues (its energy is now reduced to 95 keV) to interact with another atom, and this time loses a 30–keV photon. E1 now has 65 keV of energy remaining, and it interacts once again when it loses 60 keV of energy given off as a 60–keV photon. You may also notice that E2 only interacts with two atoms and loses 15 and 80 keV, respectively, given off as 15– and 80–keV photons. The actions of two electrons generated five photons with energies of 5, 15, 30, 60, and 80 keV. If millions of electrons interacted with millions of atoms in this manner, you can imagine the wide range of photon energies that this would generate.

Characteristic

The discussion to this point has demonstrated X-ray generation when electrons interact with the nucleus of an atom. X-rays also generate when electrons interact with tightly bound orbital electrons. This method is called collisional interaction of an electron with matter and produces characteristic radiation. The orbiting rings of electrons are called shells. For identification purposes, the electron orbital shells are given codes such as K, L, M, N, or O, to represent the relative binding energies of electrons from closest to the nucleus to farthest from the nucleus. The closer an electron is to the nucleus, the higher the binding energy. In a collisional interaction, an approaching electron has a collision with an inter-shell electron, such as an electron in the K shell of an atom of tungsten. The incident electron must have enough energy to knock an inter-shell electron from orbit, thereby ionizing the atom. The energy absorbed by the atom is equal to the binding energy of the shell. This leaves the atom in an excited state, or with an excess of energy, and an electron vacancy in a shell. Immediately after excitation, the atom returns to a normal state by emitting the energy it has absorbed in the form of X-ray photons. The atom returns to its normal state when another electron, such as from the L shell, fills the vacancy left by the ejection of the K shell electron (fig 1–4). Since the potential energy of an electron in the L shell is higher than that of an electron in the K shell, the L shell electron loses energy in the transition. The energy lost is about equal to the difference in the binding energies of the K and L shell and is given off in the form of an X-ray photon.

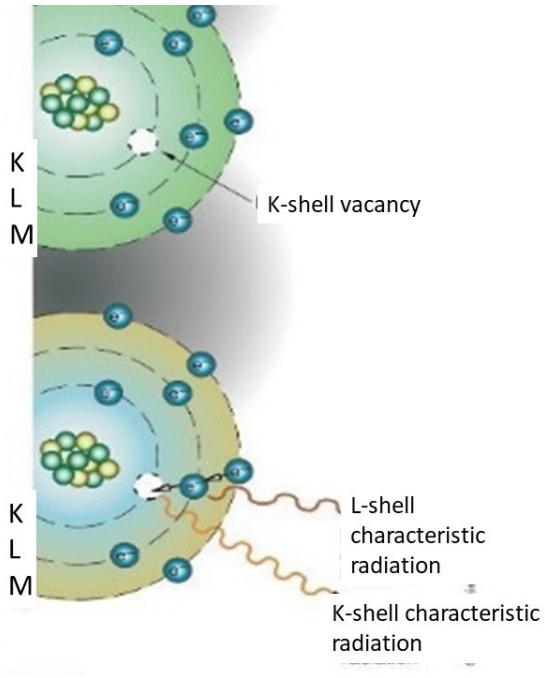

Figure 1–4. Shell vacancy action. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Although a photon has been emitted, the process is not yet completed because there is now a vacancy in the L shell, and the atom still has an excess of energy. This vacancy is also immediately filled by another electron, such as one from the M shell, and another photon is emitted equal to the energy of the difference of the transitioning electrons. This chain reaction continues with a photon given off for each electron transition, until the atom has no shell vacancies and again is in a normal state. A free electron fills the vacancy in the last shell. Figure 1–5 illustrates an atom of tungsten demonstrating photon release from the shells. The binding energies of tungsten's K, L, and M shells are 69.5, 12.1, and 2.8 keV, respectively. In figure 1–5, an incident electron collides with a K electron and ejects it from the vicinity of the atom. The electron transitions take place; as you can see two photons are emitted with energies of 59.0 keV, which is about the difference in the binding energies of the K and L shells, and 9.6 keV, which is the difference in the binding energies of the L and M shells.

When a K shell electron is ejected, the vacancy need not be filled from the L shell. It can also be filled from any shell with a higher potential energy level or even from outside the atom. In any case, the emitted photon is relative to the energy of the difference of the transition. The combined energies of the photons from a collisional interaction are equal to the binding energy of the shell from which the electron was ejected, because that amount of energy was initially absorbed by the atom. The maximum energy of a photon cannot be greater than the binding energy of the K shell of the atom under consideration or, in the case of tungsten, 69.5 keV.

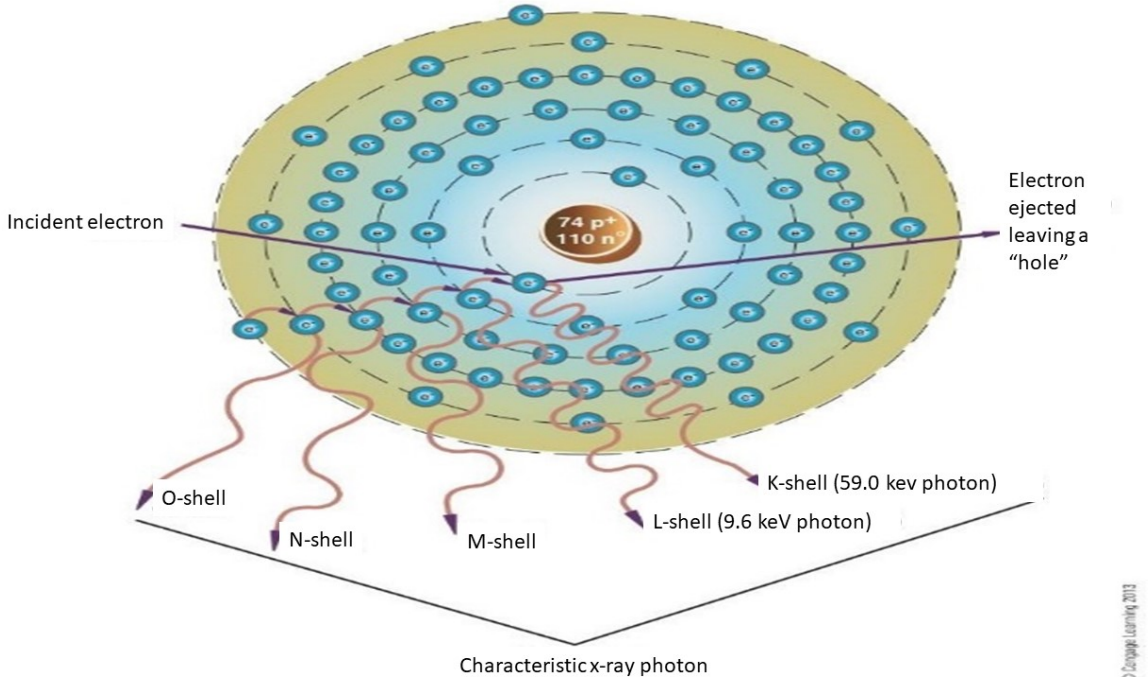

Figure 1–5. Collision of K shell electron and resultant photon action. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

The radiation produced in this manner of a collisional interaction is called characteristic radiation, because its energy is characteristic of the shell of the atom from which it came. For instance, the binding energy of the K shell in copper is 9 keV; therefore, the maximum energy of characteristic radiation that could be generated in copper would be 9 keV, an amount not usable in radiology. Tungsten, however, as stated before, can generate characteristic radiation with a maximum energy level of 69.5 keV, some of which can be useful in radiology. This is why we use tungsten as a target material within and X-ray tube, which we will cover later in this section.

Interactions of photons with matter

The definition of the term "interaction" is quite simple: one force or body having a measurable effect on another force or body. You can see daily evidence of interaction in a bowling alley, watching a sailboat at the lake, or on the job in the many uses of electrical transformers. The level of interaction you must fully comprehend at this time is what takes place when a beam of X-ray photons passes through anything having mass and occupying space. An X-ray beam consisting of photons of pure energy (Q) transfers its energy to the matter through which it is passing, whether it be air, an X-ray film or detector, or living tissues. In many cases, the interaction is not immediately evident without complicated devices to detect these events. Some, if not all, of the X-ray energy seems to disappear in certain materials. The term that best describes this phenomenon is absorption. Absorption is the process by which an X-ray photon transfers its inherent energy to the medium through which it is passing. Let's examine the process that causes these energy transfers.

Ionization

A process known as ionization brings about all the changes, and can be defined as any process which results in the removal or addition of an orbital electron from or to an atom or molecule, thereby leaving the atom or molecule with an overall positive or negative charge. Ionization can occur when an electron is struck by a photon, at which time an energy transfer takes place. Although it is technically possible for this energy transfer to take place in the nucleus, the chances of a photon reaching that vicinity are extremely remote. After an ionizing event occurs, the remaining particles are called a pair of ions, in the case of electron removal. The parent atom (minus an electron) has an overall positive charge and is known as a positive ion. The ejected electron has a negative charge and is known as a negative ion. Figure 1–6 illustrates the process of ionization. Radiation is measured by the number of ion pairs it causes and the standard unit of measure is the roentgen (R). The amount of radiation that causes approximately 2 billion pairs of ions to be formed in 1 cubic centimeter (cc) of air (at normal pressure/temperature) is considered 1 R. The two types of ionization that most generally occur in the diagnostic radiation energy range are the photoelectric effect and Compton effect. Let's take a quick look at both of these.

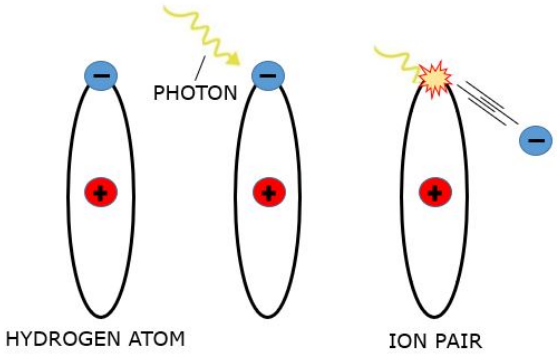

Figure 1–6. Ionization.

Photoelectric effect

This effect (fig. 1–7) is an all-or-nothing energy exchange where the photon imparts all of its energy to the electron and simply vanishes. The ejected electron, called a photoelectron, departs with all the inherent energy of the photon and can cause secondary ionization due to its increased kinetic energy. In the meantime, as the excited atom returns to the normal state, it quickly attracts another electron to fill the vacant "hole" and radiation is emitted. The energy of the radiation and the sequence of events that causes the radiation is much the same as the replenishment of an electron shell vacancy created by the ejection of an electron by an electron collision, which we discussed earlier. The photoelectric effect normally occurs with photon energies up to 100 keV.

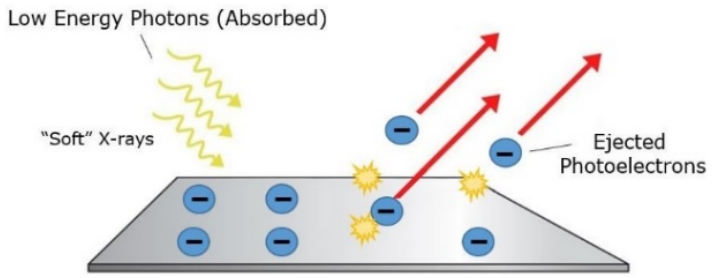

Figure 1–7. Photoelectric effect.

Compton Effect

The Compton Effect, also referred to as modified or incoherent scattering, is the result of a partial transfer of energy from an X-ray photon to an orbital electron (fig. 1–8). In this case, the photon strikes a glancing blow to the electron and ejects it from orbit. Although considerably weakened in energy (longer wavelength), the photon continues on its path. While the now "soft" photon eventually disappears via a final photoelectric effect, the ejected electron can, as in the previous case, continue on its path to cause another or secondary ionization of a nearby atom. In contrast to the photoelectric effect, the Compton Effect is predominant with highly energetic X-rays in the 100 keV–10 MeV energy range.

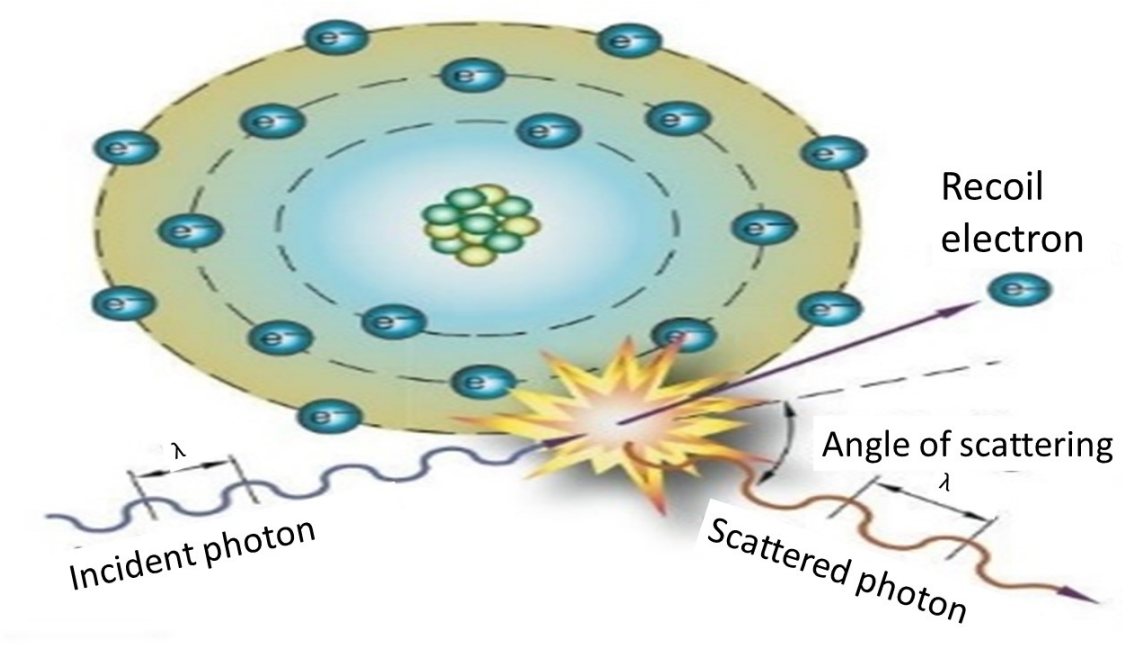

Figure 1–8. Compton Effect. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

602. Detection, measurement, and protection

Since you will be working directly in the presence of radiation, it is important that we monitor your exposure to the potentially harmful rays. Because your five senses cannot detect X-rays, we must employ indirect methods of detection, measurement, and protection.

Detection

Although the ionizing capability of X-ray is harmful and in some cases deadly to living tissue, it is this same capability that allows detection through two means—chemical and electrical change. Let's briefly examine each.

Chemical change

One example of a chemical change is the process of ionizing the silver bromide crystals in the emulsion of a film badge. The crystals exposed to ionization go through a chemical change and reduce to black metallic silver when exposed to developing chemicals. The non-ionized crystals remain chemically inert to the developer and are removed during the fixing and clearing process. Thus, the degree of darkening on a film badge, when processed under controlled conditions, determines the quantity or total dose of radiation received by the wearer.

Electrical change

The electrical changes brought about by an ionization process are also useful in detecting the presence of X-ray. It is possible to detect and measure radiation from the number of freed electrons, since an X-ray beam provides a large number of such electrons where there were essentially none before.

Measurement

Before we go any further, let's look at the five units used to measure radiation. For clarity, you may wish to refer to figure 1–9, which uses a hypothetical situation to relate four of the units and illustrates the way they would be used.

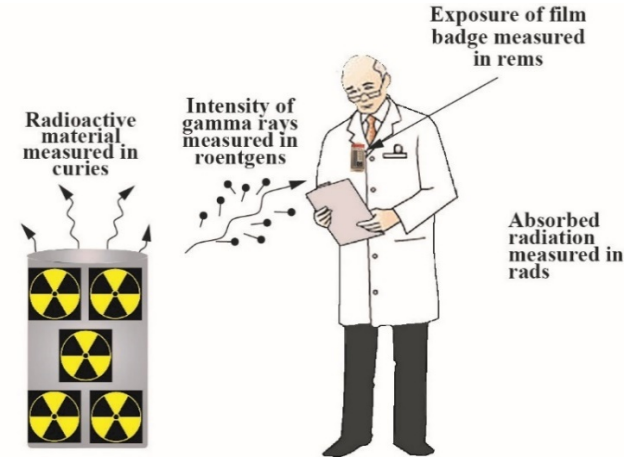

Figure 1–9. Units of radiation measurement.

Units of measure

The DOD generally uses a conventional unit system for radiology units of measure, but you might also see values or doses expressed in International System of Units (SI), considering some of the leading radiology equipment used in the Air Force comes from the international community. Since most of the testing you perform is measured in conventional units, the remainder of this lesson will focus on improving your measurement proficiency, except for a small section where we will touch on

SI equivalents for reference. The following are the five units of measure for radiation with which you must be familiar:

1. R—is the unit of measurement for radiation exposure or intensity. The official definition, however, is in terms of electric charge per unit mass of air. The charge refers to the electrons liberated by ionization. The R was first defined as a unit of radiation quantity in the late 1920s. Radiation-measuring instruments are usually calibrated in Rs. The output of X-ray machines is specified in Rs, or sometimes milliroentgens (mR). The R applies only for X-rays and γ rays, and defines the quantity required to produce a given amount of ionization (charge) in a unit mass of air. The conventional unit for exposure is the roentgen, and the SI unit is the SI unit is the coulomb per kilogram (C/kg) of air, expressed as $R = 2.58\,x\,10^{-4}\ C/kg.$
2. Radiation absorbed dose (rad)—Biological effects from X-rays are usually related to the rad and, therefore, we most often use rad when describing the radiation quantity received by a

patient. One rad is equal to 100 ergs per gram, where the erg is a unit of energy and the gram is a unit of mass. The SI unit is the gray (Gy), which is equivalent to the absorption of 1 Joule (J) of radiation energy per kg of tissue. This is not restricted to air and can be measured in other absorbing materials.

3. Radiation equivalent man (rem)—Personnel monitoring devices, such as film badges, are analyzed in terms of rems. The rem is the unit of dose equivalent (DE) or occupational exposure. It expresses the quantity of radiation received by radiation workers. Some types of radiation produce more damage than X-rays and the rem accounts for these differences in biological effectiveness. This is particularly important to persons working near nuclear reactors or particle accelerators. The SI unit is the sievert (Sv), where 1 Sv is equivalent to 100 rem.
4. Curie (Ci)—The Ci is a unit of radioactivity related to the three preceding units of radiation. This unit expresses the quantity of radioactive material and not the radiation emitted by that material. One Ci is that quantity of material in which 3.7×10^{10} atoms disintegrate per second (dps). One curie is a very large amount of radioactive material, so the millicurie (mCi) and microcurie (μCi) are common quantities that you might see (i.e., a typical nuclear medicine procedure uses activities from 0.1 to 30 mCi). The SI unit for activity is the becquerel (Bq), with the Bq representing 1 dps. See the table below for conventional to SI unit equivalents and conversions.
5. Electron volt—The energy of an X-ray is measured in electron volts (eV) or, more often, keV. An electron accelerated by an electric potential of one volt acquires energy to one eV. Most X-rays used in diagnostic radiology have energy up to 150 keV, whereas those used in radiotherapy measure in MeV. Other radiologically important energies (i.e., electron and nuclear binding energies, and mass energy equivalence) are also expressed in keV. Since diagnostic radiology is concerned primarily with X-rays, you may consider 1R equal to 1 rad equal to 1 rem. Your understanding of these terms will increase as we consider them individually in various applications.

Exposure	***Conventional Unit***	***SI Unit***	***Conversions Conv to SI***	***Conversions SI to Conv***
Exposure	roentgen (R)	coulomb/kg of air (C/kg)	$1\ R = 2.58 \times 10^{-4}$ C/kg	1 C/kg = 3876 R
Absorbed Dose	rad	gray (Gy)	1 rad = 0.01 Gy	1 Gy = 100 rad
Dose equivalent	rem	sievert (Sv)	1 rem = 0.01 Sv	1 Sv = 100 rem
Activity	curie (Ci)	becquerel (Bq)	$1\ Ci = 3.7 \times 10^{10}$ Bq	$1\ Bq = 2.7027 \times 10^{-11}$ Ci

Ion chamber

This chamber (fig. 1–10) is an example of an instrument that measures radiation dose rate. When a difference in potential is applied across the chamber and causes movement of the free electrons, the rate of current flow will reflect the amount of radiation striking the chamber. The instrument can be calibrated by using a known quantity of radiation. Figure 1–10, A shows no reading on the meter because there is no ionization (no electrons being freed); consequently, there is no electron flow. In figure 1–10, B, ionization is taking place (electrons are being freed) and the application of a potential difference is causing electron flow, which results in a reading on the meter. The meter is calibrated in Rs.

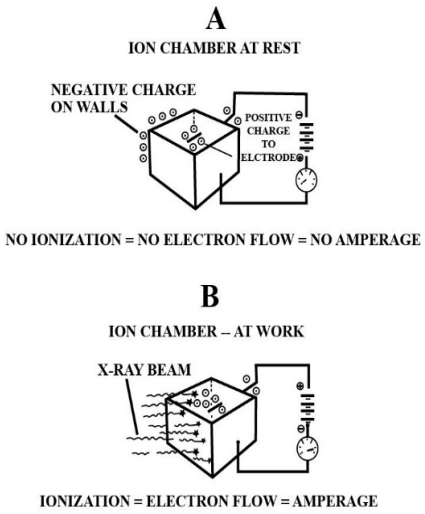

Figure 1–10. Ion chamber.

Personnel dosimetry program

If your duty assignment requires you to work on or around X-ray equipment, you will be monitored on the personnel dosimetry program. Since you will likely use personnel monitoring devices throughout your career, it is helpful to know something about this very important method of detecting radiation exposure. There are several other methods of detecting radiation; however, we will limit our discussion at this time to film badges, thermoluminescence dosimeters (TLD), and optically stimulated luminescence (OSL) dosimeters.

Film badge

The film badge is an example of a total dose detector. It is the device most commonly used to detect and measure occupational radiation exposure. The bioenvironmental engineering section in your medical treatment facility (MTF) is responsible for the personnel dosimetry program, and can advise you on the proper wearing and maintenance of film badges, should you need more specific information.

A film badge consists of a film packet in a film holder usually constructed of plastic or metal. In the film packet are two pieces of photographic film. One has a double emulsion, and one has emulsion on one side only. The single emulsion is of prime importance when loading the film holder. The film may contain only an X-ray sensitive emulsion, or it may be sensitive to X-rays, γ rays, and β particles. Special films are also available to detect and measure neutrons.

The film responds to radiation exactly as radiographic film does—ionization occurs within the emulsion layer, bringing about a chemical sensitivity to the developer solution in the ionized silver bromide crystals (as explained earlier in this lesson). Processing is carried out under extremely controlled conditions by a central agency to ensure accurate and reliable results.

Thermoluminescence dosimeters

The TLD badge contains lithium fluoride in crystalline form, either as powder or more often as a small chip approximately 3-millimeter (mm) square and 1-mm thick. When exposed to radiation, the TLD absorbs energy and stores it in the form of excited electrons in the crystalline lattice. When heated, these excited electrons fall back to their normal orbital state with the emission of visible light. The intensity of the visible light is measured with a photomultiplier tube and is proportional to the radiation dose received by the crystal.

The TLD has several advantages over the film badge. It is more sensitive and accurate than film badge monitors. Properly calibrated TLD monitors can measure exposure as low as 5 mR. The TLD monitor does not suffer from loss of information following exposure to excessive heat or high

humidity. Consequently, they can be worn for intervals of up to 3 months at a time. The primary disadvantage of TLD monitoring is cost. The price of the typical TLD monitoring service is perhaps twice that of film badge monitoring, but by reducing the frequency of monitoring to quarterly (as opposed to monthly), the cost becomes similar. The TLD has the additional advantage of simplicity. Monitoring with TLD is improving constantly and is slowly replacing monitoring with film badges.

Optically stimulated luminescence

The OSL and thermoluminescence are very similar—the only difference is the manner in which the electrons are freed from their traps. The absorption of energy from an ionizing radiation source by an insulating or semi-conducting material causes the excitation of free electrons and free holes and the subsequent trapping of these electronic species at defects (trapping states) within the material. OSLs make use of electrons trapped between the valence and conduction bands in the crystalline structure of certain minerals (most commonly quartz and feldspar). The greater the radiation energy absorbed (dose), the greater the number of trapped electrons. When it is time to assess the dose, the trapped electrons are freed by exposing the dosimeter to light (ultraviolet, visible, or infrared).

The OSL dosimeter (fig. 1–11) provides a greater degree of sensitivity by giving an accurate reading as low as 1 bulk net magnetization radiation equivalent man (mrem) for X-ray and gamma ray photons with energies ranging from 5 keV to greater than 40 MeV. The OSL dosimeter's maximum equivalent dose measurement for X-ray and gamma ray photons is 1000 rem. For beta particles with energies from 150 keV to in excess of 10 MeV, dose measurement ranges from 10 mrem to 1000 rem. In diagnostic imaging, the increased sensitivity of the OSL dosimeter makes it ideal for monitoring employees working in low-radiation environments and for pregnant workers. The advantage of using OSL dosimeters, unlike a TLD, is that they can be reread multiple times. Depending on the illumination conditions, there will be a decrease of signal of less than one percent in a second reading. OSL dosimeters can be read at room temperature. This simplifies the design of the equipment. The use of OSL powders deposited in thin layers creates a two-dimensional (2D) detector with imaging capabilities much like film. Also, used OSL badges are often archived for several years, without fading, except in extreme temperatures. OSL dosimeters, however are also more expensive to use than TLDs.

Figure 1–11. Optically stimulated luminescence (OSL) dosimeter.

Wear

Generally, the personnel monitor is worn on an area of the body expected to receive the highest exposure, such as the chest. The monitor should not be carried in a pocket or behind any obstruction (i.e., coins, combs, or cigarette packages) as they tend to absorb radiation and reduce the ultimate density reading of the film. You should avoid receiving a direct exposure of X-rays while wearing the monitor; although this may sound like an obvious assumption, maintenance accidents of this type have occurred. Direct exposures make it difficult to obtain a true density reading. Film badges should also be protected against direct sunlight to prevent thermal sensitization and possible light leaks in the wrapping paper.

When not in use, monitors are stored in a radiation-free area, along with a control dosimeter badge. The purpose of the control badge is to permit the laboratory responsible for processing and evaluating the personnel-monitoring device to take into account such factors as background radiation and temperature variations that would otherwise be recorded as an occupational exposure.

You should know that when an overexposure is indicated, an investigation is conducted to see if the exposure was indeed accidental, or if it was the result of a deliberate act or carelessness. Some of the most common causes of overexposure are:

- Deliberate exposure of the monitoring device.
- Improper storage of the monitoring device, such as in your toolbox placed on an X-ray table.
- Failure of the individual to utilize protective shielding.
- Improper work habits.
- Inadequate or defective radiation shielding.
- Unintentional wear of the film badge while receiving diagnostic or therapeutic X-rays.

Improper use of monitoring devices results in misleading reports, and a waste of time and money in unnecessary investigations. Personnel monitoring programs are designed to provide you with a means for detecting accidental exposure to radiation so you can be provided with medical treatment if necessary. It behooves you to wear the monitoring device when appropriate to do so and take required measures to ensure the success of your personnel dosimetry program.

Protection from radiation

Even though a patient knowingly receives exposure to radiation for the purposes of diagnostic advice, it does not mean they want to be exposed to more radiation than needed. Therefore, protection is not only for the operator, but also for the patient.

The most effective way to reduce an individual's exposure to ionizing radiation is by following the three basic principles: time, distance, and shielding. We sometimes refer to these three principles as the cardinal rules of radiation protection. Let's examine each of these ways in which the effects of radiation can be lessened and the characteristics of different laws that control the intensity of X-ray energies.

Time

It should be obvious that any X-ray exposure, whether to a patient or operator, should be as short as possible. The time of exposure on a per-image basis is very short (typically much less than 1 second) during radiographic procedures. During fluoroscopic procedures, which are used when dynamic information is required, the X-ray source may be on (usually intermittently) for several minutes to as long as an hour or more. To minimize exposure to radiation, individuals should reduce the amount of time they spend in the vicinity of an operable radiation source. There are two terms you should be familiar with when dealing with time and radiation together—dose rate and total dose.

Dose rate is the amount of radiation received per unit of time, such as 50 roentgen (R) per hour. A dose rate of 100 R per minute for 4 minutes would result in a total dose of 400 R.

Total dose is the total amount of radiation received, such as 50 R, with no indication of the time during which it was received. It may have been one minute or spread out over one year. As the dose gets larger, this time element becomes extremely critical and, in the case of a careless radiologic technician, perhaps even deadly. Therefore, it is necessary to be able to discuss radiation doses in terms of the rate at which they are received.

Distance

Distance is a factor for the operator and patient. The operator should be as far away from the direct beam as possible. The object is to ensure the X-ray is deflected twice before striking a bystander or operator. Distance from the X-ray source is a highly effective method of reducing the intensity of an

X-ray beam. This can be expressed in the inverse square law, which states that the X-ray or γ radiation intensity from a point source varies inversely with the square of the distance from the source. Expressed mathematically, the inverse square is:

$$\frac{I_1}{I_2} = \frac{(D_2)^2}{(D_1)^2}$$

In this equation, I_1 equals the intensity at the original distance, I_2 equals the intensity at the new distance, D_1 equals the original distance, and D_2 equals the new distance.

Suppose the intensity of an X-ray beam was 100 R per minute at a distance of 2 feet from the X-ray tube. What would the new intensity be at a distance of 4 feet? By substituting the data from this problem into the formula, the new intensity can be found: I_2 = 25 R per minute.

By doubling the distance from the X-ray tube, the intensity of the beam can be reduced to one-fourth its original value, or from 100 R per minute to 25 R per minute. This, as you can see, is an impressive reduction in intensity, which can be used to the advantage of the radiologist in keeping their radiation exposure to a minimum. Remember, this formula is applicable only to radiation from a point source, such as the target in an X-ray tube.

Shielding

Shielding is the intentional use of materials of various densities to limit, control, or modify the electromagnetic energy output—in our case, the output of an X-ray tube. To understand the effects of shielding and, therefore, be able to take advantage of it as a radiation protection tool, it is necessary to review certain facts about the interaction of X-ray photons with matter. Photon energy may be lost in many different ways; however, as mentioned earlier, in the wavelengths associated with medical X-rays, the photoelectric and Compton effects are the two predominant ways. As X-ray photons travel through an absorber, the amount of reduction, or attenuation, is determined by three important factors:

1. Photon energy factor—A factor that influences photon absorption or beam attenuation is the energy level of the photons. The higher the photon energy, the more penetrating the X-ray beam, regardless of the material used for shielding. As kilovolt peak (kVp) is increased, photon energy is increased, thereby causing more penetration.
2. Absorber density—A characteristic of an absorber, which determines its ability to absorb radiation, is atomic density. The more closely packed the atoms, the greater the probability for photon/electron interaction to take place. Because of its high absorption density, lead is an excellent shielding material, and is widely used in and around radiology departments.
3. Absorber thickness—The last factor we will mention that influences attenuation is the thickness of the absorbing material. If photon energy and absorber density remain constant, further attenuation can be accomplished by simply adding more absorber material. Simply put, if 6 inches of concrete is good, 12 inches is better.

Self-Test Questions

After you complete these questions, you may check your answers at the end of the unit.

601. Radiation

1. Match the characteristics in column A with the appropriate type of radiation in column B. Each type of radiation in column B may be used once or not at all.

Column A	*Column B*
____ (1) The spontaneous disintegration of radioactive substances.	a. β.
____ (2) A stable combination of two protons and two neutrons, the same composition as the nucleus of the helium atom, and holds a positive charge.	b. γ.
____ (3) This type of radiation results when protons and neutrons are rearranged in the nucleus of the unstable atom without emission of a particle.	c. X-ray.
____ (4) This type of radiation results in the conversion of kinetic or potential energy of electrons into another form of energy.	d. α.
____ (5) Has a penetrating ability of about 1 cm in tissue and 10–100 cm in air, and travels at 100,000 to 186,000 miles per second.	e. Radioactivity.

2. What is the name given to the type of radiation interaction known to deflect an electron from its original direction by the attractive force of the nucleus?

3. What is another name given to the production of X-rays by collisional interaction of electrons?

4. Define ionization.

5. What effect is known as an all-or-nothing energy exchange?

6. What form of diagnostic radiation energy effect is referred to as a modified or incoherent scattering, and results from a partial transfer of energy of an X-ray photon to an orbital electron?

602. Detection, measurement, and protection

1. What two methods are used in the detection of X-rays?

2. What does “rad” stand for?

3. When applying a difference in potential across the ion chamber, what does the rate of current flow reflect?

4. What is the difference between TLDs and OSLs?

5. State the main reason for shielding in an X-ray department.

6. Why is lead used as an absorber for X-rays?

1–2. Principles of X-radiation Generation

Although X-ray equipment is complex, we sometimes overlook the fact that an X-ray machine is not as complex as it may seem, if you take each component or system and break it down to its most basic operation and purpose.

This section will familiarize you with the basic circuit functions and simple schematic diagrams of an X-ray system to include principles and characteristics of X-ray tubes. We also discuss specialized accessories that aid the radiologist in the diagnosis of patients.

603. X-ray control circuit components

At this point in your career, you should have a fair working knowledge of a basic X-ray system from what you learn in the BMET basic course. This lesson will simply refresh your memory and include some new information to add to what you already know.

The control console is the apparatus that allows the technologist to adjust the X-ray tube current and voltage in order to produce a useful X-ray beam of proper intensity and penetrability for producing a good quality radiograph. The console usually provides for adjustment of line compensation, kVp control, milliampere (mA) control, and exposure time. Load compensation and space charge compensation also adjust as a result of changes to kVp and mA.

We'll discuss these subjects at this time to enable you to become more proficient in the analysis, troubleshooting, and calibration of the control circuitry. Remember, regardless of which manufacturer's unit you have, these basic concepts will be the same.

Input power

Most X-ray systems operate on either single-phase (1ϕ) or three-phase (3ϕ) power, depending on their design. Figure 1–12 illustrates the basic configuration of the two different types. If operated on 1ϕ power (fig. 1–12, A), both input lines are used to apply power to the control. If operated on 3ϕ power (fig. 1–12, B), two legs of the input are used to apply power to the control. The third leg in a 3ϕ system is used during X-ray exposure to produce 3ϕ power to the X-ray tube.

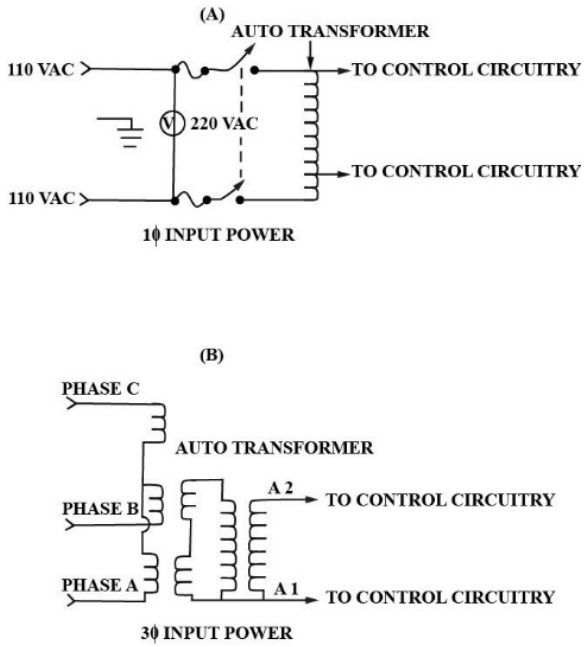

Figure 1–12. Single-phase and three-phase input power circuitry.

Autotransformers and line compensation

Since most practical X-ray units operate from standard service lines, the first circuit energized in any X-ray device is the one that connects to the service lines—this circuit will energize all other X-ray control circuitry. Before considering this circuit, let's consider some of the characteristics of the incoming line.

The voltage output of any incoming line varies from day to day, depending on the demands placed on the power source. At times, when there is a great deal of power being consumed by all sources, the voltage at the source decreases. At other times, when the demand is light, the source voltage is higher. This variation, which follows the voltage demand cycle of your facility, is commonly called line variation. Since your X-ray unit has circuits that require a specified amount of voltage for proper operation, some means to adjust and compensate for this line variation must be internally available. If no attempt were made to correct for line variation, the amount of penetrating ability of the X-rays produced would also vary due to the changing voltage supplied to the circuit. Therefore, to maintain consistent results from your X-ray unit, a line to the autotransformer circuit must have some adjustment to compensate for different line voltages, known as line compensation.

Let's begin to build our basic X-ray control circuit by looking at figure 1–13, A. For the control circuits to be supplied with the proper voltages, the volts per turn of the transformer must be correct as specified by the manufacturer. The technician needs some indicator to assure the correct adjustment of the line to the autotransformer circuit. Installing a voltmeter across a certain number of turns on the transformer can accomplish this. When the voltage across the voltmeter reaches a predetermined value, the volts per turn of the autotransformer circuit is aligned and the voltage output of the transformer is correct for the control's operation.

The voltmeter and line compensator enable the technician to adjust the volts per turn of the autotransformer. The voltmeter serves as an indicator, which usually has an indicator needle or red acceptable limit marker on the face of the meter. By turning the line compensator knob, the technician can adjust the circuit to compensate for changing source fluctuations.

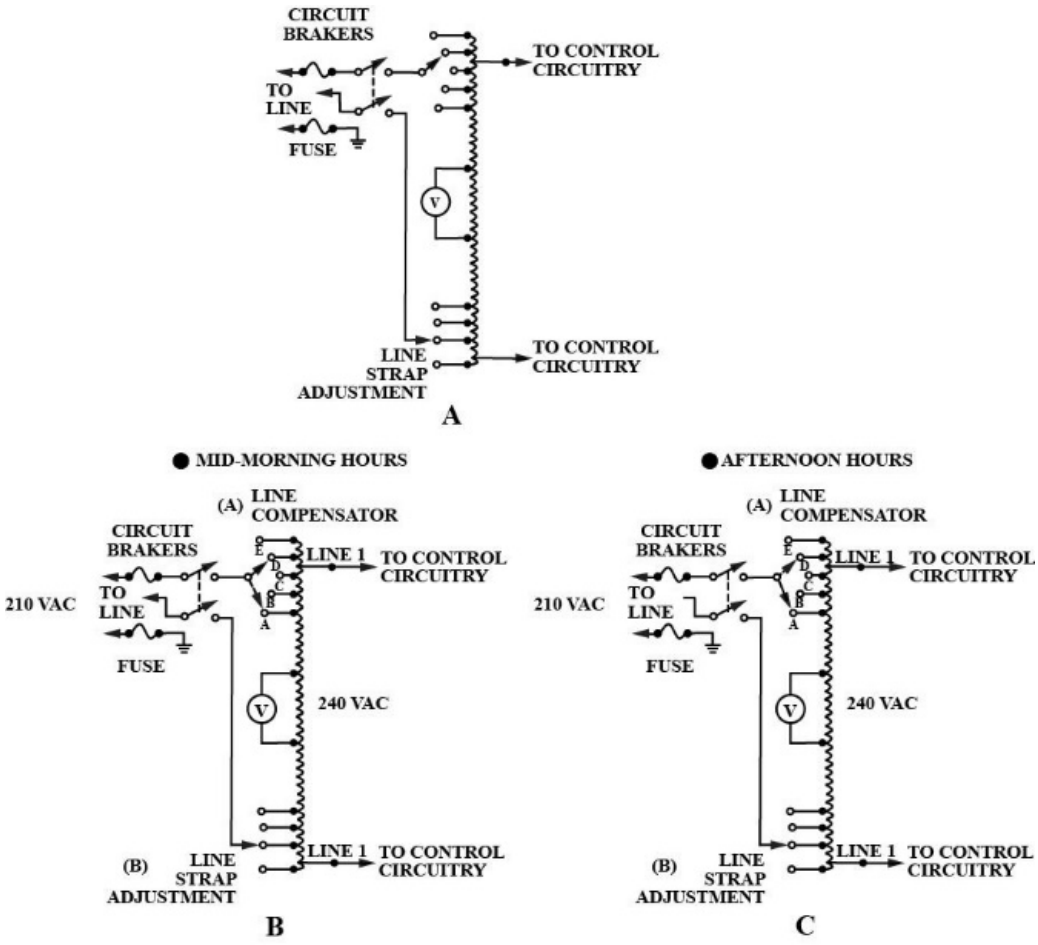

Figure 1–13. Autotransformer and line compensation circuit.

The line strap is a major adjustment accomplished by the company representative or you during the initial installation. If the manufacturer does not supply directions for connecting the line strap, start by adjusting the line compensator to the midpoint position. Connect the line strap to a midpoint position and see if you can get a red line reading on the voltmeter; if not, adjust the line strap in the direction necessary for the meter to red line.

CAUTION: During all adjustments, ensure the main breaker is in the OFF and LOCKED position.

The circuit contains the main circuit breakers or main line switch, which disconnects the service power line from the unit. Also included is a ground wire that extends from a good earth ground to the frame of the unit—this protects the operator from electrical shock should one of the wires of the incoming line short to the casing of the unit. Each side of the incoming line will also have a fuse to protect the components in the unit from damage in case of overload. A circuit breaker may accomplish overload protection in place of using fuses. The ground wire directly connects without fusing.

In most X-ray units you will encounter, line compensation is automatic and, therefore, will not require adjustment by the radiographer; however, some older units may still require manual adjustment. For this reason, as well as for you to understand how line compensation works, we will cover it in the following paragraphs.

Follow along with figure 1–13 for the following example. For our purposes, let's say the control circuitry operates on 240 volts alternating current (VAC). Due to heavy usage of electricity during

midmorning hours, the line voltage supplies only 210 VAC to the autotransformer. To apply 240 volts (V) to the control circuitry, we would move the line strap compensator switch (A) from position "D" to position "A." This would apply 210 VAC between the line strap (B) and position "A" of the switch. The windings between the line strap (B) and point "A" would induce a voltage into the windings between point "A" and the point between line 1 and line 2. This would give us the desired voltage (240 VAC) necessary for the control circuitry. Later in the afternoon, you discover the line voltage has increased to 250 VAC due to a low usage of electricity during the lunch hour. You would move the line compensator switch (A) back to the "D" position (fig. 1–13, C), which applies 250 VAC between the line strap (B) and position "D" of the switch. This voltage is too high, so your final adjustment would be to set the line compensator to position "C" to ensure the approximate operating voltage of 240 VAC throughout the duty day.

Various manufacturers may use different circuits to accomplish line compensation, but all manufacturers' circuits mirror the concepts of line compensation presented in the preceding paragraphs.

Kilovolt peak control

The kVp control, often referred to as the high-tension circuit, uses a step-up transformer to develop the thousands of volts needed to produce X-rays. Kilovolt peak control circuitry is similar to line compensation circuitry because, in most units you will encounter, it is automatic; however, you may see some older units that have manual adjustments, which allow the radiographer to manually adjust the kVp.

The primary of the high-tension (PHT) circuit includes electrical devices and conductors from the power source to the primary winding (fig. 1–14) of the high-tension transformer and back to the power source—that source being the autotransformer. The electrical devices incorporated within the basic PHT circuit of any X-ray unit include a means of adjusting the potential applied to the X-ray tube, a provision for energizing and terminating the PHT circuit, and some means of indicating the potential produced across the X-ray tube.

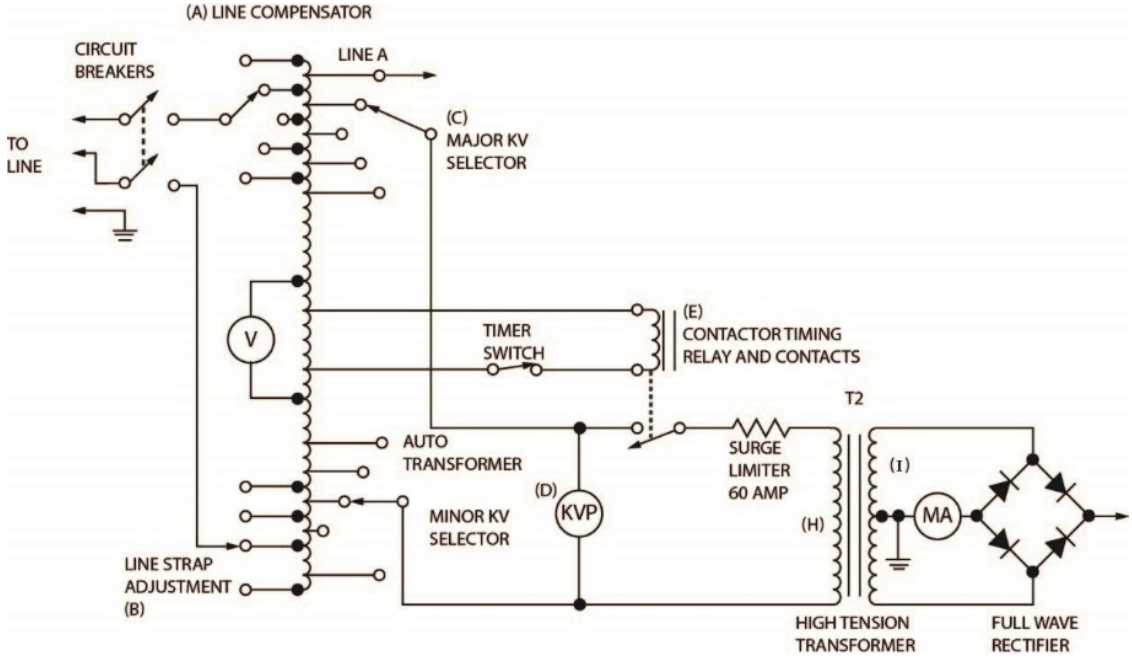

Figure 1–14. High-tension circuit, kVp.

Figure 1–14 shows the PHT circuit begins at the autotransformer with the major kilovolt (kV) selector (C) to the primary winding through the normally open (NO) contact points labeled (E), including the contactor coil. The PHT circuit then continues through to the minor kV selector at point (D) to complete the circuit back to the autotransformer.

On older units with manual adjustments, the major and minor kV selectors provide a means of increasing or decreasing the kVp across the X-ray tube by varying the potential across the primary winding of the high-tension transformer. The major kVp selector usually adjusts in increments of 10kVp, and the minor kVp selector adjusts in increments of 1–2kVp. Therefore, if the major kVp selector had 10 steps and the minor had 10 steps as well, it would yield a possibility of 100 different voltage settings.

The purpose of the contactor coil and points (E) are to energize and terminate the PHT circuit. As long as these points are open, there is no high voltage across the X-ray tube, and there is no X-ray from the tube. The contact points, as mentioned earlier, are NO and the coil is located in the operating circuit.

The secondary of the high-tension (SHT) circuit consists of the secondary winding (I) of the high-tension transformer, the X-ray rectifier circuit, and milliammeter (mA meter). The X-ray tube is also included, but is not shown in figure 1–14; it will be discussed later.

The secondary winding of the high-tension transformer is grounded at the center of the secondary windings, and the mA meter is connected at this point. The high-tension insulated cables that connect between the secondary winding and the X-ray tube are designed for use with center-grounded secondary windings. Without the center tap ground, the insulation requirements of the high-tension cable would double, and the cable would then be too cumbersome and costly. The mA meter is located on the control panel for convenience; consequently, for safety, it too must be grounded. Since the mA meter is a low-resistance device, the voltage drop across it to the center tap ground is only a few volts; however, if the meter were not grounded, it would require high-voltage insulation.

The mA control and filament circuit

We measure the X-ray tube current (also considered the number of electrons crossing from cathode to anode) in mA. The temperature of the filament determines the quantity of electrons emitted by the filament, and the filament current (mA setting), in turn, controls the temperature. The voltage applied to the primary of the filament transformer determines the current through the X-ray tube filaments.

The X-ray tube filament, as used in medical radiography, must heat up until it becomes incandescent; which requires a relatively high current to produce this temperature. The autotransformer is the power source for the primary of a step-down filament transformer to produce the required high current at a low voltage. The filament transformer also isolates the high voltage, which appears across the X-ray tube from the autotransformer circuit.

A separate circuit called the filament circuit controls X-ray tube current (fig. 1–15, J). Taps of the autotransformer provide voltage for the filament circuit. This voltage drops across precision resistors to a value corresponding to the mA station desired. X-ray tube current normally is not continuously variable; usually currents of 25, 50, 100, 150, 200, 300, and 500 large and small, and higher are provided. Large and small refers to the filament size used for a particular operator technique. The mA station precision resistors are usually wire-wound variable resistors, which you can calibrate for the individual mA stations.

A large or small filament select switch then delivers the voltage from the mA select switch, item (F), to the filament transformer. The filament transformer is a step-down, meaning the voltage supplied to the filaments is lower by a factor equal to the turns ratio (Tr) rather than the voltage supplied to the filament transformer. Similarly, the current increases across the filament transformer in proportion to the Tr, typically 5:1 or 10:1.

Looking at figure 1–15, mA selector switch (F), you see that if you decrease the resistor selected in the primary of the filament circuit, it applies more voltage to the primary of T_1, thus increasing the filament voltage in the secondary. When filament voltage is increased, filament current is increased and the X-ray tube has more electrons available for current flow, thus increasing mA in the SHT transformer.

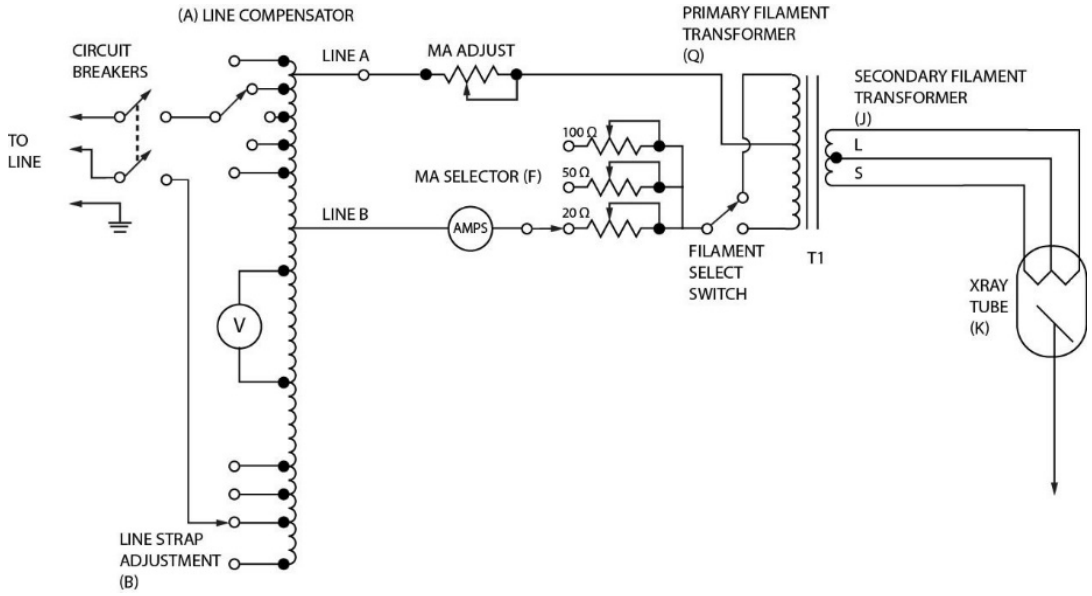

Figure 1–15. mA control and filament circuit.

Load compensation

If you place a voltmeter across a wall plug with no electrical devices or appliances connected to the line, the voltmeter will read the "no-load" voltage of the line. When you connect devices and appliances to the wall plug, the voltmeter reading will fall off. This change of line voltage is called line drop. If a sufficient number of devices are connected to the line, rated line current will flow and the voltmeter will read the "full-load" line voltage. With the no-load and full-load voltages known, the regulation of the line can be expressed as a percentage when these values are placed in the following formula:

$$\text{Percent of line regulation} = 100\left(\frac{E1 - E2}{E2}\right).$$

In this formula, E1 equals no load line voltage and E2 equals rated load line voltage.

When the line regulation is large, the difference between the no-load voltage and line voltage at various loads tends to be large. You can see the voltage delivered to the X-ray tube depends somewhat on the line regulation for a given presetting of the kVp selectors and the no-load voltage.

Line drop affects the kVp delivered by the high-tension transformer to the X-ray tube. When the X-ray tube current is made to increase, the line drop increases. This calls for an increase in PHT transformer voltage to obtain a constant kVp at the X-ray tube. The opposite would be true when the X-ray tube current is decreased. The changes in PHT transformer voltage, which compensate for line drop, are referred to as "load compensation." Figure 1–16 illustrates this concept. The load compensation is an automatic control circuit aligned with the kVp and mA selection.

The term "load" refers to current; therefore, the term "load compensation" must refer to a compensation for an additional current demand. When we increase filament current in the X-ray tube, more electrons become available for current flow from cathode to anode. This additional current through the X-ray tube increases the power demand of the circuit. You may recall the formula for

power is $I \times E = P$; so, an increase in power demand of the SHT increases the power demand of the PHT. This increased power demand on the PHT results in additional current to supply the additional power. The additional current drops the line voltage down, and the result is low kVp supplied to the PHT. Load compensation is the method used to supply the increased current demand and still maintain a constant kVp throughout varying loads.

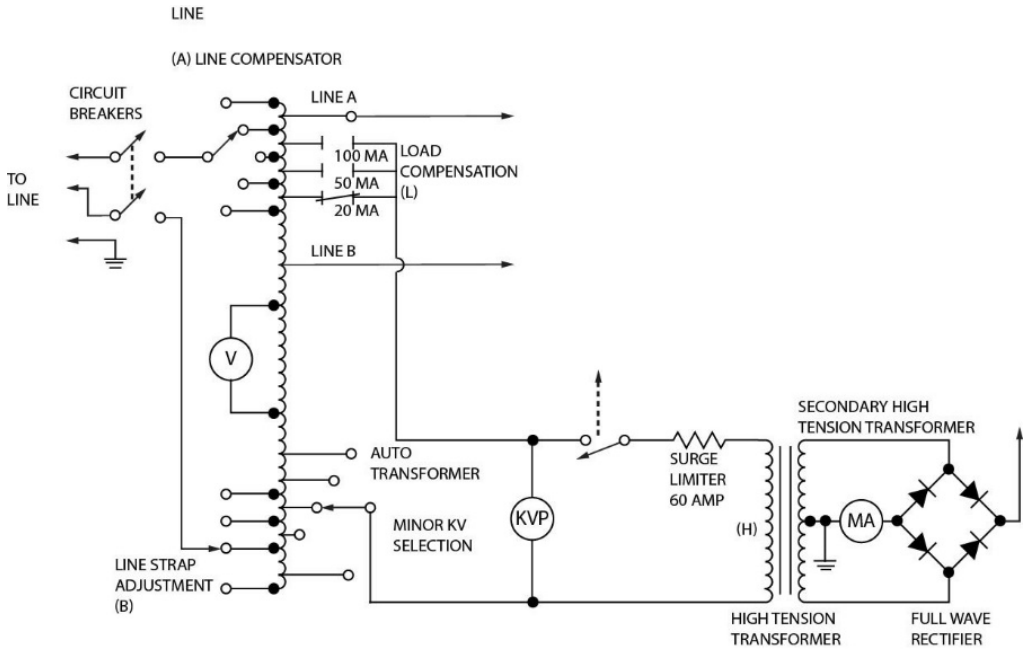

Figure 1–16. Load compensation circuit.

Use figure 1–17 for the following example. With the mA selector switch (F) in the position shown in figure 1–17 (A), let's imagine we have 20V at 5 amperes (A) of current flow in the X-ray tube filament circuit. Hypothetically, let's say that with 100 watts (W) of power (20V x 5A = 100W), the filament boils off enough electrons to produce 25 mA of current flow from the cathode to anode when the X-ray tube is conducting. By moving the kVp select switch (C), we select the kVp applied to the X-ray tube. Let's select 100 kVp for this example. When the main contactor (E) closes, you have 100kVp applied to the X-ray tube with 25 mA current flow through it. This would give us a total power requirement in the SHT of 2,500W (0.25A x 100,000V = 2,500W). Remember, we previously stated that the power to the input of the primary side of the transformer must equal the power output of the secondary side.

Therefore, if the SHT requires 2,500W, the PHT must supply 2,500W also. If the voltage across the PHT is 100V, the current through the PHT must be 25A (100V x 25A = 2,500W) to give us 2,500W. The current path for the PHT starts at the bottom of the autotransformer in figure 1–17 (A), from the major kV selector switch (C) to the minor kV selector switch (D), and then through the bottom of the PHT transformer (H). From the bottom of the PHT transformer, current flows through the surge limiter designed to protect the PHT circuitry in the event a malfunction should allow excessive current flow. The surge limiter is usually rated at approximately 60A or less. The current then flows back to the top of the autotransformer.

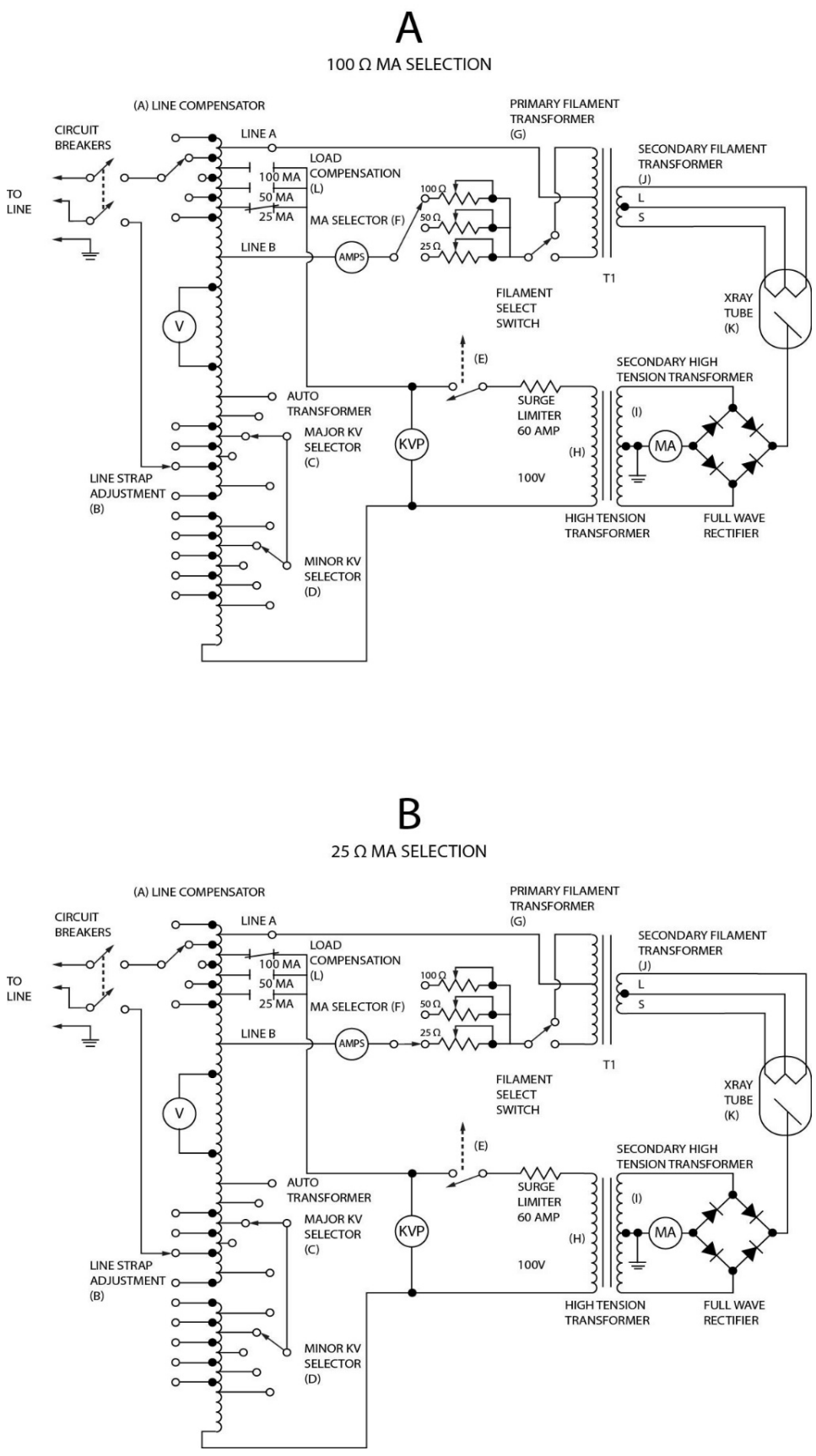

Figure 1–17. Load compensation circuit at different mA settings.

Now, you can see an exposure taken at 100,000V and 25mA will not exceed the PHT surge protection. The incoming line voltage can supply 100V at 25A without dropping significantly enough to affect kVp, as long as the X-ray system is the only system tied into the main transformer supply. However, if the X-ray system shares the same main transformer supply with another piece of equipment (e.g., another X-ray system), then the line regulation may vary enough to affect kVp output. Normally, a pre-procurement survey by you or your regional medical equipment repair center (MERC) will determine the need for a separate transformer to maintain the proper line regulation for your X-ray generator requirements.

If you move the mA selector switch (F) to the 25Ω (ohms) position (fig. 1–17, B), you would increase the filament voltage and, as a result, increase the conduction of the X-ray tube. Let's say 300W (30V x 10A = 300W) would give us 100mA of current flow from cathode to anode when the X-ray tube is conducting. With 100kVp selected, the total power requirement of the SHT would be 10,000W (100,000V x 0.100A = 10,000W). With 10,000W required in the SHT, the PHT must also supply 10,000W. If the load compensation switch (2) were left in the position shown in figure 1–17 (A), there would be 100V at 100A to supply 10,000W to the PHT (100V x 100A = 10,000W). Obviously, the circuit shown in figure 1–17 (A) cannot supply the 100A of current needed to meet the power demand of the SHT, so we must implement an alternate method of increasing power to the PHT. By using the load compensation switch (L), the voltage applied to the PHT can be increased to supply the necessary power. If switch (L) at the 25mA setting is in the opened and the 100mA is closed, then instead of applying 100W to the PHT, let's say 25V is applied to the PHT circuit (fig. 1–17, B). With 250V applied to the PHT, the current necessary to give us 10,000W would be 40A (250V x 40A = 10,000W). The circuit shown in figure 1–17 (B) can supply 250V at 40A without opening the overload surge limiter rated at 60A and without decreasing the kVp during the exposure.

As you can see in the examples of load compensation, when the load requirement of the SHT increases, the power demand on the PHT increases. To supply this additional power without causing a circuit overload or decreasing kVp due to line voltage decreasing, we increased the voltage applied to the PHT to keep the circuit current within safe limits. Therefore, load compensation is nothing more than increasing power applied to the PHT when the load requirement of the SHT increases.

Many of the X-ray generators in use by the Air Force use this or similar types of circuits to control kVp throughout various changing loads. Some generators are now incorporating falling load generators at higher frequency ranges to do the same thing, thus eliminating the need for bulky autotransformers.

Space charge compensation

The space charge effect is based on the assumption that a random cloud of electrons exists around the heated filament of an X-ray tube. This cloud of electrons has a limiting effect on the current across the X-ray tube when the tube is operating below saturation. This limiting effect exists because the electrons which move from cathode to anode and constitute the tube current, must travel through this negative electrostatic field set up by the electrons around the filament. The positive-anode potential tends to cancel out this negative field, and the electrons tend to move toward the anode. As higher and higher positive potentials are impressed on the anode, the canceling effect increases and a greater number of electrons move toward the anode, providing an increase in tube current. Space charge effect can then be defined as the rise or fall in tube current due to a charge in anode-to-cathode voltage on the X-ray tube. When the tube voltage is increased, tube current increases. Conversely, when the tube voltage is decreased, the tube current decreases. The space charge compensation circuit is designed to keep mA stable when the kVp is changed. Simply, if you dial up 100mA you want 100mA no matter what kVp is selected. Well, as just explained, a 50kVp anode is less attractive to the thermionic cloud than a 100kVp anode. Something must be done so mA will always be the same.

If the tube voltage is raised high enough, the space charge about the cathode is completely canceled and a further rise in tube voltage will not give rise to an increase in tube current. When this situation

occurs, the X-ray tube is said to be operating above saturation. This means the electrons are not allowed to form a cloud around the cathode, but are moved across the tube as soon as they are boiled off of the cathode.

From our previous explanation, you can see that below saturation the tube is basically a function of tube voltage, whereas above saturation the tube current is a function of filament temperature. Theoretically, the previous statement is correct; however, the tungsten emitter or cathode is not absolutely pure. This results in a condition where below saturation tube current may vary slightly with filament temperature, and above saturation the tube current varies slightly with tube voltage. X-ray tubes are generally operated above saturation, under which a problem exists in maintaining current throughout the usable range of kVp as desired in medical radiology. Radiographic techniques demand a constant tube current independent of tube voltages. To operate the tube in this manner, it is necessary to change the filament temperature each time the X-ray tube voltage is changed. Figure 1–18 (M) illustrates this circuit—called the space charge compensation circuit.

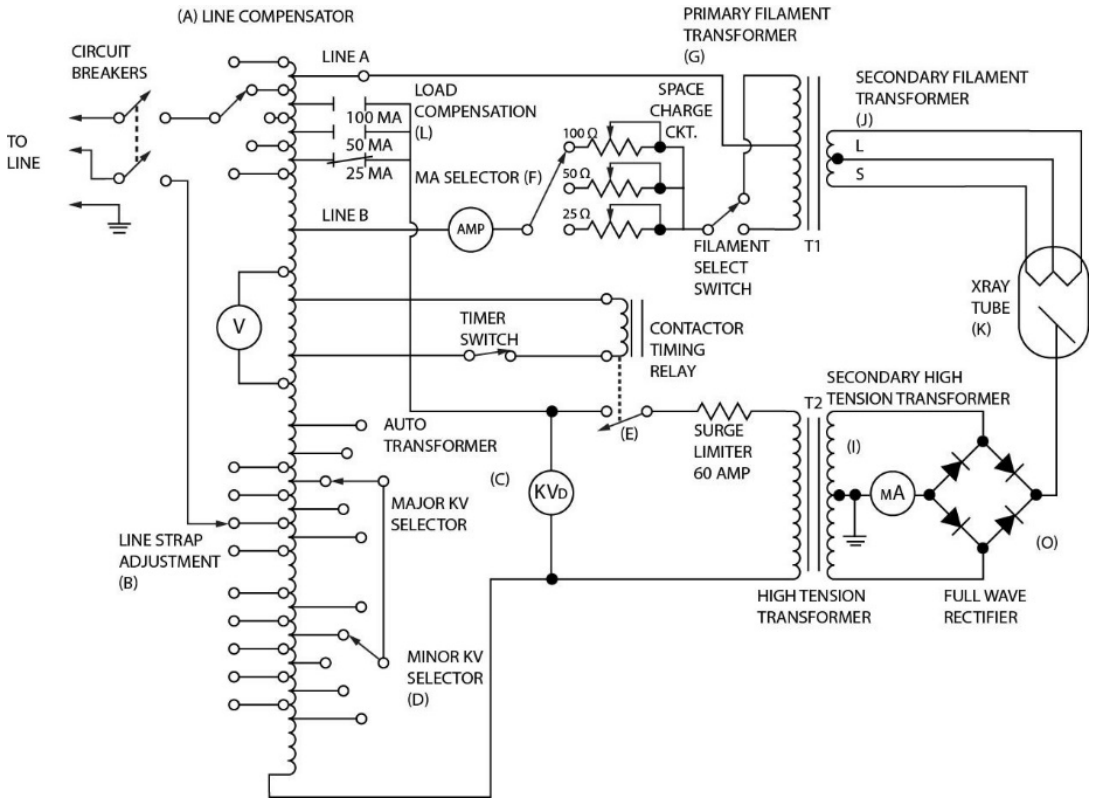

Figure 1–18. Space charge compensation circuit.

As different mA selections are made, precision variable resistors are tapped to vary the filament current and, thus, raise or lower the filament temperature to compensate for various kV selections. Space charge compensation can be summed up as an attempt to obtain a constant tube current, regardless of the tube voltage. To accomplish this, as tube voltage is raised, filament temperature must be lowered. Conversely, as tube voltage is lowered, filament temperature must be raised. The amount of compensation necessary is dependent upon the mA value selected and, thus, varying amounts of compensation must be used for different mA values selected. You should note that as higher mA values are selected, a lower value resister in the mA circuit is used. The reason for this is the higher the mA selected, the more space charge compensation is needed. Space charge compensation becomes more critical at the high and low values of kVp.

604. High-voltage rectification

When alternating current (AC) power is available from locally installed generators or commercial power lines, the electronic equipment in your hospital is normally powered from this source. The AC power may be 1ϕ or 3ϕ, and of various voltages (115 and 208V being the most common). Regardless of the type of power supply used, the rectification principles are the same. Before discussing power supplies, let's take a brief look at some rectifier devices.

Rectifier devices

Rectifier devices are used to change AC to pulsating direct current (DC). They are thermionic (tube type) or semiconductor (solid-state). The specific type used in a particular power supply is determined by the age of the unit, its use, and under what conditions it is to operate. The thermionic types are divided into gas-filled and high-vacuum tubes. The semiconductors include selenium, germanium, and silicon types.

Rectifier devices have various ratings, which must be observed for proper operation. These ratings are the peak-inverse voltage, average DC current, and peak current. The peak-inverse voltage is the maximum potential that can be applied across the rectifier without breaking down; that is, when the anode is negative with respect to the cathode. The value of the average DC current that can be continuously passed through the rectifier is called the DC current rating. This rating depends upon the type of filter used and load current being drawn. The maximum peak current is the value of surge current the rectifier is subjected to during operation. This rating also depends upon the type of filter circuit used. For example, the peak current may be several times the average DC current when using a capacitor-input filter, whereas the peak current may not be more than double the average DC current when using a choke-input filter network.

Semiconductor rectifiers have the advantage of not requiring filament circuitry. Their smaller size, however, limits their current and peak-voltage handling capacity. Selenium rectifiers of 10, referred to as STICKS, are widely used and have an AC rating of about 26V per cell. When used for high-voltage supplies, a number of cells are connected in series or arranged in a voltage multiplier configuration. Germanium and silicon rectifiers have AC voltage ratings up to 1,000V per cell and are extensively used in modern X-ray equipment because of their compact size. Special consideration must be given to protect solid-state devices from current overloads because of the small rectifying positive/negative (p/n) junction area. Overheating caused by excessive current flow will destroy the p/n junction and render the device useless.

Now that we've briefly covered the rectifier devices used in X-ray high tension (HT) power supplies, let's put them in a 1ϕ and 3ϕ system and see how they work.

Single-phase rectifiers

You may recall from basic electronics that DC power can be obtained by connecting a suitable rectifier directly across the AC line voltage, but in most cases, the AC line voltage is not enough to supply the high voltage required for the equipment. Transformers are normally used to produce a variety of voltages for rectification. They may be step-up or step-down, depending on the application. You may also recall 1ϕ rectifiers are divided into three groups: half-wave, full-wave, and bridge (fig. 1–19). Let's review these three circuits.

Half-wave

A half-wave rectifier circuit is arranged so the output current flows only during one-half of the input AC sine wave. This principle (fig. 1–19, A) shows this rectifier has some disadvantages, which are common to all half-wave transformer rectifiers. Filtering is difficult if the load draws appreciable current, and the transformer has a tendency to saturate because the secondary current flow is in one direction only. The efficiency of a half-wave rectifier is low because it uses only one-half of the input AC sine wave. This limits its use to X-ray equipment that requires a low-current drain, such as a dental X-ray unit that normally operates between 10 and 15mA.

Full-wave

A full-wave rectifier circuit is arranged so current flows in the output in the same direction during each half-cycle of the AC input to the rectifier. We can get full-wave rectification by using two diodes connected (fig. 1–19, B). The cathodes of each diode are connected and the junction is connected to the anode end of the X-ray tube. The cathode of the X-ray tube is connected to the center tap of the secondary transformer. The high-voltage secondary may be a center-tapped winding or may be made up of two separate windings connected at the middle.

The secondary AC winding may be considered a voltage source, which produces a voltage of the waveform shown in curve 2 of figure 1–19 (B). This voltage is impressed across diode D1 and the X-ray tube in the series. During the half cycle marked T1, the anode of D1 is positive with respect to its cathode, enabling current to flow in the direction of the solid arrow. This current causes a voltage drop across the X-ray tube, which makes its anode more positive than its cathode. During this same half-cycle, the voltage across points BC makes the anode of D2 negative with respect to its cathode, as shown in curve 3, and this diode is cut off. A half-cycle later, at T2, the voltage on the anodes of the two diodes is reversed; D2 is now conducting and D1 is cut off. The current that passes through D2 flows in the direction indicated by the dotted arrows. This current also produces a positive pulse of voltage across the X-ray tube, as shown at T2 on curve 5. A comparison of curve 5 with curve 4 shows only one diode is conducting at any given instant. The current being contributed by D2 flows through the X-ray tube in the same direction as the current contributed by D1.

You may recall from previous electronic fundamentals that there are two pulsations of current in the output for each cycle of AC in the input. You may also recall the full-wave rectifier is more efficient than the half-wave rectifier and has much less ripple effect. The ripple frequency is twice the input frequency, which also makes filtering easier. Transformer core saturation is greatly reduced, provided both halves of the transformer produce equal voltages and the diodes have identical characteristics.

Bridge

A bridge rectifier, you recall, consists of four diodes connected as shown in figure 1–19 (C). The input to this circuit is applied to two diagonally opposite corners of the network, and the output is taken from the other two corners.

During one half-cycle of the applied voltage (waveform 1, fig. 1–19, C) point A becomes positive with respect to point B. During this time, the voltage across AB is felt across a load consisting of D1, the X-ray tube, and D3 in series. The voltage applied across these diodes makes their anodes more positive than their cathodes and current flows in the path indicated by the solid arrows. The waveform of this current is shown by waveforms 2 and 3. One half-cycle later, D1 and D3 are cut off or reverse biased, and current (waveforms 4 and 5) flows through diodes D2 and D4, and the X-ray tube in the direction indicated by the dotted arrows. The current through the X-ray tube is always in the same direction. This current develops a voltage (output waveform 6). The bridge rectifier is a full-wave rectifier because current flows in the X-ray tube during both halves of a cycle of applied AC voltage.

Bridge versus full-wave

One advantage of a bridge over the full-wave rectifier is, with any given transformer, the bridge circuit produces a voltage output of nearly twice that of the full-wave rectifier. We can illustrate this point by assigning values to some of the components in figure 1–19 (B and C). Presume the same transformer is used in both figures and the peak voltage developed between A and B is 1,000V. In the full-wave circuit (fig. 1–19, B), the peak voltage from the center tap (c) to either A or B is 500V. Because D1 conducts on one half-cycle and D2 on the other half-cycle, the maximum peak voltage rectified is 500V. Therefore, the maximum voltage developed across the X-ray tube is 500V, less the small voltage drop across the diode while it is conducting. In the bridge circuit (fig. 1–19, C), the maximum voltage rectified is felt across the secondary of the transformer (1,000V), which is also felt across the X-ray tube. Thus, you can see the bridge circuit produces a higher output voltage than the full-wave rectifier when using the same transformer.

The second advantage of the bridge circuit is the peak inverse voltage across the diodes is half the peak inverse voltage impressed across the diode in a full-wave rectifier, when designed for the same output voltage. For example, if the two circuits are to produce a 500V output, the transformer secondary in the full-wave rectifier must have a 500V output and a 1,000V peak, while the bridge rectifier has only a 500V peak. When D1 in figure 1–19 (B) is cut off, its anode is made negative with respect to its cathode by a maximum of 1,000V. In figure 1–19 (C), when D1 and D3 are cut off, D4 is conducting and has a very small voltage drop across it. Therefore, practically the entire secondary voltage (Es) is applied across D1. This voltage however, is half (500V) as compared to the full-wave voltage drop across D1.

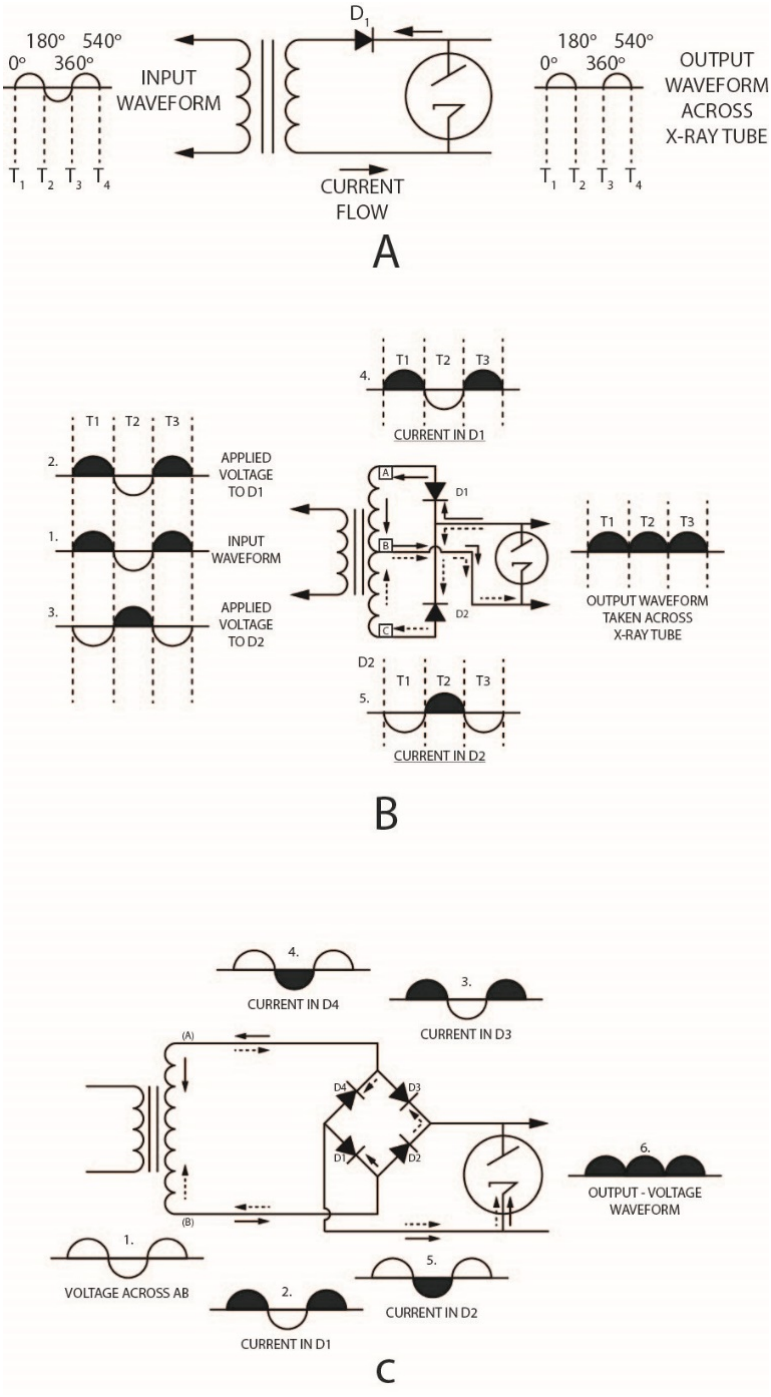

Figure 1–19. Three types of single-phase rectifiers.

Polyphase rectifiers

When constant DC power requirements reach approximately 1 kilowatt (kW) or higher, polyphase rectifiers are normally used. The output of a polyphase rectifier, especially the full-wave type, is closer to a pure DC than that of a comparable 1ϕ rectifier; thus, less filtering is needed. In addition, the quality of X-ray generation is increased and exposure times can be shortened. Polyphase power supplies normally use 3ϕ inputs connected to locally installed generators or commercial power lines. 3ϕ power is simply three 1ϕ sources separated by one-third of a cycle or 120°.

In a 3ϕ system, the voltage is usually raised or lowered by means of three 1ϕ transformers, or by one 3ϕ transformer. In either case, the windings are connected in a wye-wye, delta-delta, wye-delta, or delta-wye configuration (fig. 1–20).

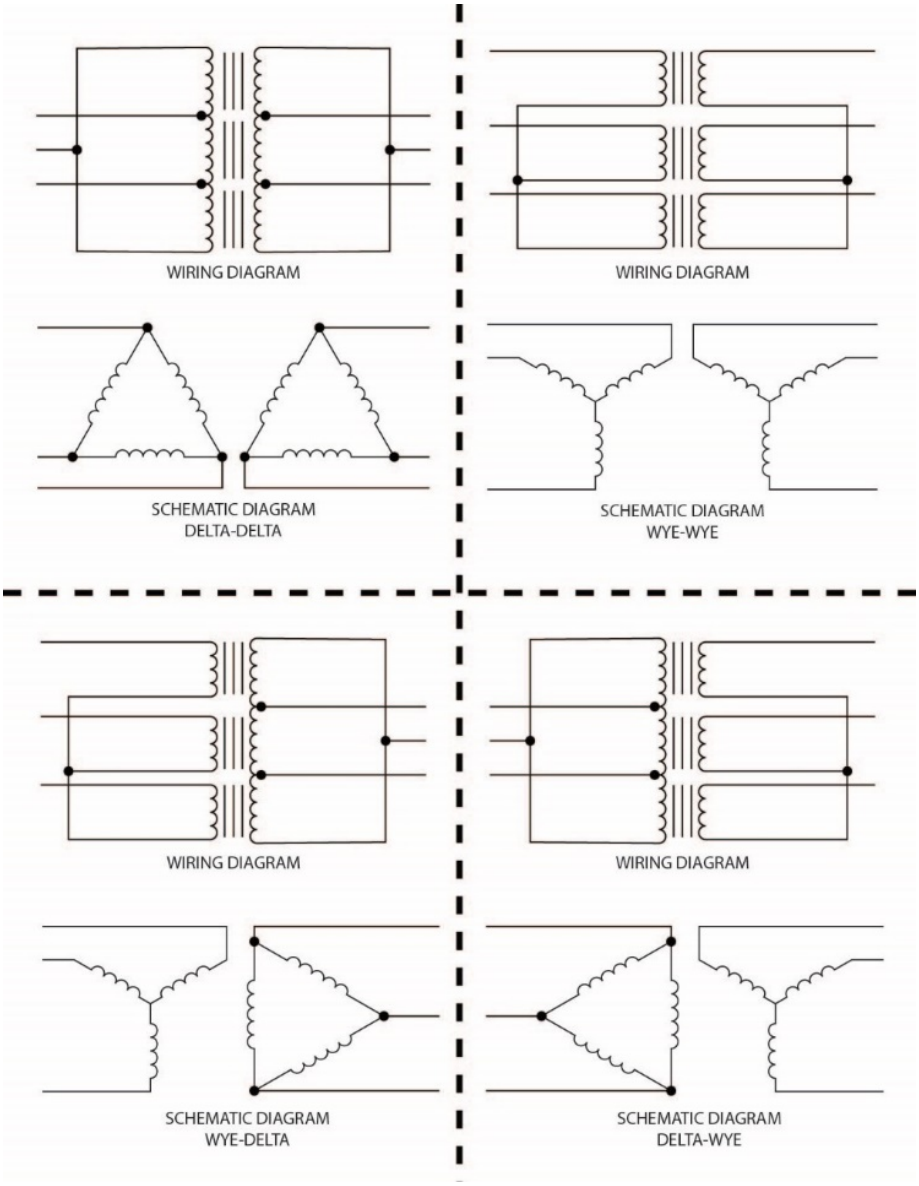

Figure 1–20. Three phase transformer configurations.

In the illustration, the primary windings are shown on the left-hand side and the secondary windings on the right of each connection. The output AC voltage available at the secondary of any 3ϕ transformer depends upon its Tr and the type of configuration used. The AC output of the wye-wye and delta-delta configuration depends strictly upon the Tr of the transformer. This relationship is expressed as:

$$Es = Ep \times Tr.$$

In this equation, Es equals secondary voltage, Ep equals primary voltage, and Tr equals the turns ratio of the transformer.

The Tr of the transformer is equal to the number of turns in the secondary divided by the number of turns in the primary.

In the wye-delta configuration, the output is determined by:

$$Es = \frac{(Ep \times Tr)}{1.73}.$$

You can see the reason for this by referring to the wye-delta configuration. The primary wye configuration means the line voltage is applied to two coils in series and not across any single primary winding. The result is the voltage in each phase winding is decreased to the value of the line voltage divided by 1.73. The value 1.73 (equal to the square root of three) takes into consideration 120° of difference in phase angle between the two windings. The secondary winding is delta connected and delivers the full Es to each phase line. Thus, the output voltage is $^{1}/_{1.73}$ of the voltage that would be obtained from either wye-wye or delta-delta configurations.

The output voltage of the delta-wye configuration is determined by:

$$Es = Ep \times Tr \times 1.73.$$

The effect is just the opposite of that produced in the wye-delta configuration. The Ep is the full-line voltage and because of the secondary wye connection, the Es is 1.73 multiplied by the phase voltage.

The most commonly used 3ϕ transformer is manufactured with three cores and two windings on each core. Figure 1–21 shows its construction; it can be connected in any of the basic configurations shown in figure 1–20. Because of the many factors involved, we will not give a detailed description of the construction; however, we will clarify the basic operation.

In figure 1–21, windings A1, B1, and C1 are the primary windings. A2, B2, and C2 are the secondary windings. The C1 and C2 windings are on the same core in the wiring diagram detailed on the A-construction illustration, and are drawn at the same angle in the primary and secondary of the schematic diagram (fig. 1–21, B) to show they are on the same core. The same holds true with the other windings. For example, the A1 and the A2 windings are drawn vertically in the schematic diagram; this indicates they are wound on the same core whether or not the letter designations of A1 and A2 are used. A current in the A1 winding induces a voltage in the A2 winding; a current in the B1 winding induces a voltage in the B2 winding; and so forth.

Now that you are familiar with the types of connections and the reasons for them, let’s incorporate them into some typical 3ϕ rectifier circuits.

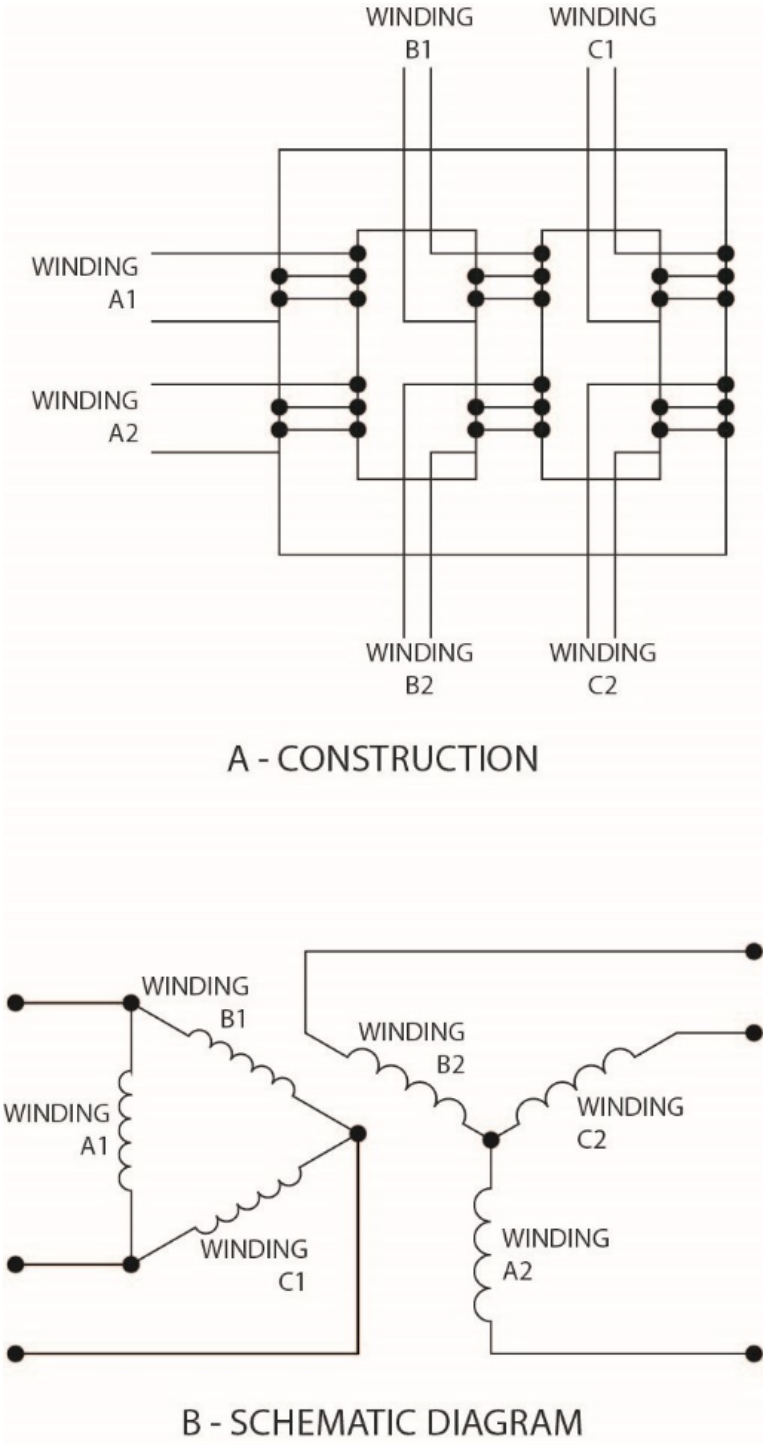

Figure 1–21. Three-phase transformer construction.

Three-phase rectifiers

Figure 1–22 shows a half-wave, 3ϕ rectifier. Notice the delta-wye transformer configuration is used, thereby taking advantage of the additional step-up action afforded by this type of configuration. In this circuit, each diode draws current one-third of the time; that is, each diode is a half-wave rectifier and conducts during the time its anode is positive (fig. 1–22, B). In this figure, numbers 3, 1, and 2 represent the secondary transformer voltages applied to the diodes, and A, B, and C represent the rectified DC output voltage from the diodes. Notice, as the result of rectification, the output current is three pulses of DC for each 3ϕ cycle of AC. The section of the curve labeled a-b-c represents the output of diode one (D1) during the fraction of the cycle this diode is conducting; section c-d-e represents the output of diode two (D2) during the fraction of the cycle it is conducting; and section e-f-g represents the output of diode three (D3) during the time it is conducting. The output total waveform depicted as solid lines in the figure show the output would require less filtering than the output of a 1ϕ, half-wave rectifier. The ripple frequency equals the supply voltage multiplied by the number of phases; in our previous example of a 60 hertz (Hz) AC input, that gives us a 180Hz ripple.

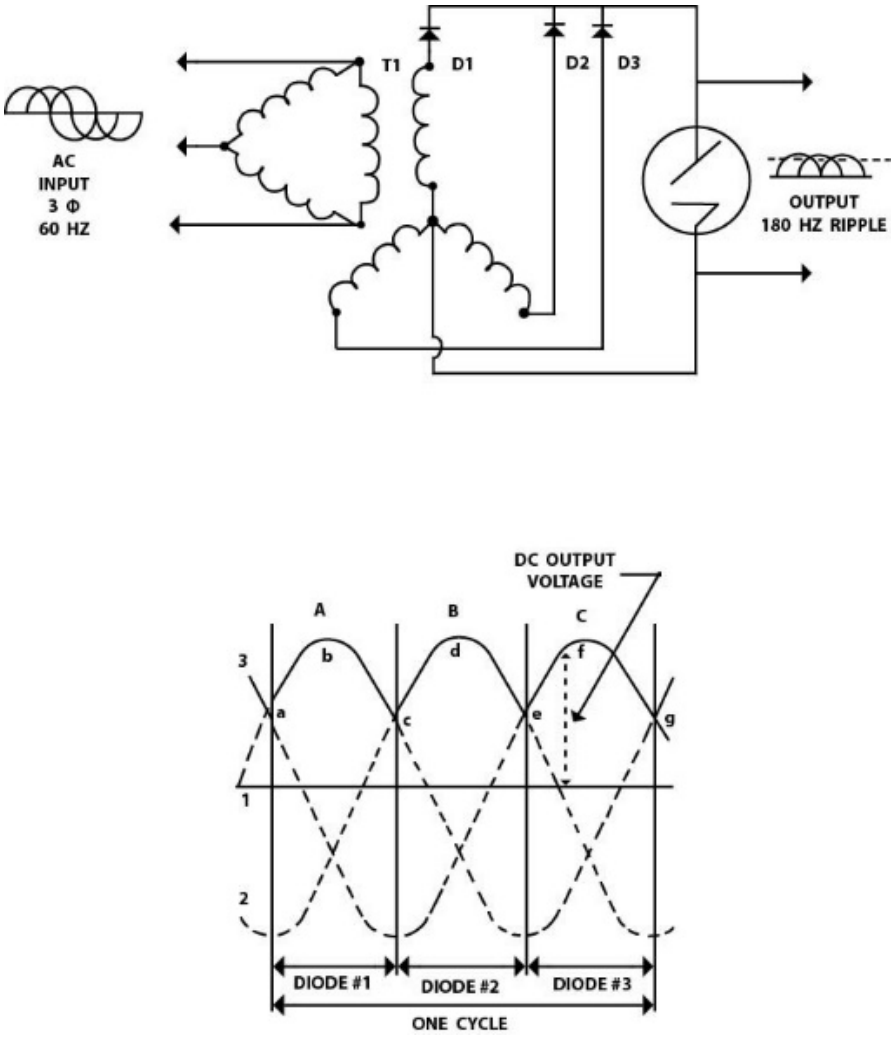

Figure 1–22. Three-phase, half-wave rectifier.

Figure 1–23 (A) shows a full-wave, 3ϕ power supply circuit. This type of circuit is commonly used because of the high-powered load requirements used in large X-ray equipment. Because of the delta-wye transformer configuration, it develops an Es 1.73 times the Es.

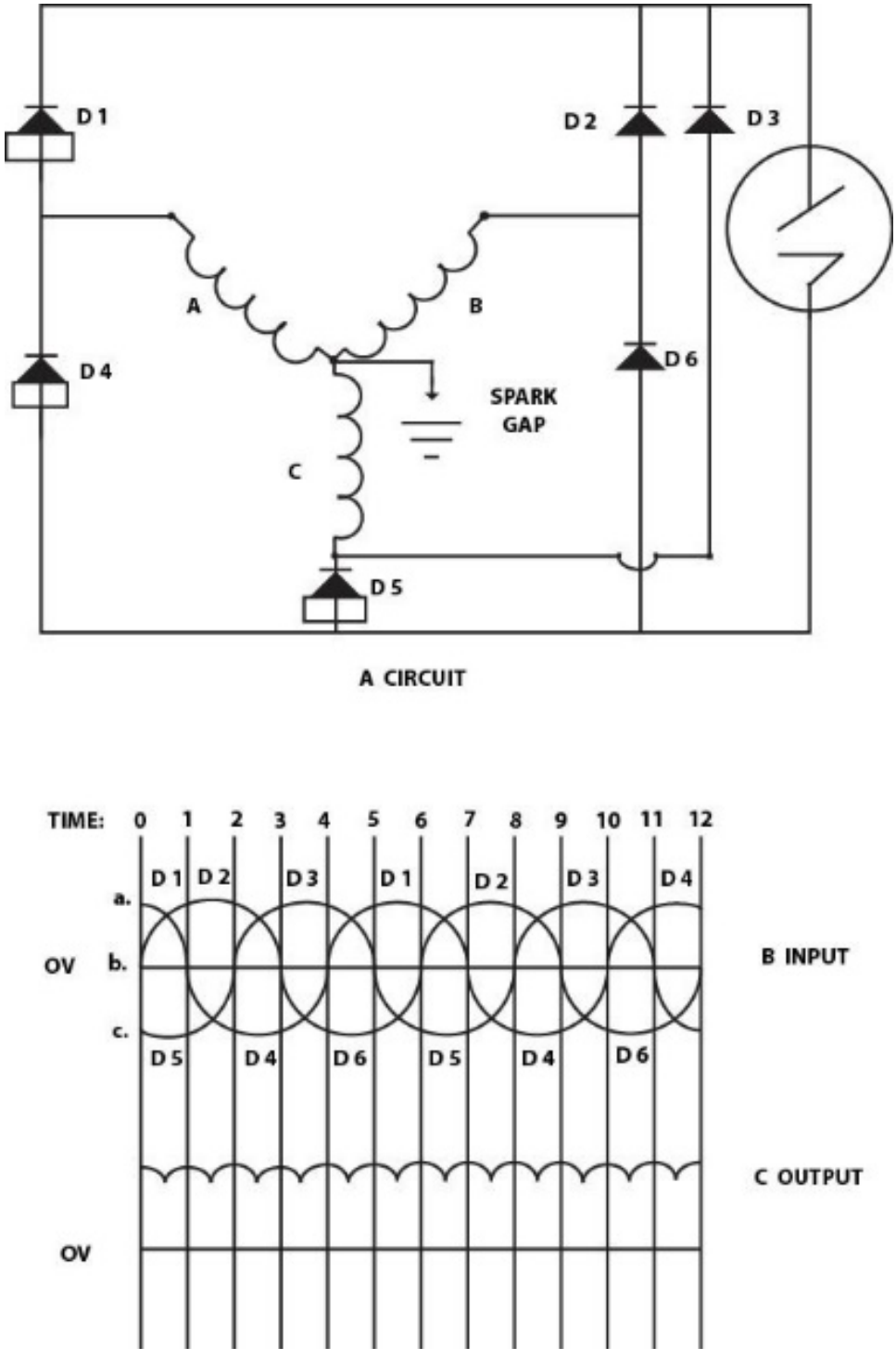

Figure 1–23. Three-phase, full-wave rectifier.

Figure 1–23 (B) shows the phasing of the Es (phases a, b, and c), and indicates the points at which maximum positive voltage appears on the anodes of diodes 1, 2, and 3, and maximum negative

voltages on the cathodes of diodes 4, 5, and 6. Figure 1–23 (C) illustrates the total output current and is subdivided according to the conduction periods of the six diodes.

Referring to figure 1–23 (A and B), let's start on the curve where a maximum positive voltage exists on the anode of D1 in figure 1–23 (A) (time "0"). With the top of winding "A" positive, the current path would be from the bottom of winding "C" through D5, through the X-ray tube to D1, and back to the top of winding "A." Current would also flow from the top of winding "B," through D6, the X-ray tube, and again back to D1 to the top of winding "A." Sixty degrees later (time "1"), winding "A" would be at 0V, with winding "B" positive and winding "C" at a negative potential. The current path would now be from the top of winding "A" through D4, the X-ray tube, through D2 and then to the top of winding "B." Current also flows from the bottom of winding "C" through D5, the X-ray tube, and back to the top of winding "B" through D2.

You can see from figure 1–23 (B) that 180° after time "0," the bottom winding "C" would be positive, with winding "A" negative and winding "B" at 0V. The current path in figure 1–23 (A) would be from the top of winding "A," through D4, the X-ray tube, and down through D3 to the bottom of winding "C." Current would also flow from the top of winding "B," through D6, the X-ray tube, and again down D3 to the bottom of winding "C."

Looking at figure 1–23 (B), you can see that every 60° the current path changes, resulting in six pulses of energy through the X-ray tube within the 360° time period from time "0" to time "6." The result is a ripple frequency three times higher (fig. 1–23, C) than that of a 1ϕ, full-wave rectifier.

High frequency generation

One of the newest methods of producing high voltage in X-ray equipment is high frequency generation. You may also hear these types of generators called medium frequency generators, but we'll only use the term high frequency generators in this career development course (CDC). High frequency X-ray units have many advantages over those previously mentioned. The biggest advantage is high frequency generators produce a very stable and nearly ripple-free voltage to the X-ray tube, which enables very efficient and uniform X-ray production throughout an entire exposure. Another advantage is high frequency units can operate on 1ϕ or 3ϕ power sources, and don't require the special voltage regulators needed in conventional units. High frequency X-ray units also require smaller components than conventional equipment; therefore, these units generally take up much less space—this also makes these generators ideal for portable units. Another advantage is high frequency units are generally easier to calibrate and maintain than other units. Since we've talked about all the good points of high frequency X-ray, let's take a look at how it works.

High frequency generators take the incoming AC line voltage and immediately convert it to DC, which provides a constant source for the X-ray unit—this is advantageous because it allows better kVp regulation. In conventional X-ray units, the kVp is taken from the AC input, which is subject to voltage fluctuations; these fluctuations will cause a corresponding fluctuation in the kVp waveform. Because high frequency generators use a filtered DC supply, the voltage is stable regardless of the AC input.

After the AC input is rectified, it is filtered and sent to a power inverter, or chopper circuit. This inverter converts the DC voltage into high frequency pulses that simulate an AC waveform. These pulses are then sent to the high voltage transformer, which steps up the voltage for the X-ray tube. Because of the transformer action, the power is automatically converted back to AC, so it must again be rectified. After this, the voltage is filtered and smoothed for the actual X-ray tube. Due to the effectiveness of this type of circuitry, the amount of ripple in the voltage applied to the tube is usually less than 1%. As previously mentioned, this allows a very stable and constant voltage to the X-ray tube, thus providing improved image quality with a lower patient dose. Figure 1–24 shows a sample diagram of a high frequency X-ray generator.

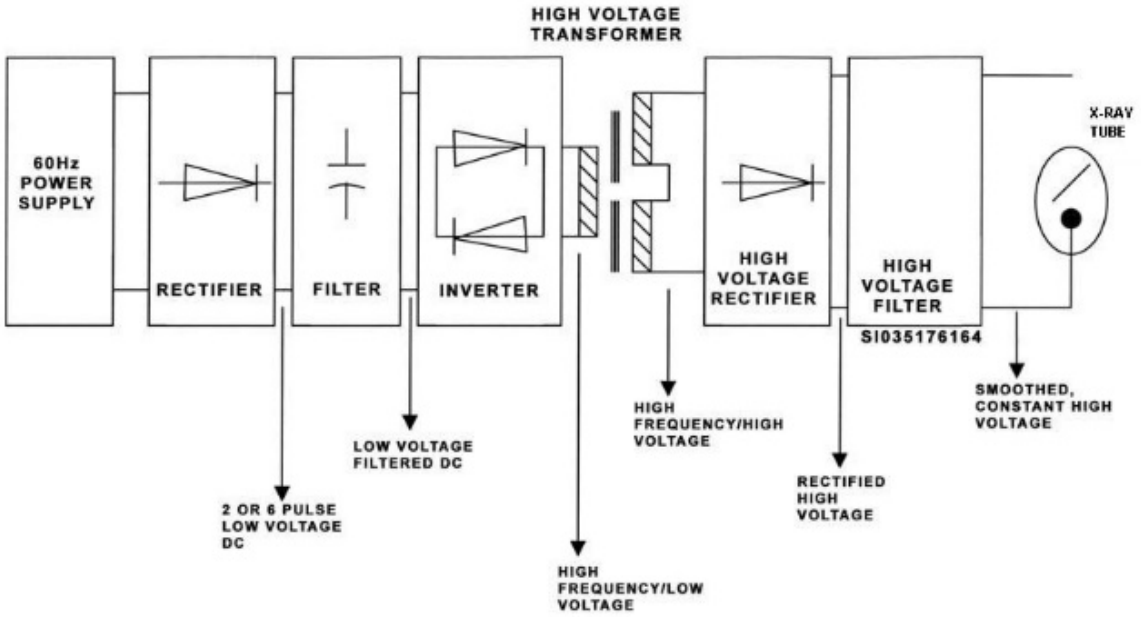

Figure 1–24. High frequency X-ray block diagram.

Falling load generators

One final type of X-ray generator you need to be familiar with is the falling load generator. We will not go into great detail on this particular unit, although you do need to know a little about it so you understand the concept. This particular unit gets its name because it works using the falling load principle. Let's take a look at how this works.

Before we see how the falling load principle works, let's review how a traditional X-ray system works. In a system that relies on fixed techniques, there are particular settings available for the radiological technician to select. For example, if the technician needs an exposure of 240mAs, then they would need to select the mA setting of 240 and the exposure would take 1 second (remember, mAs is a result of mA x time).

NOTE: For our example we won't worry about the kVp setting.

Now, let's compare the above example to an X-ray unit that uses the falling load principle. For this type of machine, the radiologist does not set an exposure time, but tells the generator to produce a particular mAs at a particular kVp. The exposure will begin at the highest mA available, and then drop the tube current during the exposure at given points in time to produce the requested mAs. Let's look at an example: If the operator requests a mAs of 240, the exposure will begin at the highest possible setting—for our example, we'll say 600mA for the first 0.1 second (fig. 1–25). At this time, the 600-mA tube limit is reached and the generator will automatically reduce the mA to 500 until it reaches the tube limit again (0.2 second later, or 0.3 second from exposure start). Once again, the generator reduces the mA to 400 until the limit is reached (0.2 second later, or 0.5 second from exposure start). This produces 240 total mAs.

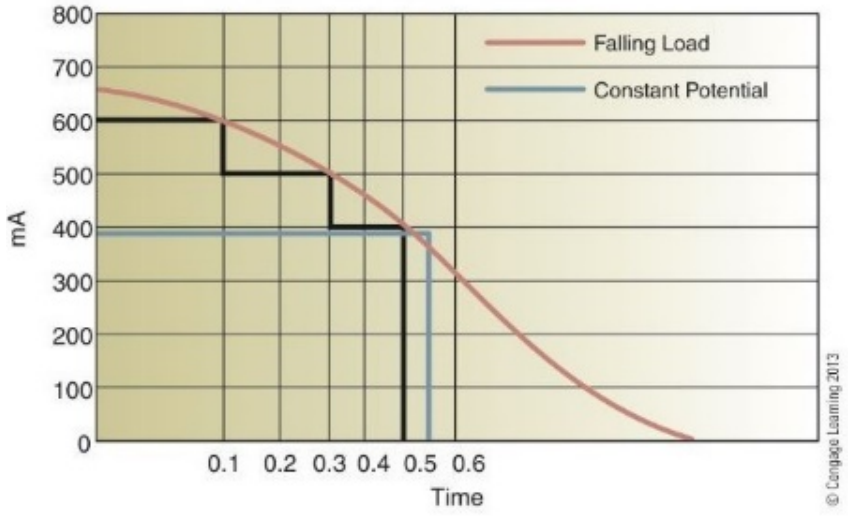

Figure 1–25. Falling load generator. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

To make this explanation easier, look at the following chart that demonstrates the principle:

Operating mA and time	Resulting mAs
600mA x .10 sec	60mAs
500mA x .20 sec	100mAs
400mA x .20 sec	80mAs
Total time = 0.50 sec	Total mAs = 240

As you can see, the requested mAs was achieved in 0.50 second as opposed to the 1 second required for a fixed mA technique.

While you can see the advantages of using the falling load principle in an X-ray unit, there are many disadvantages that come with this type of equipment. First, these units are usually very expensive (traditional units are expensive, but these are usually considerably more!). Most falling-load generators will step down the mA at 70–80 percent of the tube loading capacity. This feature has a disadvantage in that constant-potential generators may actually make better use of tube loading capacity on very short exposures, where they may use 80–90 percent of the tube loading capacity.

Also, because the tube is operated at maximum mA during the start of an exposure, the negative effects of heat shortens the life of the tube. When the technologist cannot select the mA setting, there is less flexibility available for certain examinations.

605. Timing devices

In general, the closing and opening of the PHT transformer circuit governs the time period during which the X-ray tube emits X-rays. After the main switch of the unit closes and all devices are in readiness for the exposure, the PHT transformer circuit closes and the X-ray tube starts emitting X-rays. This continues until the PHT circuit opens again. To control the length of the time period between closing and opening of the PHT circuit, some type of timing device is used. The timing device may close and open the PHT circuit by remote control of a relay, or by the use of electronic devices, such as silicon-controlled rectifiers (SCR).

Radiographic exposures may vary from the millisecond (ms) range to several seconds. The importance of timing circuits is to reduce the exposure of X-ray dosage to the patient. Faster film screens and rare-earth films have also led to the reduction of X-ray exposure to the patient.

Fluoroscopic exposures are controlled by means of a simple foot-operated switch, and the time period of exposure is determined by the time the footswitch is depressed. Again, due to improved fluorescent screens and controls, the time of fluoroscopic exposure has also been reduced. While older units previously used timing devices such as mechanical and synchronous timers, for this lesson, we will look at a couple newer and commonly used types of timers.

Milliampere-Second Timers

Most modern X-ray equipment is designed for accurate control of tube current and exposure time. The mAs timer is usually designed to provide the highest safe tube current for the shortest time of exposure for any mAs selected. In a mAs timer, the product of mA and time determines the number of X-rays emitted and, therefore, the density on the film. A mAs timer is a special kind of electronic timer that monitors the product of mA and time. Because this type of timer monitors the actual tube current, it is located on the secondary side of the PHT. When the secondary side of the high-voltage step-up transformer attains the desired mAs, it interrupts the circuit to terminate the exposure. These timers are used in falling-load generators as well as some capacitor discharge units. Because the high-voltage capacitor, timer, and X-ray tube are operating on the same circuit, mAs value remains constant even when there is a slight fluctuation in the capacitor charging current.

Automatic Exposure Control Timers

The previous versions of timers are satisfactory, but do not compensate for human error, such as the technician's personal judgment of the right technique, pathological changes, and effects of age. To improve the quality and consistency of developed films, automatic exposure controls (AEC) or automatic exposure devices were developed which do not require adjustments from the technician and operate on the amount of radiation transmitted through the patient. Because patients vary in size and shape, an AEC device is very useful in achieving consistent X-ray film densities, which can be difficult when manually setting exposure factors without AEC.

The AEC works on the principle of the photoelectric effect. To help explain this, we can say that when light strikes a special combination of metals, one of the following happens:

- A voltage may be generated (photovoltaic).
- A change in resistance will occur (photoconductive).
- An electron, or electrons, may be ejected or emitted (photoemitting).

The difference in AEC systems lies in the type of device used to convert radiation into electricity. Two types of AEC systems have been used: phototimers (which use a photomultiplier tube) and ionization chambers. Phototimers represent the first generation of AEC systems used in radiography, and it is from this type of system that the term phototiming has evolved. Phototiming specifically refers to the use of an AEC device that uses photomultiplier tubes or photodiodes. The more common type of AEC system uses ionization chambers. Regardless of the specific type of AEC system used, almost all systems use a set of three radiation-measuring detectors, arranged in some specific manner. See the dark grey detectors in figure 1–26 as an example of a typical AEC layout. The radiographer selects the configuration of these devices, determining which of the three individually or in combination actually measures radiation exposure reaching the image receptor. These devices are variously referred to as sensors, chambers, cells, or detectors. These radiation-measuring devices are referred to here for the remainder of the discussion as detectors.

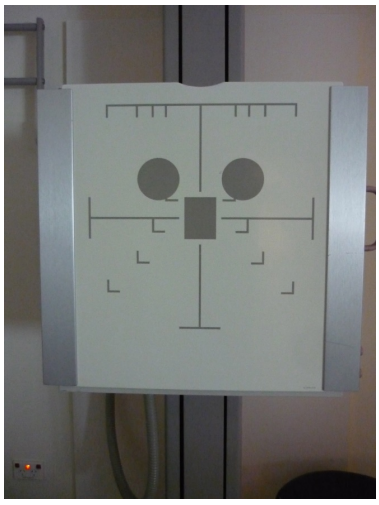

Figure 1–26. Typical AEC detector layout.

Phototimers

Using a phototimer, X-rays are converted into visible light by a fluorescent material, usually a fluorescent screen located between the patient being examined and the photomultiplier tube, and this light is directed toward the photocathode. A photomultiplier tube is an electronic device that converts visible light energy into electrical energy. Phototimer AEC devices are considered exit-type devices, because the detectors are positioned behind the image receptor so that radiation must exit the image receptor before it is measured by the detectors. Light paddles, coated with a fluorescent material, serve as the detectors, and the radiation interacts with the paddles, producing visible light. This light is transmitted to remote photomultiplier tubes or photodiodes that convert this light into electricity. This current is directly proportional to the intensity of light falling on the cathode—the greater the light intensity striking the photodiode, the greater the current flow to the output anode of the photomultiplier. When it receives a sufficiently large charge, the timer trips and the radiographic exposure is terminated. This electrical charge is in proportion to the radiation to which the light paddles have been exposed.

The current from the anode of the photomultiplier charges a capacitor over a period of time. Thus, the total charge on the capacitor is proportional to the total light, which is proportional to the total radiation during the same period. The voltage across the capacitor is fed into the circuits of a phototimer where it is amplified and used to terminate the X-ray exposure. Therefore, it can directly control the exposure timer in accordance with the X-ray required for correct blackening (or density) of the film. Phototimers, however, have largely been replaced with ionization chamber systems.

Ionization chamber systems

Ionization chamber AECs (fig. 1–27) use a thin, parallel-plate ionization chamber, which is positioned immediately above the image receptor. Because thin, parallel-plate chambers are only about 5 mm in thickness, they do not cause an appreciable shadow on the image. It is possible to image the location of the ionization chambers by using an automated AEC exposure at the lowest kVp possible.

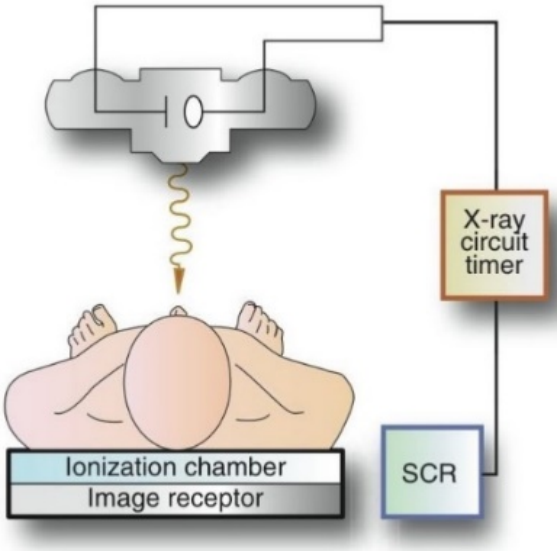

Figure 1–27. Ionization chamber AEC. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

An ionization or ion chamber is a hollow cell that contains air and is connected to the timer circuit via an electrical wire (fig. 1–28). Ionization-chamber AEC devices are considered entrance-type devices because the detectors are positioned in front of the image receptor (between the patient and the image receptor) so that radiation interacts with the detectors just before interacting with the image receptor.

It's important its absorption is kept as low as possible to maintain the image on the X-ray film. Usually, three detector chambers are located within the table with corresponding rectangles labeled on the tabletop to assist the technician in the proper positioning of the patient. Since your five senses can't detect the presence of X-ray, the ion chamber is a good indirect method of measuring radiation.

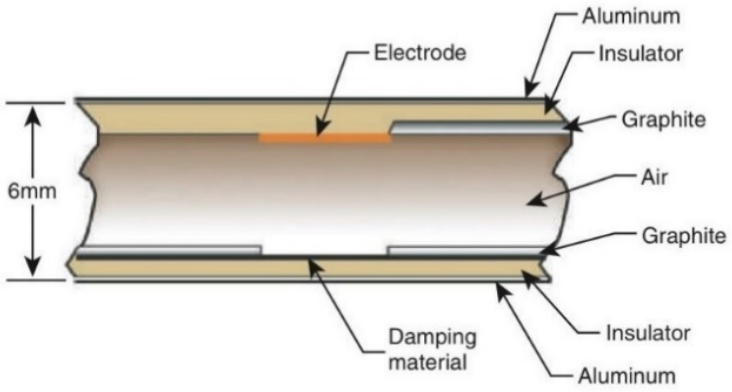

Figure 1–28. Parallel-plate Ionization chamber AEC. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Remember, in the ionization process, X-ray energy strikes an atom and an ion pair is formed. The ion pair consists of the remaining atom and a free electron. It is possible to detect and measure radiation from the number of freed electrons within an ionization chamber. When the ionization chamber is exposed to radiation from a radiographic exposure, the air inside the chamber becomes ionized, creating an electrical charge. This charge travels along the wire to the timer circuit. When it receives a sufficiently large charge, the timer trips and the radiographic exposure terminates. This electrical charge is proportional to the radiation to which the ionization chamber has been exposed.

Compared with phototimers, ion chambers are less sophisticated and less accurate, but they are less prone to failure. Most of today's AEC systems use ionization chambers. When using an AEC, problems may be avoided by considering the minimum reaction time and the backup time.

Minimum reaction time

All AECs have a minimum reaction (or response) time, which is determined by the length of time necessary for the AEC to respond to the radiation and for the generator to terminate the exposure. This delay is caused primarily by the turn-off delay in the high-voltage circuits. Old phototimers had a minimum reaction time of 0.05 second or less. Modern ionization chambers with SCRs may have a minimum reaction time of less than 0.001 second. AECs are sometimes incapable of terminating exposures quickly enough, especially with extremely high-speed image receptors during high-kVp chest radiography. In this instance, radiographic technology has outstripped itself and either a manual time or a slower image receptor must be used, although some manufacturers use an exposure-monitoring circuit to terminate low-dose-rate exposures quickly.

Imaging small body parts with an AEC may not be possible. If such an image is overexposed, the AEC should be turned off and a manual exposure should be used.

Backup time

Nearly all units equipped with automatic exposure permit (and may have electronic interlocks that require) a manual backup time to be set. There are numerous reasons why an automatic exposure control may be improperly set. For example, if one forgets to activate the AEC in a wall unit, this may leave an AEC in a table unit waiting for exposure by a tube that is instead directed toward the wall (and through a patient). Because the wall AEC is not receiving any radiation dose, the exposure would not cease until the tube overload protector activated. Not only is this an expensive waste of tube life, the radiographer error would cause an excessive radiation dose to the patient. The image

would also be overexposed, requiring it to be repeated, adding to the patient's dose again. Backup times cannot exceed the tube limit and should be set at 150 percent of the anticipated manual exposure mAs. According to Code of Federal Regulations (CFR) 21, generators must terminate the exposure at 600 mAs for exposures above 50 kVp and 2,000 mAs for exposures below 50 kVp (primarily during mammography).

606. X-ray tubes

Now that we have discussed much of the circuitry related to X-ray devices, it is time to discuss where the magic actually happens, the X-ray tube. The electrical production of X-rays is only possible under very special conditions, including a source of electrons, an appropriate target material, a high voltage, and a vacuum. The X-ray tube is the device that permits these conditions to exist, and it is within the tube that X-ray photons come into existence. The tube consists of a cathode and an anode enclosed within an envelope and then encased in a protective housing. The X-rays, utilized in diagnostic radiography and in much of radiotherapy, are produced by accelerating electrons in an electric field between two electrodes.

Basic principles of an X-ray tube

The two electrodes in an X-ray tube have a large potential difference between them, usually not less than 40kVp. The electrode that attracts electrons has a small plate made of a heavy metal on its surface, or the whole electrode may be made of such metal. Heavy metal is used because its atoms have large numbers of electrons orbiting around their nuclei, and the electron orbits constitute regions of strong negative electric fields.

Since the accelerated electrons are negative electric charges, forces of repulsion retard them as they approach the atoms at the surface of the heavy metal plate and interact with their associated negative electric fields. Also, because the strength of an electric field varies inversely with distance, the force of repulsion increases as the accelerated electrons approach the fields. The accelerated electrons lose their kinetic energy very rapidly at the surface of the metal plate, and energy conversion consequently occurs. This kinetic energy of the accelerated electrons is converted in three principal ways:

1. A very small fraction (approximately 0.2 percent) is converted to X-rays.
2. Approximately 99.8 percent is converted into heat by increasing the thermal vibration of the atoms of the plate, the temperature of which may rise considerably.
3. Some of the electrons have sufficient energy to eject orbital electrons from the atoms of the plate material, which are ionized. The secondary electrons produced in this way may escape from the surface of the plate and subsequently be recaptured by it, producing further heat or X-ray.

The process of interaction between the electrons and the material of the electrode towards which they are accelerated is, therefore, primarily one of retardation and energy conversion. For simplicity, the electrons are often described as bombarding the electrode, or other similar terms. Such terms are convenient and used throughout this volume; but when they occur, you need to remember they imply the more precise and complex process we just discussed.

The two electrodes, and the accelerated electrons moving between them, are parts of a circuit, which, in its simplest form, is completed by a supply source such as the secondary winding of a high-tension transformer and connecting cables. The electrons, as well as supplying the energy required for X-ray production, also make up the current in that part of the circuit. During the flow of current through the completed circuit, electrons are displaced from the atoms in the material of the heavy metal plate and are replaced by the accelerated electrons after they have been retarded and kinetic energy conversion has occurred.

Since electrons cannot be sufficiently accelerated or their behavior properly controlled in air or any other gas at atmospheric pressure, the two electrodes are sealed in an enclosure in which the required

degree of low pressure, or vacuum, can be maintained. The enclosure is usually made of glass (called a glass envelope) and, together with the electrodes, is called an X-ray tube (fig. 1–29).

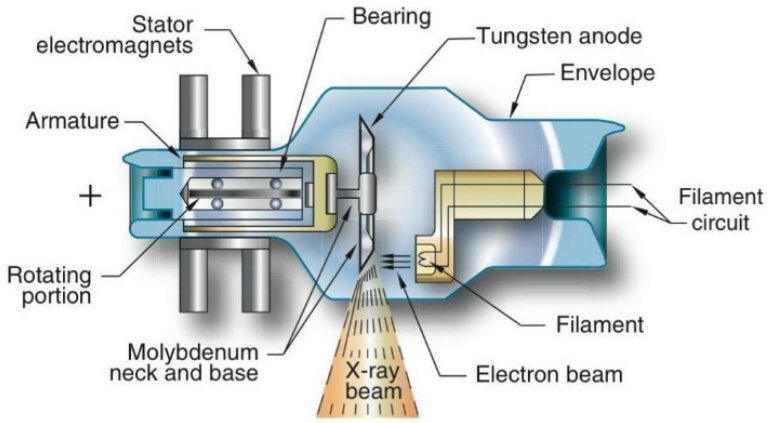

Figure 1–29. Typical X-ray tube. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

The electrode where the electrons originate is called the cathode, because it is normally at a negative potential. The electrode at which the electrons are arrested (bombard) is called the anode, because it is normally a positive potential. The high potential difference between the cathode and anode establishes an electric field in the region between them; this field causes electrons from the cathode to be accelerated towards the anode. The limited area of the anode surface, which is actually bombarded by electrons, is called the tube focus, focal area, or the focal spot.

As you can see, the X-ray tube is a vacuum or semi-vacuum enclosure containing two electrodes and is, therefore, an electronic diode designed for the specific purpose of accelerating electrons to high velocity and converting part of their kinetic energy into X-rays. Older types of X-ray tubes, no longer in general use, were an example of the cold cathode gas diode. The type of X-ray tube most widely used today is the hot cathode (thermionic) vacuum diode. You may also encounter special-purpose triode tubes also called thermionic grid controlled tubes.

Let's examine the specific features of an X-ray tube that distinguish it from other diodes:

- The shape and size of the tube must be designed to prevent external arcing (or electric discharge) between the electrodes, having regard to the high potential difference between them and the efficiency of any insulating medium surrounding the tube.
- The great quantities of heat generated at the focus, and in the anode, generally require that they be provided with an adequate cooling system to prevent melting or other thermal damage.
- The high kVp applied to the anode and cathode affects the penetrating power and intensity of the resulting X-ray. The electron current across the tube affects the intensity of the X-ray. It is desirable to be able to accurately determine radiographic exposure factors. The tube should be of a type that permits independent variations of potential difference and current flow.

Cathode assembly

The cathode is the negative side of the x-ray tube. The function of the cathode is to produce a thermionic cloud, conduct the high voltage to the gap between cathode and anode, and focus the electron stream as it heads for the anode. The cathode is a complex device and is referred to as the cathode assembly. This assembly consists of the filament or filaments, focusing cup, and associated wiring.

Filament

The filament is a small coil of thin, thoriated tungsten wire. The wire is about 0.1–0.2 mm thick and the coil 1–2 mm wide by 7–15 mm long. It is set in the cathode assembly within the focusing cup (fig. 1–30). Tungsten is the material of choice because of its high melting point (3,370°Celsius [C]). Rhenium (melting point of 3,170°C) and molybdenum (melting point of 2,620°C) are also desirable materials. The high melting point permits the filament to operate at the high temperatures required of an X-ray tube. In addition, tungsten is not easy to vaporize (turn into a gas). Vaporization produces particles that deposit on other surfaces and reduce the vacuum within the tube. The length and width of the filament have a great effect on the ability of the particular X-ray tube to image fine details.

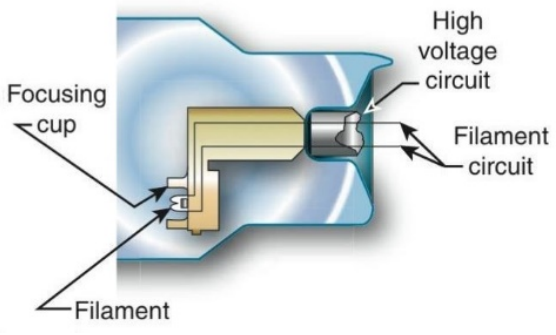

Figure 1–30. Cathode assembly. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Focusing cup

Electrons are focused to a specific area on the target by means of a focusing cup. The focusing cup is merely a negatively charged depression in the cathode structure, which partially surrounds the filament. When the electrons are boiled off the filament, they tend to spread out due to electrostatic repulsion. The purpose of the negatively charged focusing cup is to condense the electrons into a small beam. This concept is illustrated in figure 1–31.

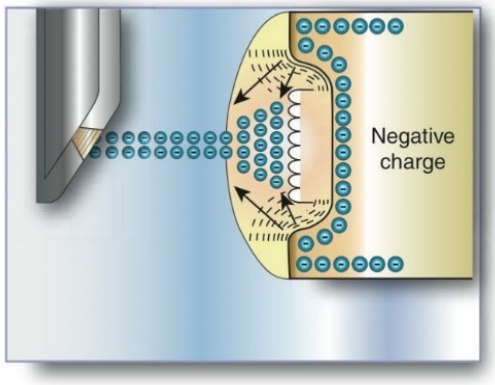

Figure 1–31. The effects of a focusing cup on the electron stream. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Target material

Since the development of X-ray tubes, tungsten has most often been used as the material for target construction because of its high Z (74). As the Z of the target increases, so does the energy of its characteristic radiation. As we discussed previously, the energy to characteristic radiation produced in tungsten reaches 69.5keV, a value considered useful in conventional radiography. A target with a higher Z would produce more energetic characteristic radiation and would appear better suited for X-ray production; however, the target material must meet other requirements in addition to the high Z.

One such requirement is a high melting point. The production of X-rays is a very inefficient process, with less than 1 percent of the energy supplied to the X-ray tube converted into X-rays. The remaining 99 percent is converted into thermal energy (heat). Consequently, the target temperature frequently reaches high proportions. Tungsten's high melting point makes it suitable to withstand these extremely high temperatures. The target must also be able to conduct heat away, as it is generated, to keep temperatures below the melting point. Tungsten's heat conductivity, while not extremely high, is also considered acceptable in view of its other excellent characteristics.

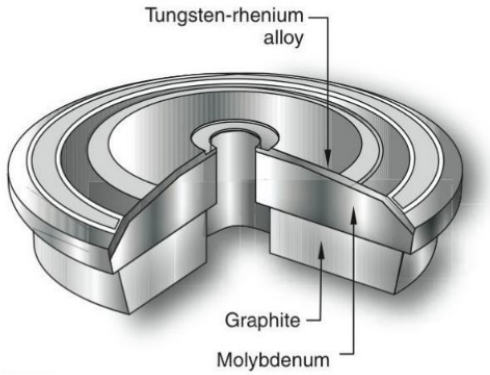

Figure 1–32. Rotating anode made of rhenium-tungsten alloy. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Some X-ray targets are also constructed with a rhenium-tungsten alloy (fig. 1–32). The addition of rhenium makes the target surface more resistant to surface etching at high temperatures. Molybdenum (Mo) is used for the target material in some X-ray tubes made especially for mammography (or X-ray procedures involving the breasts). The characteristic radiation produced in Mo is somewhat lower than that produced in tungsten, which makes the X-ray beam more suitable for mammography.

Anode construction

Even with tungsten's high melting point, overheating of the target to the point of melting, cracking, and pitting is a continuing problem. One way to provide better heat dissipation is to embed the tungsten in copper. Copper conducts heat better than tungsten, and carries the heat away at a much faster rate. Another way to reduce heat buildup is to rotate the anode. By continuously spinning the target, the focal spot presented to the electron beam is constantly changing, thereby making a focal track. This spreads the electrons and heat over a larger area.

Anodes are divided into two types: stationary and rotating (fig. 1–33). Rotating anodes turn during the exposure, thus presenting a much larger target area. Modern rotating anodes permit bombardment of a given area of the target for only 7–50 microseconds. The faster the anode rotates, the better the heat dissipation. The use of stationary anode X-ray tubes has become limited to low-power functions, such as those of dental units. Nearly all units designed for diagnostic radiography utilize rotating anodes because of their greater efficiency. We will discuss how rotating anodes work in depth later; but for now, let's continue with the general anode principles.

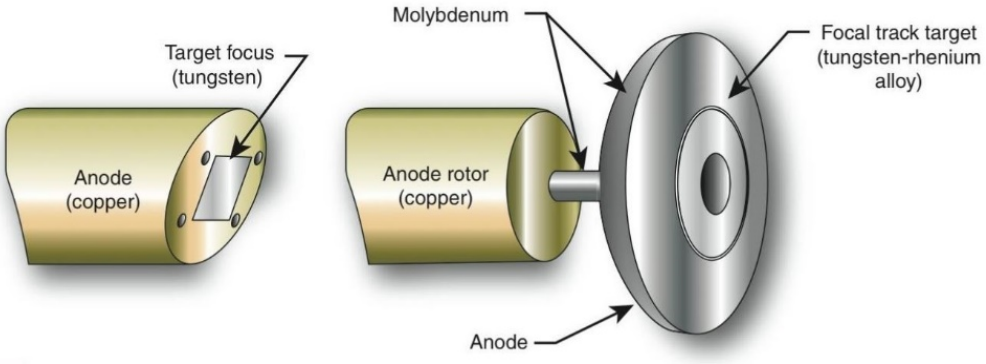

Figure 1–33. Stationary and rotating anodes. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Focal spot (actual)

The area of the target bombarded by the electron stream is known as the focal spot (fig. 1–34). The size of the actual focal spot significantly affects the heat loading capacity of the tube. With larger focal spots, greater heat loading is possible. The size of the focal spot is determined by a combination of three factors:

1. Size and shape of the filament.
2. Size and shape of the focusing cup.
3. Angle of the target surface.

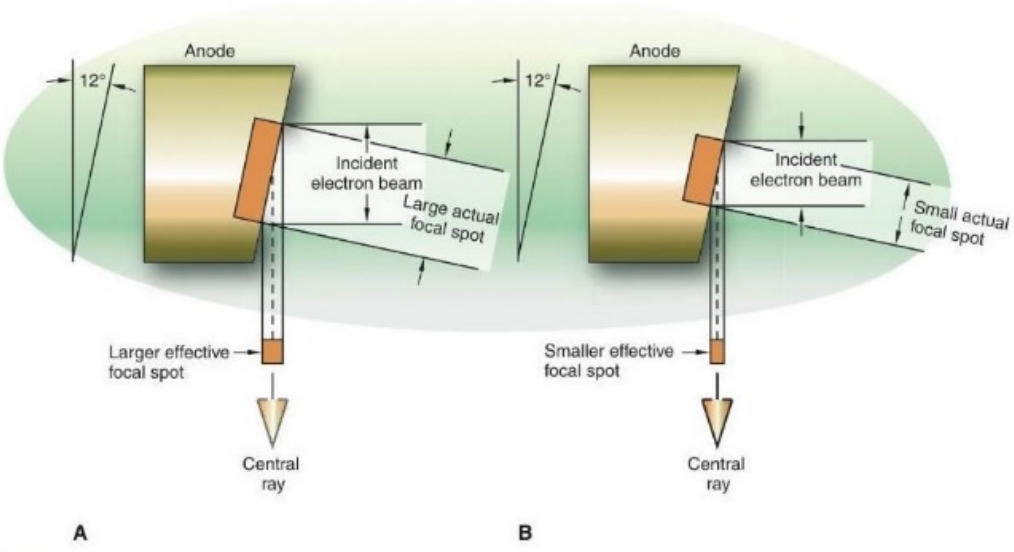

Figure 1–34. Focal spots. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Focal spot (effective)

The focal spot, as it appears from directly beneath the tube at right angles to the electron stream, is called the effective focal spot. The size of the effective focal spot is a very important factor in a diagnostic X-ray tube because it affects the detail on a radiograph—that is, the smaller it is, the better the detail or resolution. X-ray tubes have effective focal spots from 0.1 to 1.5 mm. The size of the effective focal spot is determined by the size of the actual spot and the angle of the target. Notice how in figure 1–34 the target angles are the same, however, the difference in the size of the actual focal spot effects the size of the effective focal spot. Also, notice in figure 1–35 that the electron beams and actual focal spots are the same size, yet when the angle of the surface area changes, so does the size of the effective focal spot. This concept is known as the line focus principle.

Line focus principle

Before development of the rotating anode, another design, known as the line focus principle, was incorporated into X-ray tube targets to allow a large area for heating and maintaining a small focal spot. By angling the target (fig. 1–35), the effective area of the target becomes much smaller than the actual area of the electron interaction. The effective target area, or effective focal spot size, is the area projected onto the patient and image receptor. The smaller the target angle, the smaller the effective focal spot size. Diagnostic X-ray tubes have target angles varying from about 5–15°. The advantage of the line focus principle is it simultaneously provides the sharpness of image of a small focal spot and the heat capacity of a large focal spot.

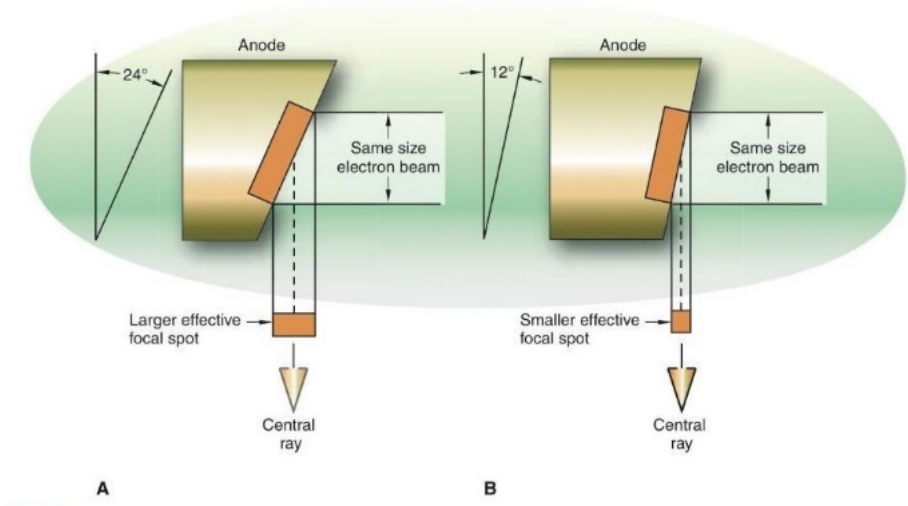

Figure 1–35. Line focus principle. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Anode heel effect

Because the anode is angled, the intensity of the X-ray beam along the longitudinal axis of the tube varies. This variation in intensity results from absorption of some photons by the target. Consider the illustration in figure 1–36, where several photons are given off at a point within the target. Those photons, which make up the anode side of the X-ray beam, stand a greater chance of being absorbed because they travel through more target material than those that make up the cathode side of the beam. (Notice the different distances from the point where the photons are given off to the edge of the target.) Thus, the intensity of the X-ray beam is greater on the cathode side. This non-uniformity is known as the anode heel effect.

The anode heel effect can be used to an advantage when X-raying parts of uneven thickness or density, such as the lower leg. By placing the proximal or nearest end of the leg under the cathode side of the tube, the finished radiograph would have balanced density. A disadvantage of the anode heel effect can be experienced when parts of even density and thickness are X-rayed.

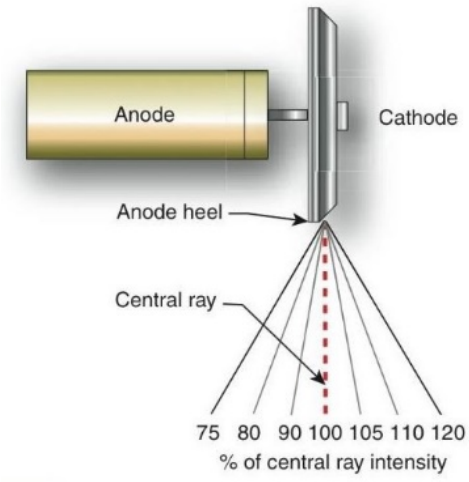

Figure 1–36. Anode heel effect. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

The focal film distance and the area of the X-ray beam to be used must be considered in connection with the anode heel effect. In other words, the anode heel effect can cause an exposure problem at short focal film distances, while at long focal film distances the problem is less likely to exist.

The intensity percentages given in figure 1–36 are designed for a 20° target. As the target angle becomes smaller, the difference in the intensity percentages becomes greater. This further increases the non-uniformity of the X-ray beam. When using tubes with small target angles, extra care is taken to avoid the unbalanced density the anode heel effect can cause on radiographs. The anode heel effect can also be used to a technician's advantage. By positioning the tube with the cathode side toward the thicker portion to be exposed, such as an abdomen, you can take advantage of this greater intensity and not have to increase the kVp or mA for the thickest part being imaged.

X-ray coverage

The target angle also affects the total area of X-ray coverage. As the angle is reduced, so is X-ray coverage (fig. 1–37). The smaller target has less coverage than the larger target at an equal distance from the tube. X-ray coverage usually is of no concern, but at the small target angles, it may interfere with certain examinations.

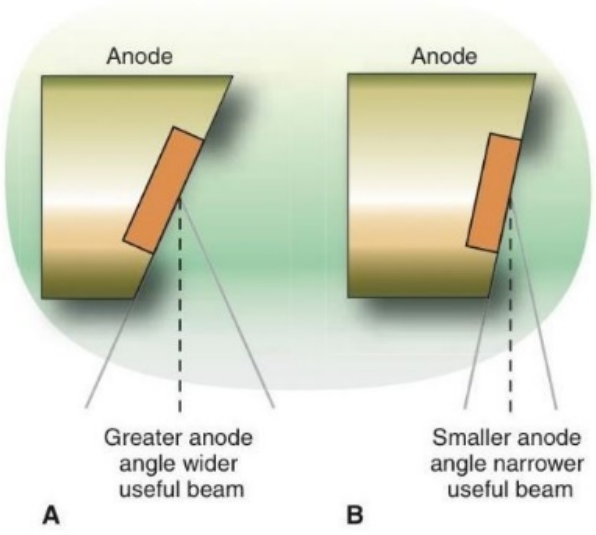

Figure 1–37. Target angle and X-ray coverage. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

For example, a 12° target at the standard 40-inch focal film distance (FFD) covers an area with 8.5-inch radius; a 10° target at 40-inch FFD covers only an area with a 7-inch radius. Clearly, these two targets will not cover a 14 x 17 film at 40 inches—that requires an 11-inch radius. To determine X-ray coverage, the tangent of the target angle (tan) is multiplied by the FFD, which equals the radius of the area covered (RC). The formula is:

$$RC = \text{Tan} \times FFD.$$

If the target angle is 20° (tan is 0.364) and the FFD is 40 inches, the RC, using the formula, is:

$$RC = 0.364 \times 40 = 14.5 \text{ inches.}$$

Rotating anode

We briefly mentioned rotating anodes in our discussion of anode construction. Let's now take a deeper look at the concept of rotating anodes. In low-capacity X-ray tubes, the cooling of the target is achieved by air. In the high-capacity X-ray tubes, required in most modern radiographic techniques, a considerable increase in the capacity of the target to withstand heat is achieved by using a rotating anode. This results in the electronic bombardment of the target over a wide area without altering the size of the focal spot (fig. 1–38).

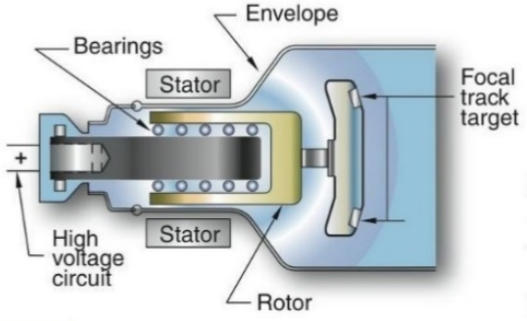

Figure 1–38. Rotating anode. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Many difficulties had to be solved in the development of rotating anodes. For example:

- The high speed of the rotating anode required excellent bearings.
- The heat generated in the tube posed several serious problems because of the low tolerance of certain bearing materials to heat.
- Bearings had to be capable of operating over a wide range of temperature.
- Lubrication of the vacuum enclosed a "hot" rotor, impractical with liquid lubricants; it was solved by the use of metallic lubricants, such as lead, gold, barium, or silver.

In the majority of X-ray examinations, the X-rays are generated by a rotating anode X-ray tube. Stationary anode tubes are mostly used in dental radiography and in low-capacity portable X-ray units.

Heat dissipation

We said the use of rotating anodes would increase the capacity of the target to withstand heat. To see how this is accomplished, refer to figure 1–39 for the following explanation. By comparing the target area of the stationary anode to the target area of the rotating anode, it can be determined that the actual target area for the stationary anode is 1 mm x 4 mm = 4 mm^2; the actual target area of the rotating anode is 2π x 1 mm x 4 mm = 754 mm^2. If the rotating anode has a diameter of 7 cm, then the radius of the target area is approximately 3 cm or 30 mm. Thus, you can see the rotating anode tube provides several hundred times more area for the electron beam to interact than a stationary anode tube.

NOTE: The symbol π stands for Pi, which is shown numerically as 3.14.

Increasing the anode rotation speed can further increase heating capacity. Most standard speed rotating anodes revolve at 3,600 revolutions per minute (rpm). The anodes of high-capacity tubes rotate at approximately 10,000 rpm.

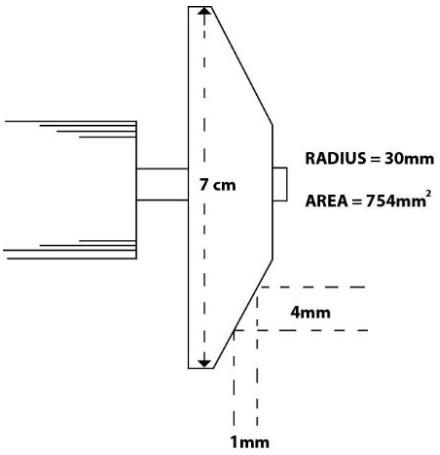

Figure 1–39. Rotor capacity.

With the advent of the rotating anode, the advantage over stationary anodes is a much higher heat capacity, referred to as heat units (HU). This makes possible:

- A considerably greater load in terms of mA per unit time for a given focus size.
- A reduced focal size for a given mA load per unit of time.
- A reduction of time for a given value of mA seconds for a given focus.

Rotation

The rotation of the anode is accomplished by attaching the anode to the rotor of an induction motor contained within the vacuum of the X-ray tube. Let's quickly review how alternating-current induction motors work. They utilize a rotor coil with the exterior magnetic field supplied by several pairs of electromagnets, thus producing a strong magnetic field, increasing the power of the motor and permit- ting it to run at any desired speed. Instead of supplying current to the coil in the magnetic field, an induction motor uses a device called a rotor. A rotor consists of bars of copper around an iron core (fig. 1–40). No commutator or slip rings are required. Instead, a device called a stator is used. A stator consists of pairs of stationary magnets (or, more commonly, electromagnets) arranged around the rotor. The stator electromagnets are energized in sequence. As the copper bars of the rotor reach a point where the motor principle forces are equalized, the multiphase current activates the next pair of electromagnets and the motor principle forces pull the rotor around to the next position. This continuing sequential energization causes the motor principle forces to continually turn the rotor.

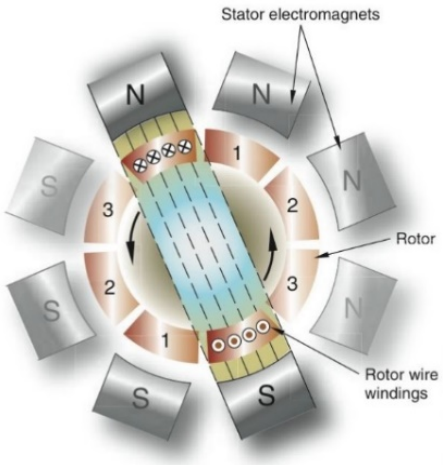

Figure 1–40. Rotor construction and stator design. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Because magnetic lines of force will pass through glass and a vacuum, the electromagnets, or stator, can be positioned outside the vacuum tube, thus avoiding interference with the high voltages required to produce X-rays. The speed of anode rotation may be as high as or higher than 10,000 rpm for special-purpose X-ray tubes. Standard speed is controlled by 60Hz power, while the high-speed rotor operates on 180Hz power.

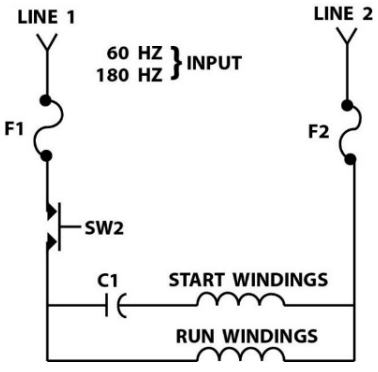

Figure 1–41. Rotor control circuitry.

The anode rotation is basically the same as in a capacitance start induction motor. An electrical current is induced into the starter field windings, which creates a torque and causes the anode stator to rotate. In figure 1–41, when switch 2 is closed, current flows from line 1 to C1 and to the start windings. The capacitive reactance (Xc) and the inductive reactance (X_L) are determined by the value of the capacitor, inductor, and input frequency. The capacitor and inductor are selected to provide a phase shift of 45°. The run winding provides an additional 45°-phase shift, making a total phase shift of 90° between the start and run windings.

The effect is the start and run windings act like a two-phase (2ϕ) stator and produce the revolving field required to start the stator. Once the stator has started turning, it will continue to rotate by itself. Since it isn't possible to rotate the anode by hand because it is sealed in a glass vacuum envelope, an electronic field must be used.

By using some simple math, we can determine the speed of the rotor if we know the input line frequency. With a line frequency of 60Hz, the rotor would be turning 60 complete turns per second (the rotor turns one complete revolution for every input cycle). At 60 revolutions per second, the rotor would be turning at 60 x 60, or 3,600 rpm. To increase the rotation speed, the frequency of the power applied to the stator windings must be increased. For example, if 180Hz power is applied to the stator windings, the magnetic fields will rotate 180 times per second—this gives you approximately 10,800 rpm (or about 10,000 rpm with some stator slippage).

When an X-ray tube is used with high-speed anode rotation, it becomes necessary to bring the rotating anode down to a slower speed at the end of an exposure by the use of an electronic brake. If the anode were allowed to coast to a stop, it would slow down through what are called resonant speeds, which can cause a harmonic vibration large enough to shatter the glass envelope that houses the cathode and anode assembly. In addition, the bearings of the rotating anode would be subjected to premature wear and subsequent failure. For these reasons, a DC current, which works as an electrodynamic brake, is induced into the stator windings at the end of each exposure and brings the rotating anode down to a few hundred rpm in a matter of seconds. The anode is then allowed to coast to a stop to help further dissipate heat buildup. Please note that heat is also carried away from the anode by an oil cooling system that surrounds the glass tube, and also a fan assembly attached to the outer casing to circulate ambient air around the enclosed oil casing.

To ensure tube longevity, an X-ray interlock, or protection circuit, is used in conjunction with the stator windings to ensure the rotating anode is spinning prior to the beginning of an exposure. This type of protection device is usually a current-sensing circuit, meaning that if no current is flowing through the stator windings, the tube protection circuit locks out the main control circuit and an exposure cannot be taken.

X-ray tube rating charts

All X-ray tubes can be thought of as a fuse—too much current and irreparable damage will result. X-ray tubes must be operated within the limitations set by the manufacturer. If not, they can be permanently damaged by excessive heat that results when improper warm-up and exposure techniques are used. To prevent this kind of damage to tubes, you and the technicians must be aware of tube rating charts and how to apply them in the daily operation of the X-ray unit. With each tube, the manufacturer supplies a series of rating charts to be used only with that particular tube.

Heat unit calculations

The amount of heat generated within a tube depends upon the electrical energy in the tube and is measured in heat units (HU), which is the product of the formula:

$$\text{HU} = \text{kVp} \times \text{mA} \times \text{time}.$$

Let's determine how many HU would be generated in a tube given the following exposure factors: 75 kVp, 100 mA at 0.5 second:

$$\text{HU} = 75\text{kVp} \times 100\text{mA} \times 0.5\,\text{seconds} = 3{,}750.$$

To determine HU in 3ϕ equipment, we use two equations:

$$\text{For 6-pulse, HU} = \text{kVp} \times \text{mA} \times \text{time} \times 1.35.$$

$$\text{For 12-pulse, HU} = \text{kVp} \times \text{mA} \times \text{time} \times 1.41.$$

Let's do some calculations using these formulas to familiarize you with this concept. Determine how many HU you would build up with a 6-pulse system using the following exposure factors: 75 kVp, 100 mA at 0.5 second.

$$HU = 75kVp \times 100mA \times 0.5\,second \times 1.35 = 5{,}063.$$

As you can see from these calculations, a 3ϕ, 6-pulse generator delivers approximately 35 percent more power than a 1ϕ unit.

Now, determine how many heat units you would build up with a 12-pulse system using the same exposure factors for comparison: 75 kVp, 100 mA at 0.5 second.

$$HU = 75kVp \times 100mA \times 0.5\,second \times 1.41 = 5{,}288.$$

As you can see by comparing the three power systems, 3ϕ, 12-pulse generators deliver the most power at approximately 41 percent more than 1ϕ and 6 percent more than 3ϕ, 6-pulse systems, but at a higher HU per exposure.

These factors are added in because the rectification is better in a 3ϕ, 6-pulse system than a 1ϕ system. Since the peak kVp is higher, therefore, more HU are generated. It's the same story with 3ϕ, 12-pulse—it is even more efficient and has a better peak voltage than the other two; therefore, the larger factor must be figured in because more HU are generated.

Rating chart selection

It is important to use the proper chart when applying tube-rating charts to determine HU build-up. For example, if the X-ray unit has full-wave rectification, the rating chart for full-wave rectification must be used, since tube capacities vary with the type of rectification used. We have already seen tube capacities also vary with 1ϕ and 3ϕ generators. Generally, factors, such as those we mentioned, need to be determined only once and remain constant throughout the use of the equipment unless a major modification changes the X-ray tube type itself. One factor that can vary from exposure to exposure, and has a definite effect on tube capacity, is focal spot size. You and the radiology department must know the size of the focal spot before appropriate HU calculations can be determined.

Radiographic rating chart

This chart shows the maximum exposure factors allowable for a single radiographic exposure. Figure 1–42 illustrates two such charts. One chart is for a 1 mm (small focal spot) and the other is for a 2 mm (large focal spot). The factors to be considered on these charts are mA, kVp, and exposure time—when any two of the factors are known, the third can easily be found. For example, with the large focus using 150 mA on the left margin on chart B, follow the line across the chart until it intersects with the 90kVp line. From that point, follow an imaginary line straight down to the exposure time scale. The maximum exposure time in this example would be 4.5 seconds. Anything above that would exceed the tube's capacity and could damage the tube. Using the same procedure on the small focal spot chart A, you can see for the same 150 mA at 90kVp, you would be allowed to make a 0.75 second maximum exposure. You can see by using this comparison of the small focal spot tube versus the large that the load capacity of the small is considerably less than the large focal spot tube.

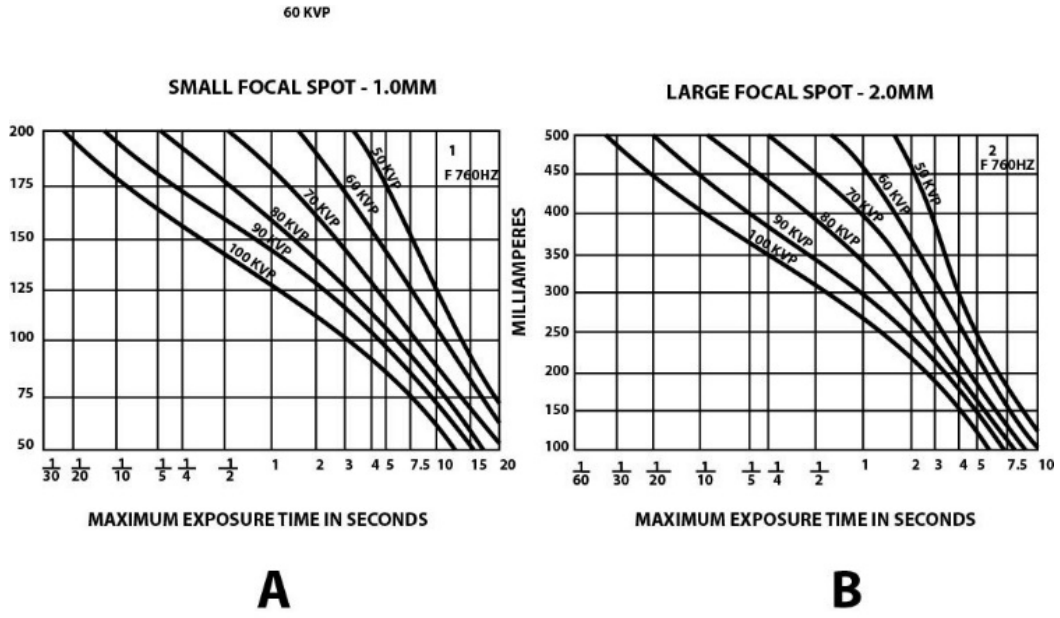

Figure 1–42. Radiographic tube rating chart.

Anode cooling chart

The anode cooling chart reflects the maximum number of HU the anode can store and the rate at which it can dissipate the heat. Figure 1–43 illustrates an anode cooling chart with a storage capacity of 175,000 HU. If a series of exposures totaling 100,000 HU was made over a period of a few seconds, and it was necessary to repeat the series, how much time, if any, would have to be allowed in between? Since the anode in this particular tube can only be subjected to 175,000 HU, and 100,000 HU were introduced in the first series, an additional 100,000 HU series without delay would mean 200,000 HU to the anode, or 25,000 HU more than the maximum allowable. Before exposing the second series, the tube must be allowed to cool to 75,000 HU. By following the cooling curve, it can be seen that a drop from 100,000 HU would take approximately 1.5 minutes. The second series can then be repeated after 1.5 minutes without exceeding the anode storage capacity. However, since you are now at maximum storage capacity, before this series can be repeated a third time, it would be necessary to allow approximately 4.25 minutes cooling time to dissipate 100,000 HU.

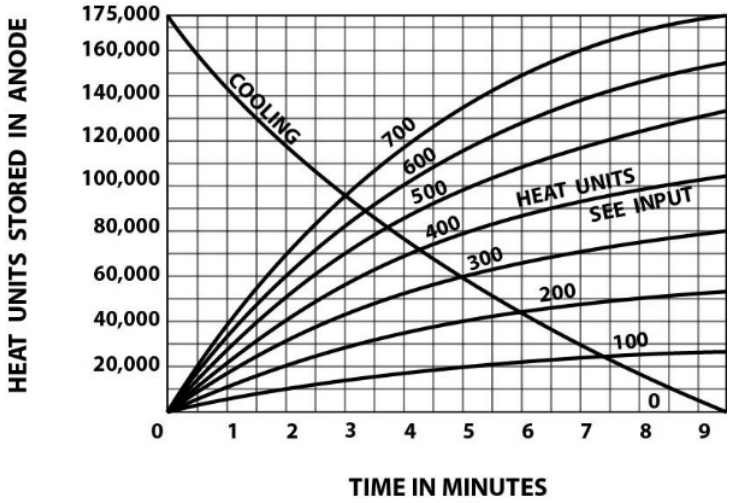

Figure 1–43. Anode cooling chart.

Housing cooling chart

As with the rate of heat dissipation, the storage capacity of the tube housing is also limited. Figure 1–44 illustrates the heat capacity of the tube housing is considerably greater than the anode capacity, while the heat dissipation rate is less. The housing cooling chart is basically used in the same manner as the anode cooling chart. For example, suppose 50,000 HU were added to the tube every minute. You can see from the chart in figure 1–44 that the tube housing can't dissipate 50,000 HU every minute, even with the air circulator, as seen on the cooling curve. Therefore, the tube housing would eventually reach its maximum loading capacity, and further cooling time between series would need to be allowed. In actual practice, the anode will usually reach its limits long before the housing. However, there are instances, as when the forced-air fan for the housing is not functioning, when these charts can be useful.

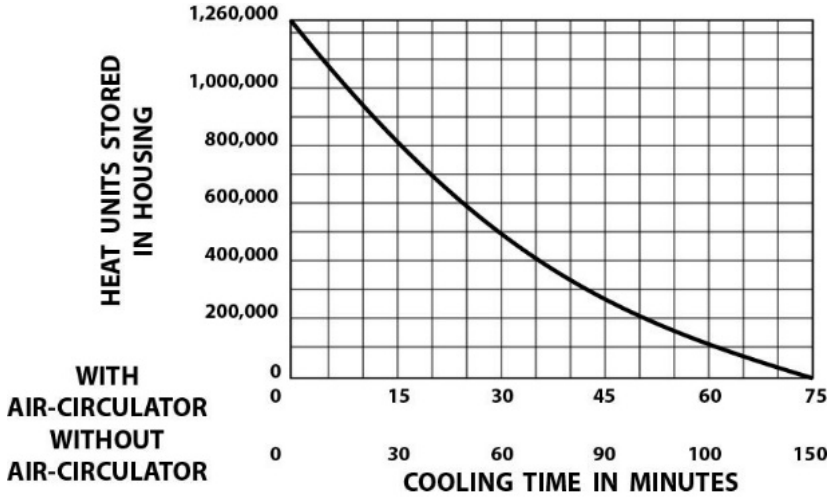

Figure 1–44. Housing cooling chart.

607. Digital radiography

In simple terms, digital radiography differs from conventional radiography as digital radiography does not use film; rather, it uses other types of radiation detectors. These detectors have an electrical output proportional to the radiation intensity. The initial output signal may be analog; if so, it is then converted to digital form, processed by a computer, and then displayed on a video monitor. As the Air Force has essentially phased out X-ray film as a radiographic image capture medium, we will conclude the X-ray process using digital radiography as our exposure receptor. For the purpose of the lessons to follow, we may continue to use the term "film" as a general image capture concept.

Take a moment to think about some of the advantages of a digital image over conventional film. One of the advantages is a digital image likely saves many dollars over the long run because we eliminate the cost for film processing (as well as no more repairs on the film processor!). In addition, the hassles of dealing with the chemicals used in film processing are eliminated. Another benefit is a digital image takes up much less space (although it may take up a lot of computer memory, its physical dimensions are small) than a traditional image. One last advantage is digital images can be transferred between MTFs or other medical centers with relative ease, compared to mailing or hand-carrying a traditional image.

Hypothetically, an "ideal" digital radiography system would have the following characteristics:

- Image quality as good as, or better than, the best traditional X-ray system.
- Excellent image resolution.
- Low patient dose.

- High durability and low maintenance (especially compared to traditional radiographic systems).
- Easily connects to the hospital information system (HIS).

Of course, one major consideration of these systems is the cost—they are not cheap! Then again, medical equipment and cheap are rarely used in the same sentence! However, as technology improved, the benefits of a digital system began to far outweigh those of a traditional system making them much more of a commonplace.

Digital X-ray imaging can be divided into two general classes: computed radiography (CR) and digital radiography (DR). Let's explore each.

Computed radiography

CR, also known as a cassette system, was the first type of digital radiography to be introduced. While it cannot be considered a truly digital system, it is a step in that direction and a step away from the traditional radiological system. CR imaging uses a photostimulable, phosphor-based storage imaging plate to replace the standard image intensifier screen and X-ray film. This plate has a phosphor layer doped with special chemicals that is energized when exposed to X-rays, causing a latent image to be temporarily formed. This image is scanned with a CR plate reader, which brings the image out, and is then viewed with an extremely sensitive photomultiplier tube. The image is then digitized and stored for display on a computer monitor or printed out on a laser imager. A bright light is used to erase the imaging plate, which can then be reused.

The image plates, or CR plates, used with this type of system are similar to traditional film cassettes and are available in the same cassette sizes. In fact, these CR plates can often be used in traditional X-ray systems with only minor modifications to the system. The workflow is also very similar to a traditional system: the X-ray is taken; the CR plate is removed and placed into an image plate reader; the reader extracts the image from the plate; and the image is presented. However, this is where the major difference appears. In a traditional system, the image is printed on X-ray film, but, when using CR plates, the image can be printed with a laser imager or sent to a computer workstation to be viewed on a monitor.

The CR imaging plates work on the principle of photostimulable luminescence. This basically means when the CR plate is exposed to ionizing radiation, it stores the image as energy within its phosphor layer. This energy is then released in the form of light when the CR plate is stimulated by a light source (within the CR plate reader). The light within the plate reader (usually infrared or red laser) has a longer wavelength than the characteristic emissions of phosphor to activate the process of luminescence. The luminescence given off from the CR plate is detected by an image detector and converted into a digital signal.

The imaging plate (fig. 1–45) consists of several layers and is located within a cassette, just like a traditional X-ray cassette. Within the imaging plate, the protective layer is used to protect the phosphor layer during handling. The most important layer is the phosphor layer, which traps the latent image. The reflective layer consists of light reflective materials to increase the effectiveness of the image capture process. The conductive layer is made up of conductive crystals in some type of binder and is used to reduce problems caused by electrostatic charges, thus increasing the sharpness of the image. The support layer is a polyester base similar to that of a standard image-intensifying screen, and helps to strengthen and hold together the imaging plate. The light shielding layer is used to prevent light leakage into the plate from the backside of the plate. Last, the backing is made from a soft polymer and makes handling the CR plate easier, and also helps protect the plate. There is usually a barcode label on the backing, which provides a serial number for tracking patient data.

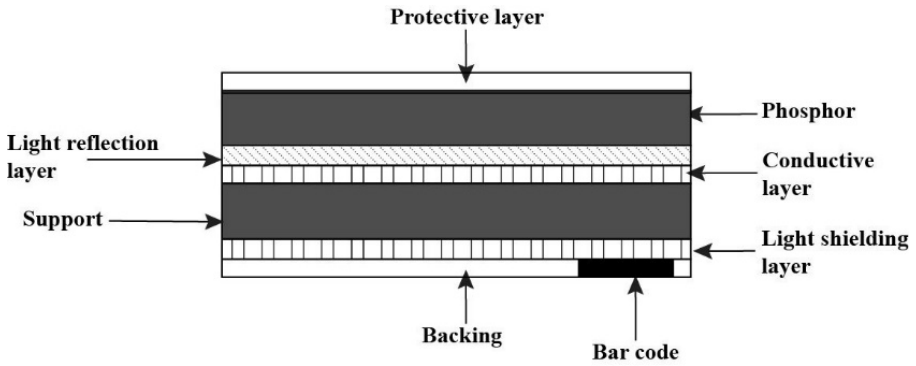

Figure 1–45. Typical CR plate construction.

Digital radiography

DR, also known as a cassette-less system, is more of a truly digital system. DR systems typically have the detector and reader that are a permanent part of a table or wall unit; therefore, a cassette is not needed. Newer technology has now made some of these systems wireless and they can resemble a sealed cassette that can be moved around a room from a table bucky to an upright unit. With these systems, great care must be taken in moving the wireless DR unit as it is a very expensive device. With DR systems, the image is acquired and sent directly to the display monitor without the need for the radiographer to physically move the detector for the image to be processed. This makes these systems faster and more efficient at creating an image. DR can be divided into two broad classes: charge-coupled devices (CCD) and flat panel detectors. CCD detectors were the first type of DR detector on the market, so we will start our discussion with them.

Charge-coupled device detector systems

A CCD detector system is composed of an optical detector chip, which contains millions of independent picture elements (pixels) arranged in an array pattern. When incoming X-rays strike the detector, a charge is induced, thus illuminating the pixels in a pattern representing the X-ray image. This pattern is then read out pixel-by-pixel, digitized, and transferred to computer memory. The image information is then converted into visible light by an X-ray scintillator (a scintillator is a material that emits light when particles, in this case X-rays, strike it). The visible light is then transmitted to a CCD detector similar to those used in video cameras. Finally, the CCD detector transforms the visible light into a digital image.

One problem a CCD detector system presents is the CCD detector is physically smaller than the image area. To compensate for this, light emitted from the scintillator must be reduced in size before it strikes the CCD—normally a demagnifying system made up of lenses or fiber optics handles the task. Because the demagnifying system reduces the size of the field that strikes the CCD detector, the image quality is degraded to some extent; therefore, this is not the ideal detector system.

Flat panel detector systems

The second type of detector system in use is the thin-film, flat panel type. The advantage these types of detectors have over the CCD detectors is their detector surface areas are the same size as the image being detected; therefore, no demagnification takes place. Most flat panel detector systems employ a thin-film transistor (TFT) array. This type of detector is fabricated like an “electronic sandwich” by implanting the electronics in a glass substrate of several layers. Normally, the readout electronics are situated at the bottom and charge-collecting capacitors are placed in the middle of the sandwich. Depending on the type of detector, the top of the array is made up of X-ray or light sensitive materials, or a combination of both. Detectors using a scintillator and light-sensitive photodiode are called indirect conversion TFT detectors because the X-rays are first converted to light by the scintillator, which is then converted to an electrical signal by the photodiode. Detectors using only

X-ray sensitive photoconductors are called direct-conversion TFT detectors because the step of converting X-rays to light is eliminated. Figure 1–46 is a graphic representation of the three DR detector systems covered in this lesson.

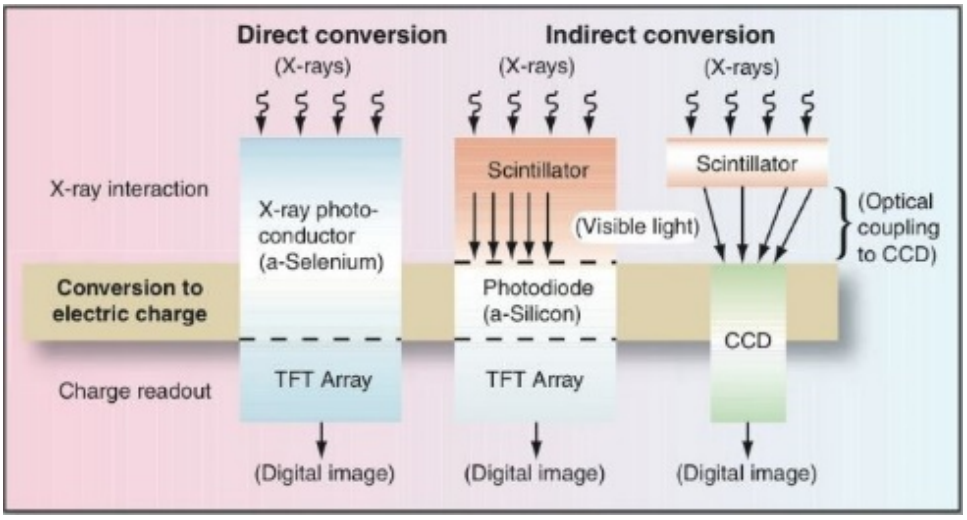

Figure 1–46. Comparison of CCD, indirect and direct flat panel detector systems. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Refer to figure 1–47 for the remainder of this discussion. Whether a detector is direct or indirect, the basic principle of operation after the incoming X-rays are converted to an electronic signal remains the same.

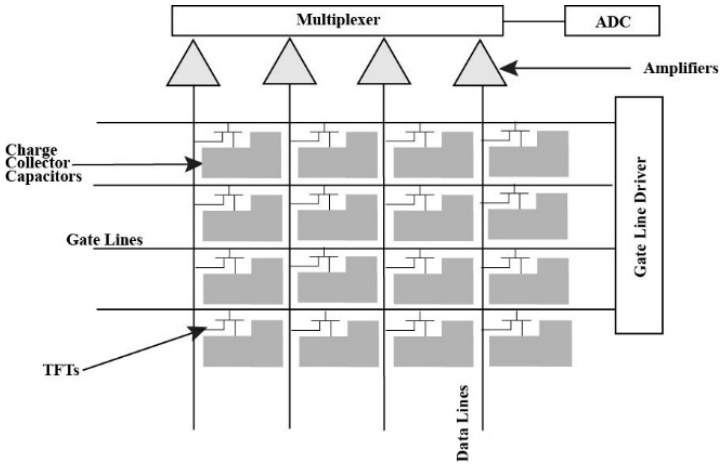

Figure 1–47. Construction of a flat panel detector system.

The charge-collector capacitor drains the charge generated by the photodiode or photoconductor. Each individual charge-collector capacitor and TFT represents a pixel of an X-ray image; the charge collected on each is proportional to the X-rays received by the detector. During the readout process, these charges are read out one row at a time with the gate lines controlling the readout. When the gate line voltage is set high, all transistors of the row conduct the pixel charges collected since the last readout cycle to the adjacent data lines. Charge amplifiers and a multiplexer integrated in the detector then read out the signal for that row. This means every data line transports the information for one pixel during the readout of one row, and there is one data line for every line of the detector.

Indirect detector systems

The main difference between the two systems lies on the top of the detector. The indirect system uses a scintillator, reflective layer, and graphite protective coating (fig. 1–48). The scintillator contains a layer of thin, rod-shaped cesium-iodide (CsI) crystals, which produce light when an X-ray is absorbed. The columns of CsI reduce the light spread interacting with the amorphous silicon. The light is then reflected onto the TFT array, and the photodiodes in the array produce an electrical signal.

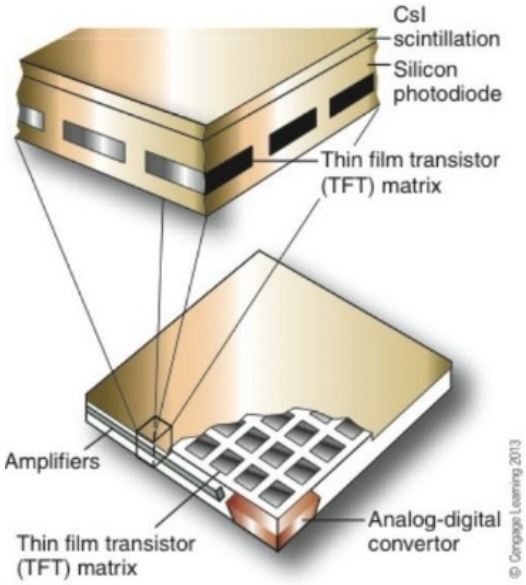

Figure 1–48. Indirect amorphous silicon flat panel imaging plate system. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Direct detector systems

As we stated earlier, this type of detector system eliminates the intermediate step of converting X-ray energy into light—the X-ray energy is converted directly into an electrical signal. Rather than detecting light, the TFT array detects an electrical signal. The high-voltage charge at the top surface of the amorphous selenium layer results in the ionization caused by the X-ray photons to free electrons for collection by the electrodes at the bottom of the selenium layer (fig. 1–49). The selenium layer directly converts X-rays to an electrical charge. The charge is collected and then transmitted through thin film detectors to the computer for processing.

Unlike the indirect detector system, the direct system does not use phosphors or a scintillator, but employs a direct conversion process. Because of this, the direct method generally produces a higher quality image.

DR is an ever-evolving field and there are always new technologies on the horizon. Manufacturers realize this is the future of X-ray imaging; therefore, you can look for new types of digital imaging technologies to be released and adopted as acceptable replacements for traditional X-ray imaging techniques.

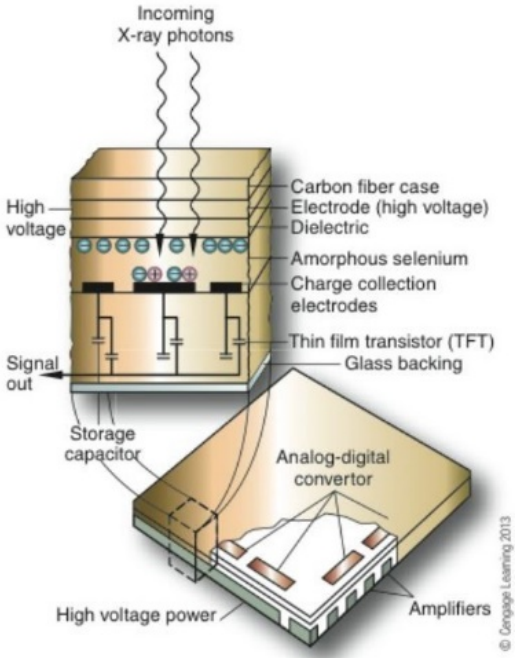

Figure 1–49. Direct conversion amorphous selenium flat panel imaging plate. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Self-Test Questions

After you complete these questions, you may check your answers at the end of the unit.

603. X-ray control circuit components

1. Identify the lettered areas in figure 1–50, which depicts a basic X-ray generator.

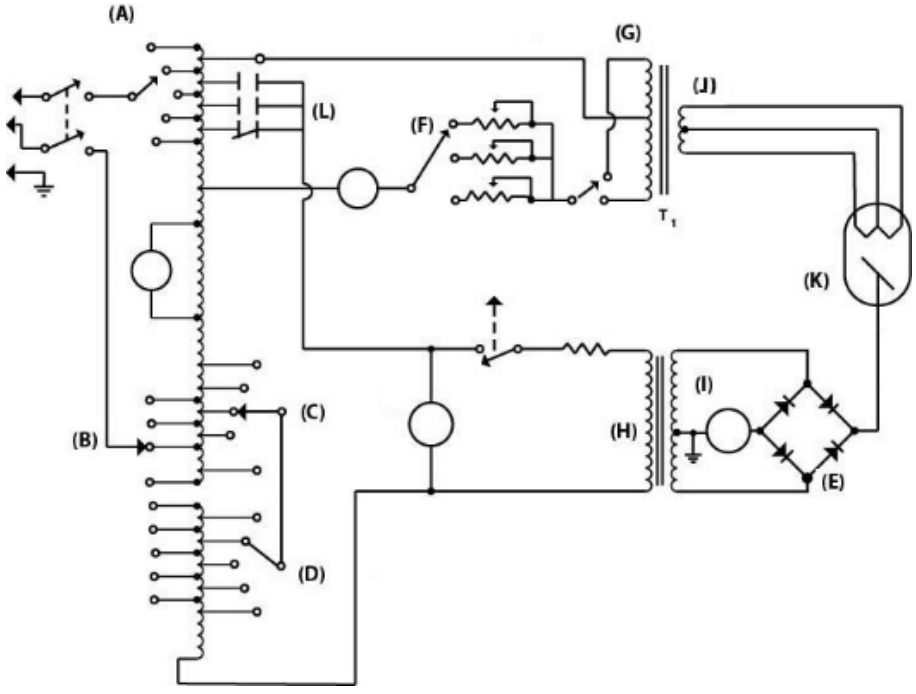

Figure 1–50. Basic X-ray generator.

604. High-voltage rectification

1. What are rectifiers used for in X-ray?

2. What is peak inverse voltage?

3. What do they call the value of the average DC current that can be continuously passed through the rectifier?

4. What will overheating caused by excessive current flow do to a rectifier?

5. List the three groups of 1ϕ rectifiers that were discussed.

6. The efficiency of the 1ϕ, half-wave rectifier limits its use to what type of X-ray systems?

7. What are two advantages of the bridge rectifier over the full-wave rectifier?

8. Name the four configurations used to raise voltage in a 3ϕ system.

9. What does the AC output of the wye-wye and delta-delta configuration depend upon?

10. What formula is used to determine output voltage in a delta-wye configuration?

11. What is the biggest advantage of high frequency X-ray units over traditional X-ray units?

12. In a falling load type of X-ray generator, what causes a shorter tube life?

605. Timing devices

1. Why are timing circuits important?

2. What determines the number of X-rays emitted and, therefore, the density on the film in a mAs timer?

3. What was developed to improve the quality and consistency of developed films?

4. List three things that can happen in an AEC when light strikes a special combination of metals.

5. What are the two types of AEC systems?

6. Explain the proportionality of current to light intensity in a phototimer.

7. Where is the ion chamber located in the timing circuit?

8. What factor determines the minimum reaction time of an AEC?

9. What percent of the anticipated manual exposure mAs should you set the backup timer to?

606. X-ray tubes

1. What percent of the energy is actually used in the production of X-rays?

2. Why are X-ray tubes sealed in a vacuum?

3. What is the name of the electrode at which the electrons originate in the tube?

4. State the name of the electrode in which electrons are arrested or bombarded.

5. What is applied to the anode and cathode of an X-ray tube, which affects the penetrating power and intensity of the resulting X-ray?

6. What is the function of the cathode in an X-ray tube?

7. Explain the purpose of the focusing cup.

8. Identify the metals used as target material for the X-ray tube.

9. What two methods are used for heat reduction of an anode?

10. Name the two types of anodes used in X-ray tubes.

11. What is affected when you change the effective focal spot size?

12. When talking about the anode heel effect, which side of the tube has the greater intensity? Why?

13. How is further heat dissipation accomplished in high-capacity X-ray tubes?

14. What are three advantages of a rotating anode over a stationary anode?

15. How is the rotation of the anode accomplished?

16. What is the determining factor in the rpm of a rotating anode?

17. State the reasons for using a braking system on a modern rotating anode X-ray tube.

18. How many HU would be generated in a 3ϕ, 6-pulse system with the following factors: 100kVp, 100mA at 0.75 second?

19. What is the main purpose of an anode cooling chart?

607. Digital radiography

1. List three advantages of DR over conventional radiography.

2. What characteristics would an "ideal" DR system have?

3. What are the two general classes of digital X-ray imaging?

4. Match the statements in column A with the appropriate type of radiography in column B. Items in column B are used more than once.

Column A	*Column B*
____ (1) A truly digital system.	a. CR.
____ (2) Workflow is similar to conventional radiography.	b. DR.
____ (3) May have a CCD detector system.	
____ (4) The first type of DR to be introduced.	
____ (5) Traditional systems can be converted to with only minor modifications.	
____ (6) May have a flat-panel detector system.	

5. What principle do the imaging plates within a CR system work on?

6. List the layers and the purpose for each within a CR imaging plate.

7. What are the two broad classes of DR X-ray systems?

8. What is a scintillator?

9. What is the one problem associated with CCD detector systems?

10. What are the two types of flat panel detector systems?

11. Why is the direct detector system known as a “direct system?”

Answers to Self-Test Questions

601

1. (1) e.
   (2) d.
   (3) b.
   (4) c.
   (5) a.
2. Bremsstrahlung.
3. Characteristic.
4. Any process that results in the removal or addition of an orbital electron from or to an atom or molecule, thereby leaving the atom or molecule with an overall positive or negative charge.
5. Photoelectric effect.
6. Compton Effect.

602

1. (1) Chemical.
   (2) Electrical.
2. Radiation absorbed dose.
3. The amount of radiation striking the chamber.
4. The manner in which the electrons are freed from their traps.
5. Limit, control, or modify the electromagnetic output.
6. High absorber density.

603

1. (A) Line compensator.
   (B) Line strap adjustment.
   (C) Major kVp selector.
   (D) Minor kVp selector.
   (E) Full wave rectifier.
   (F) mA selector.
   (G) Primary filament transformer.
   (H) PHT.
   (I) SHT.
   (J) Secondary filament transformer.
   (K) X-ray tube.
   (L) Load compensation.

604

1. To change AC to pulsating DC.
2. The maximum potential that can be applied across the rectifier without breaking down.
3. DC current rating.

4. Will destroy the p/n junctions and render the device useless.
5. (1) Half-wave.
   (2) Full-wave.
   (3) Bridge.
6. X-ray equipment that requires a low current drain, such as a dental X-ray unit.
7. (1) The bridge circuit produces a voltage output nearly twice that of the full-wave rectifier.
   (2) The peak inverse voltage across the bridge diodes is half that of the full-wave rectifier.
8. (1) Wye-wye.
   (2) Delta-delta.
   (3) Wye-delta.
   (4) Delta-wye.
9. Strictly upon the Tr of the transformer.
10. Es = Ep x Tr x 1.73.
11. They produce a very stable and nearly ripple-free voltage for the X-ray tube, which enables very efficient and uniform X-ray production throughout the entire exposure.
12. Excess heat produced from operating the tube at maximum mA during the beginning of an exposure.

605

1. Because they reduce the exposure of X-ray dosage to the patient.
2. The product of mA and time.
3. AECs or automatic exposure devices, as they do not require adjustments from the technician and operate on the amount of radiation transmitted through the patient.
4. (1) A voltage may be generated (photovoltaic).
   (2) A change in resistance will occur (photoconductive).
   (3) An electron, or electrons, may be ejected or emitted (photoemitting).
5. (1) Phototimers (which use a photomultiplier tube).
   (2) Ionization chambers.
6. This current is directly proportional to the intensity of light falling on the cathode—the greater the light intensity striking the photodiode, the greater the current flow to the output anode of the photomultiplier.
7. In front of the image receptor so that radiation interacts with the detectors just before interacting with the image receptor.
8. The length of time necessary for the AEC to respond to the radiation and for the generator to terminate the exposure.
9. 150 percent.

606

1. 0.2.
2. Because electrons cannot be sufficiently accelerated or controlled in air or any other gas at atmospheric pressure.
3. Cathode of the tube.
4. Anode of the tube.
5. High kVp.
6. To produce a thermionic cloud, conduct the high voltage to the gap between cathode and anode, and focus the electron stream as it heads for the anode.
7. To condense the electrons into a small beam, because when electrons are boiled off they tend to spread out due to electrostatic repulsion.
8. Tungsten, rhenium-tungsten alloy, and Mo.
9. (1) Embedding the tungsten in copper, since copper has higher heat conductivity than tungsten.
   (2) Rotating the anode.

10. Stationary and rotating.
11. Radiographic detail.
12. Cathode side; more photons are absorbed by the anode side due to the angle of the target.
13. By incorporating a rotating anode in the tube.
14. (1) Greater load in terms of mA per unit of time for a given focus size.
    (2) A reduced focal size for a given mA load per unit of time.
    (3) A reduction of time for a given value of mAs for a given focus.
15. By attaching the anode to the rotor of an induction motor contained within the vacuum of the X-ray tube.
16. Frequency of input power.
17. To prevent harmonic vibration damage to the X-ray tube and to prolong the life of the bearings.
18. 10,125.
19. To determine the maximum number of HU the anode can store and the rate at which it can dissipate.

607

1. (1) A digital image saves money over the long run because film-processing costs are eliminated.
   (2) Film-processing chemicals do not have to be dealt with.
   (3) Images stored digitally save storage space over traditional films.
2. Image quality as good as, or better than, the best tradition X-ray system; excellent image resolution; low patient dose; high durability and low maintenance (especially compared to traditional radiographic systems); and easily connects to the HIS.
3. (1) CR.
   (2) DR.
4. (1) b.
   (2) a.
   (3) b.
   (4) a.
   (5) a.
   (6) b.
5. Photostimulable luminescence.
6. Protective layer is used to protect the phosphor layer during handling; phosphor layer traps the latent image; reflective layer increases the effectiveness of the image capture process; conductive layer reduces problems caused by electrostatic charges, thus increasing the sharpness of the image; support layer strengthens and holds together the CR plate; light shielding layer prevents light from leaking into the plate from the backside; and backing makes handling the plate easier and helps to protect it.
7. (1) CCDs.
   (2) Flat panel detectors.
8. A material that emits light when X-rays strike it.
9. The CCD detector is physically smaller than the image area; this requires the image to be reduced by a demagnifying system, which ultimately degrades image quality.
10. (1) Direct.
    (2) Indirect.
11. Because it directly converts X-rays into an electrical signal and eliminates the intermediate step of converting X-rays to light.

Complete the unit review exercises before going to the next unit.

Unit Review Exercises

Note to Student: Consider all choices carefully, select the *best* answer to each question, and *circle* the corresponding letter. When you have completed all unit review exercises, transfer your answers to the Field-Scoring Answer Sheet.

Do not return your answer sheet to the Air Force Career Development Academy (AFCDA).

1. (601) When a negatively charged electron is deflected from its original course by the attractive force of a nucleus, it is called
   a. line.
   b. deflection.
   c. characteristic.
   d. Bremsstrahlung.

2. (601) What is *not* a factor that affects the energy of Bremsstrahlung radiation?
   a. Charge of the nucleus.
   b. Striking force of the electron.
   c. Original kinetic energy of the electron.
   d. How close the electron comes to the nucleus.

3. (601) What term best describes X-ray energy that disappears in certain materials?
   a. Attenuation.
   b. Absorption.
   c. Contrast.
   d. Density.

4. (601) What is the process that results when an atom or molecule gains or loses an orbital electron , thereby leaving the atom or molecule with an overall negative or positive charge?
   a. Ionization.
   b. Attenuation.
   c. Microwave.
   d. Photoelectric.

5. (601) What is the all-or-nothing energy exchange in a photon where the photon imparts all of its energy to an electron and simply vanishes?
   a. Photoelectric effect.
   b. Vanishing effect.
   c. Compton effect.
   d. Flash effect.

6. (602) What two methods can be used to detect the ionization effect of X-rays?
   a. Electrically and by damage incurred.
   b. Chemically and electrically.
   c. Chemically and visually.
   d. Electrically and visually.

7. (602) The output of X-ray machines is specified in
   a. curies.
   b. roentgens.
   c. radiation absorbed doses.
   d. radiation equivalent man.

8. (602) What is the energy of an X-ray measured in?
   a. Curies.
   b. Roentgens.
   c. Electron volts.
   d. Radiation absorbed doses.

9. (602) What does an ion chamber measure?
   a. Electron volts.
   b. Speed of X-rays.
   c. Distance of source.
   d. Radiation dose rate.

10. (602) Which section is responsible for the personnel dosimetry program in your facility?
   a. Radiology.
   b. Public health.
   c. Medical maintenance.
   d. Bioenvironmental engineering.

11. (602) How often must thermoluminescence dosimeters be exchanged?
   a. Monthly.
   b. Every two months.
   c. Every three months.
   d. Every six months.

12. (602) Which is *not* an advantage of using optically stimulated luminescence (OSL) dosimeters over thermoluminescence dosimeters (TLD)?
   a. Less expensive.
   b. Can be reread multiple times.
   c. Can be read at room temperature.
   d. Signal loss less than one percent in a second reading.

13. (602) What is used in the radiation exposure badge storage area to record background radiation and temperature variations?
   a. Nothing.
   b. Radiation probe.
   c. Control film badge.
   d. Radiation detector and thermometer.

14. (602) What is *not* a common cause for overexposure of the personnel film badge?
   a. Failure to use protective shielding.
   b. Improper work habits.
   c. Exposure to sunlight.
   d. Improper storage.

15. (602) What is *not* a way to reduce radiation exposure?
   a. Time.
   b. Distance.
   c. Shielding.
   d. Personnel dosimetry.

16. (603) What type of *input* power do most X-ray units operate on?
   a. Single-phase (1ϕ) or high frequency.
   b. Three-phase (3ϕ) or high frequency.
   c. 1ϕ or 3ϕ.
   d. 3ϕ only.

17. (603) The quantity of electrons emitted by the filament of an X-ray tube is controlled by the
    a. kilovolt peak setting.
    b. milliampere setting.
    c. X-ray tube size.
    d. time setting.

18. (603) Which X-ray unit circuit controls kilovolt peak (kVp) throughout varying loads?
    a. Milliampere control.
    b. Load compensation.
    c. Line compensation.
    d. kVp control.

19. (603) What X-ray unit circuit keeps tube current constant throughout varying tube voltages?
    a. Space charge compensation.
    b. Milliampere control.
    c. Load compensation.
    d. Line compensation.

20. (604) What are rectifier devices used for in X-ray equipment?
    a. Changing alternating current to pulsating direct current.
    b. Stepping up transformer voltages.
    c. Voltage limiting.
    d. Voltage steering.

21. (604) What term describes the *maximum* potential that can be applied across a rectifier without it breaking down?
    a. Peak current.
    b. Peak-inverse voltage.
    c. Average direct current.
    d. Maximum breakover voltage.

22. (604) What is *not* a single-phase rectifier?
    a. Bridge.
    b. Full-wave.
    c. Half-wave.
    d. Delta-wye.

23. (604) The characteristic of high-frequency generators that allows them to provide better kilovolt peak (kVp) regulation than standard X-ray generators is they
    a. use smaller components.
    b. use no alternating current power.
    c. can operate on single- or three-phase power.
    d. use filtered direct current power to supply the kVp.

24. (605) According to Code of Federal Regulations (CFR) 21, X-ray generators must terminate the exposure at
    a. 600 mAs for exposures above 50 kVp and 2,000 mAs for exposures below 50 kVp.
    b. 600 mAs for exposures above 90 kVp and 2,000 mAs for exposures below 90 kVp.
    c. 300 mAs for exposures above 50 kVp and 1,000 mAs for exposures below 50 kVp.
    d. 300 mAs for exposures above 90 kVp and 1,000 mAs for exposures below 90 kVp.

25. (606) How much of the electron action within an X-ray tube is converted to heat?
    a. 0.2 percent.
    b. 40.0 percent.
    c. 60.9 percent.
    d. 99.8 percent.

26. (606) What material is *most* often used for X-ray tube targets?
    a. Molybdenum.
    b. Tungsten.
    c. Rhenium.
    d. Copper.

27. (606) What factor affects the size of the X-ray unit's focal spot?
    a. Distance of the focusing cup from the anode.
    b. Angle of the target surface.
    c. Position of the filaments.
    d. Charge on the anode.

28. (606) What term describes the X-ray unit's focal spot as it appears from directly beneath the tube at right angles to the electron stream?
    a. Effective focal spot.
    b. Actual focal spot.
    c. Focal track.
    d. Focal zone.

29. (606) Using these target angles, which provides the *smallest effective* focal spot for an X-ray unit?
    a. 5°.
    b. 10°.
    c. 15°.
    d. 20°.

30. (606) What is the speed of an X-ray unit's *standard* rotating anode?
    a. 2,000 revolutions per minute (rpm).
    b. 3,600 rpm.
    c. 10,000 rpm.
    d. 10,800 rpm.

31. (606) What slows an X-ray unit's high-speed rotating anode down through the harmonic range *without* shattering the glass tube?
    a. Direct current.
    b. Alternating current.
    c. Oil sealed bearings.
    d. Frictionless bearings.

32. (606) Calculate the heat units (HU) for a three-phase, 12-pulse generator with a technique of 100 kilovolts peak (KvP) and 600 milliamperes (mA) for 0.01 seconds.
    a. 600 HU.
    b. 810 HU.
    c. 846 HU.
    d. 60,000 HU.

33. (606) Which X-ray tube rating chart is used to determine the *maximum* number of heat units (HU) that can be stored for a series of exposures?
    a. Radiographic tube rating chart.
    b. Housing cooling chart.
    c. Anode cooling chart.
    d. HU cooling chart.

34. (607) With only minor system modifications, a traditional X-ray system can be converted to a
    a. computed radiography system.
    b. daylight radiography system.
    c. network radiography system.
    d. digital radiography system.

Please read the unit menu for unit 2 and continue ➔

Student Notes

Unit 2. X-ray Systems

THE use of X-rays is the most satisfactory method for discovering many conditions that might not otherwise be observed until it's too late for treatment. It is less invasive than exploratory surgery and essentially eliminates the recovery time. This is all possible because of an X-ray's ability to pass through tissues and create exposures, allowing us to view a person's anatomy through images.

X-ray and related equipment will be some of the most important, as well as time-consuming, apparatus you will work on in your career. Because of this, we will take the time to explore some of the more important aspects of X-ray equipment. We will start by looking at fixed radiologic and fluoroscopic systems, and then cover various other standard X-ray systems. Next, we will cover some of the support equipment related to X-ray systems and end by discussing information related to equipment maintenance.

2–1. Fixed Systems

As an aid in diagnosis, X-rays provide two different methods of examination. One method is known as radiography, a permanent record of the X-ray image of various parts of the body made digitally or on sensitive film. The other method is known as fluoroscopy, the visual study of the X-ray shadow as it appears on a fluorescent screen, which enables the radiologist to observe internal motion of the organs. We will spend some time in the next few lessons on the fixed (non-mobile) versions of these systems. Let's start by discussing a typical radiologic system and its various uses.

608. Typical radiologic systems

The most common type of X-ray unit you will see throughout an MTF is the fixed radiologic system. Because of this, we will start our journey through X-ray systems here. Let's start with clinical applications.

Clinical applications

Standard radiography presents an image on a sensitive film. An X-ray film, or radiograph, of any part of the body is similar to a photograph, except it is fundamentally a shadowgraph rather than a picture produced by reflected light. The source of light is an X-ray tube instead of an ordinary lighting apparatus. The "blacks and whites" of the film depend on the different densities of the tissues situated

in the path of the rays. The amount of X-ray that reaches the film after passing through the patient determines the amount of exposure to the film. The greater the density of the object radiographed, the lighter that particular area appears on the film.

The radiograph, therefore, is not a picture, but a record of shadows of different body parts that vary according to their respective densities and absorption rates. It is from a highly skilled interpretation of these shadow images that the radiologist makes a diagnosis. We use these images to diagnose injuries or disease conditions, or to recognize abnormalities that lie within the body. Of course, bones are easily visible using traditional X-rays; therefore, we often use X-rays for the detection of bone-related problems, such as fractures, arthritis, and other bony abnormalities. A chest radiograph series is a key study to check for pneumonia in the lungs or other problems in the chest. While these are two of the most common uses of a standard radiographic system, there are also many other uses. We will not attempt to make you an expert on every possible technique; hopefully by now, you have spent some time in your X-ray department and are aware of the numerous applications of X-ray systems. Most modern X-ray systems consist of many interrelated components.

Components

We will show you the physical layout of each component to the X-ray room (fig. 2–1) and give the clinical application of each when applicable. The first component we will discuss is the X-ray tube.

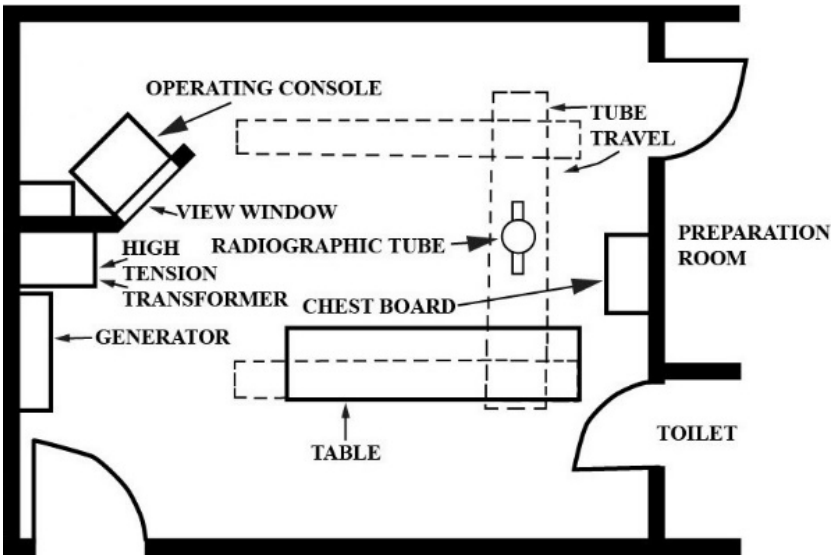

Figure 2–1. Typical X-ray room layout.

X-ray tube

As you can see in our room layout, the radiographic tube is located over the tabletop. The tube itself is the source of X-ray for the unit. As we already covered much of the tube's internal X-ray generation components in the last unit, we will just focus on the housing and related components in this section. The X-ray tube in figure 2–2 illustrates how the tube is housed in a steel case, and lead-lined to prevent radiation leakage. The case has a window port allowing the useful beam to exit at the proper angle. The entire tube housing is filled with a pure oil to assist in heat transfer and cooling of the X-ray tube. The area where the beam exits the housing, is also where a collimator or beam-limiting device will mount to the tube.

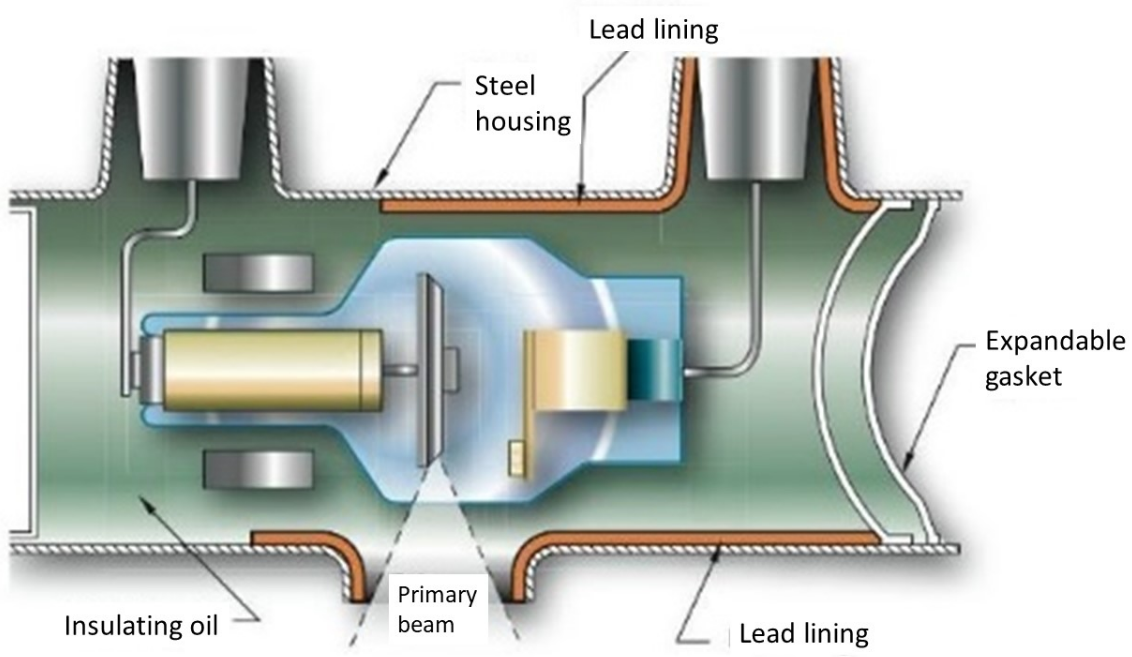

Figure 2–2. X-ray tube inside housing. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Beam-limiting devices

This is a device attached to the X-ray tube housing to direct and restrict the X-ray beam to the body part under examination. The simplest beam-limiting device is a cone, which is simply a round or rectangular lead-lined tube attached to the tube. Several examples are illustrated in figure 2–3; these are mostly restricted to dental X-ray devices. This type of device provides an X-ray beam of a fixed size and shape.

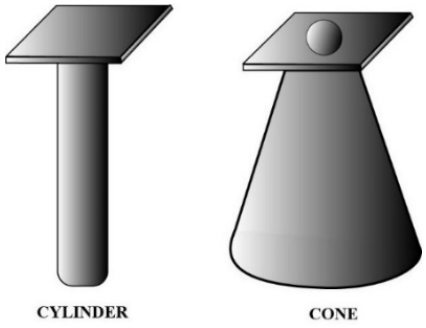

Figure 2–3. Cones for beam-limiting.

The most common beam-limiting device is called a collimator or variable aperture collimator (fig. 2–4). Not all X-rays are emitted precisely from the focal spot of an X-ray tube. Some X-rays are produced when projectile electrons stray and interact at positions on the anode other than the focal spot. Such radiation is called off focus radiation and tends to diminish the sharpness of the radiograph. To control off focus radiation, a first-stage entrance-shuttering device, consisting of multiple collimator blades, protrudes from the top of the collimator into the X-ray tube housing. The second-stage collimator shutter leaves are usually lead, at least 3-mm thick. They work in pairs and are independently controlled, thereby allowing for rectangular and square fields.

Light localization in a typical variable aperture collimator is accomplished with a small lamp and mirror. The mirror must be far enough on the tube side of the collimator leaves to project a sufficiently sharp light pattern through the leaves when the lamp is on. The lamp, mirror, and collimator leaves must all be adjusted so the projected light field coincides with the X-ray beam. If the light and X-ray beam do not coincide, adjustment of the mirror or lamp is usually necessary.

Nearly all light localizing devices used today are automatic. They are called positive beam-limiting (PBL) devices. When film is loaded in the bucky tray and clamped into place, sensing devices in the tray identify the size and alignment of the cassette. An electronic signal is transmitted to the

collimator housing and actuates synchronous motors that drive the collimator leaves to a pre-calibrated position, so the beam is restricted to the size of the film being used. When properly adjusted, the automatic collimator provides an unexposed border on all sides of the finished radiograph.

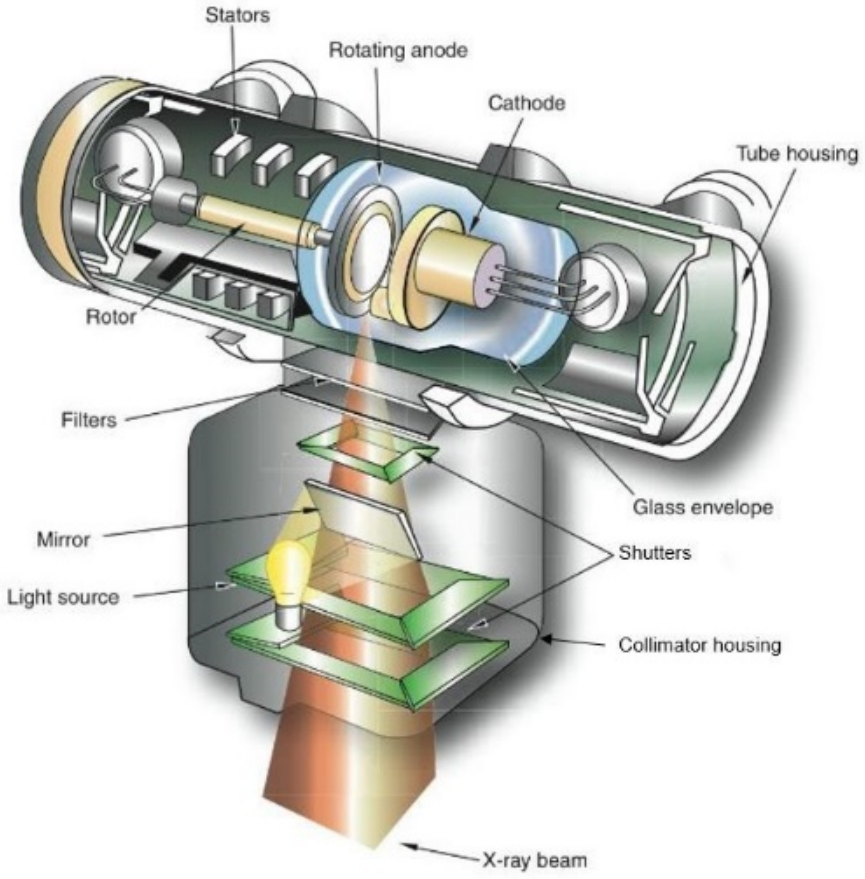

Figure 2–4. Collimator. **(Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)**

High-voltage transformer

The high-voltage transformer converts normal building AC power to the high voltage necessary to operate the X-ray tube. The transformer includes the rectifiers we have discussed to convert the AC to DC. The approximate output range of typical X-ray units is 40,000–150,000V. The transformer itself is filled with pure insulating oil.

Console

The operating console (fig. 2–5) is the apparatus that allows the technician to control X-ray tube current and voltage, ensuring a useful beam of proper intensity and penetrability for producing a good-quality radiograph. The console usually provides control of line compensation, kVp, mA, and exposure time.

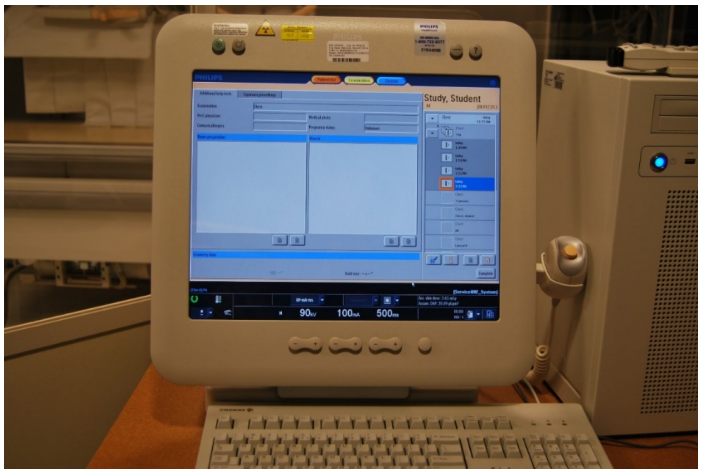

Figure 2–5. X-ray console.

Tilt table

The radiographic table is designed to support the patient in a position that will enhance radiographic examination. The tilt table is basically a simple device relative to the control circuits; however, the table does play an important role in the success of the overall procedure. The tabletop must be uniformly radiolucent to easily permit X-rays to pass through. Most tabletops use carbon graphite fiber to reduce absorption of photons. The table has several special functions to include:

- A means to position the patient automatically by the use of a tabletop, motor-drive assembly.
- Can be rotated from 0° horizontal to a 90° vertical position for special-purpose techniques (fig. 2–6).
- Houses several special components, such as the cassette holder and bucky device.
- May contain an AEC circuit that automatically terminates the exposure to the film, depending on patient density, for consistent radiographs throughout a wide range of patient densities.
- May contain an extra X-ray tube for special procedures to include a spot film tower and an image intensifier.

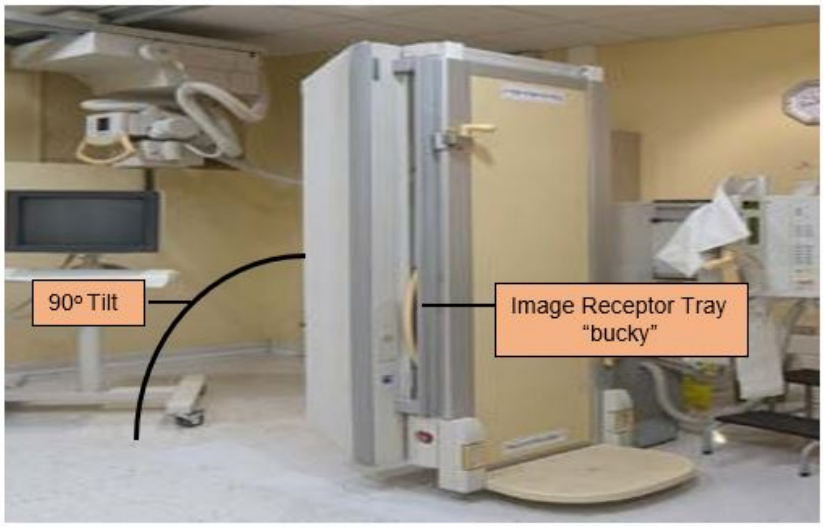

Figure 2–6. X-ray tilt table.

Cassette holder

A cassette holder is a device that holds the cassette in place while an X-ray is taken. In basic radiography units, the cassette holder is located just under the tabletop, so the patient can be positioned between the X-ray tube and the film. Another common location for a cassette holder is mounted on a wall stand.

Wall stand

The wall stand, also known as a chest stand, is an upright cassette holder or bucky unit and is a common and useful ancillary piece of equipment in any radiographic room (fig. 2–7). Chest radiography should routinely be done in an upright position; there are numerous other procedures that are best done upright as well (e.g., acromioclavicular joints, abdominal obstructive procedures, cervical spine, etc.). Upright cassette holders may or may not include a radiographic grid. Upright bucky units may include the same equipment as the bucky tray located in a table (movable radiographic grid, cassette tray, and automatic exposure control sensors).

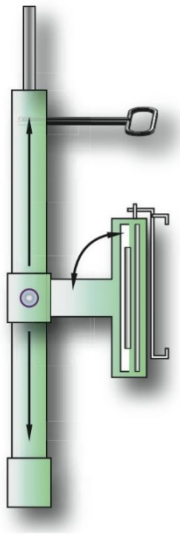

Figure 2–7. Wall stand. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Safety

During the design of fixed X-ray imaging facilities it is necessary to ensure that the layout of the equipment and shielding of the room are such that the exposure to personnel and members of the public within adjacent areas is within the recommended equivalent dose limits. This is possible by using structural protective barriers made of materials having effective X-ray attenuating properties and of thicknesses sufficient to reduce exposures to the desired levels. Commonly used materials include lead sheet, concrete, lead glass, steel, and leaded acrylic.

Protective barriers are classified as either primary barriers or secondary barriers. Primary barriers can be struck by the primary beam, or useful beam, exiting the X-ray tube. Secondary barriers can only be struck by scattered and leakage radiation. The primary beam cannot be directed at secondary barriers. Secondary barriers are always thinner than primary barriers. Barrier requirements are dependent on the type of equipment, its layout within the room, the occupancy of adjacent areas, and other factors. Such shielding should always be designed by a qualified diagnostic radiological physicist.

The X-ray room is generally lined with at least a $^{1}/_{16}$″ sheet of lead or lead equivalent to protect the neighboring rooms from scatter radiation. The X-ray control is behind a protective barrier to shield the X-ray technician during exposures. The X-ray room layout further demonstrates the typical arrangement of table, tube, cassette holder, and high voltage transformer so that ease of patient transportation on gurneys can be accomplished.

609. Typical fluoroscopic systems

X-ray examinations by fluoroscopy make it possible to study internal organs in motion; therefore, they are vitally important in modern diagnosis. A fluoroscopic examination, for instance, is often used to observe the natural real-time action of the heart, stomach, and GI tract.

Originally, fluoroscopic studies were done on a fluoroscopic viewing screen treated with chemicals that fluoresce (give off visible light) when exposed to the action of X-rays. When X-rays strike such a screen, either directly or after passing through some part of the body, an image in visible light is produced on the screen. The brilliance of the light and detail of the image depend on the quantity and distribution of the rays striking the screen. The more dense or thicker a given area of tissue, the more rays will be absorbed and the darker the shadow on the screen.

In fluoroscopy, the images of various organs are seen as dark shadows against an illuminated background. The brightness of an image on a fluoroscopic screen is reduced to a minute fraction of its original value by the absorption of X-rays in a patient's body when the patient is between the X-ray source and the screen. As a result, it is difficult to properly discriminate between contrasts or general sharpness in the image. Fluoroscopy requires intensification of X-ray image brightness on the fluoroscopic screen. This image is usually viewed electronically by using a television monitoring system and can be videotaped for further reference.

The fluoroscopic examination is an important part of the radiologist's diagnostic process. While a fluoroscopic exam enables the physician to visualize organs in motion, it can also visualize:

- Insertion of catheters into blood vessels of the heart.
- Introduction of opaque substances into organs.
- Parts of the body in different planes and projections.

System configuration

In most situations, you will not encounter fixed X-ray systems designed strictly to do fluoroscopic examinations. The typical system is capable of fluoroscopic and radiographic use. What is normally found in Air Force (AF) hospitals is a basic radiographic system, as described in the preceding lesson, with additional components to provide fluoroscopic capability (fig. 2–8).

Rather than add enough components to create two separate systems, some components serve dual purposes. For example, fluoroscopic capability usually requires the addition of a second X-ray tube to the radiographic system. However, the same control and transformer that operates the radiographic tube normally operate this tube by use of a switching circuit. There are some components that cannot be shared for this specialized procedure. They include the X-ray tube, image intensifier or other image acquisition method, and the monitoring system. Let's briefly discuss each major component.

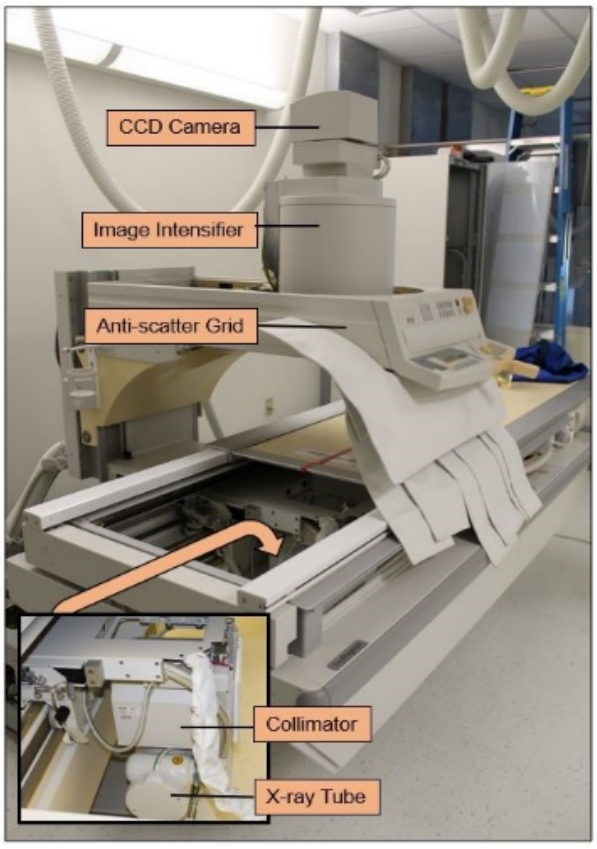

Figure 2–8. Typical fluoroscopy room layout with components.

X-ray tube

The fluoroscopic tube is housed inside the tilt table. Fluoroscopic X-ray tubes are very similar to diagnostic tubes except that they are designed to

operate for longer periods of time at much lower mA. Where a typical diagnostic tube operates at 50–1,200 mA, the fluoroscopic mA range is 0.5–5.0 mA. The kVp of operation depends entirely on the section of the body being examined. The fluoroscopic X-ray tube and image receptor are mounted on a C-arm to maintain their alignment at all times (fig. 2–9). This also keeps the tube target fixed to prevent a source-to-object distance (SOD) of less than 15" (38 cm). The fluoroscopic tube is operated by a foot switch, which permits the fluoroscopist to have both hands free for patient positioning. Modern fluoroscopic equipment allows the radiologist to select an image brightness level; this is subsequently maintained automatically by varying the kVp, mA, or, sometimes, both. Such a feature of the fluoroscope is called automatic brightness control (ABC), automatic gain control (AGC), or AEC (which you learned about in unit 1).

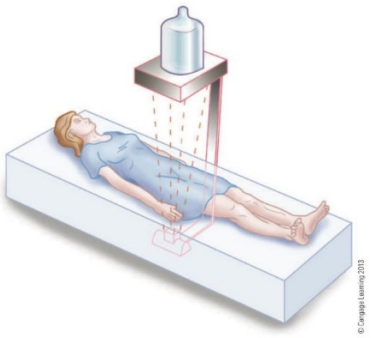

Figure 2–9. C-arm tube alignment. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Image acquisition components

Current fluoroscopy systems fall into two distinct categories: image intensifier and flat panel detector (FPD). Image processing on a fluoroscopic unit is either captured indirectly or directly. The method most common to fluoroscopy involves using an image intensifier, an indirect image acquisition process, that turns X-rays into light and then processes the image. The direct method involves using FPDs in place of an image intensifier, which captures the image without needing conversion to light. Let's look at each of these systems.

Image intensifier

The more conventional and older design is the image intensifier system. The image intensifier tube is a complex electronic device that receives a remnant X-ray beam, converts it to light, and increases the light intensity. Figure 2–10 is the rendering of an image intensifier tube. The tube is usually contained in a glass envelope that provides some structural support and, more importantly, maintains a vacuum. When installed, the tube is mounted inside a metal container to protect it from rough handling and possible breakage.

The X-ray beam, after passing through the patient, enters the vacuum glass envelope where the X-ray image strikes a fluorescent screen called the input phosphor. There, the X-ray is converted into visible light. This light is proportional to the amount and intensity of the radiation (X-ray image). The input screen is separated from the next stage, the photocathode, only by a thin transparent piece of film. As its name implies, the photocathode reacts to the light, stimulating the production of photoelectrons. These two stages are physically located close together for a reason. Think about this, what would happen to the light intensity if the photocathode were located farther away from the input phosphor? Your light would not be as strong and wouldn't stimulate the production of photoelectrons. The number of photoelectrons produced is directly proportional to the number of light photons made.

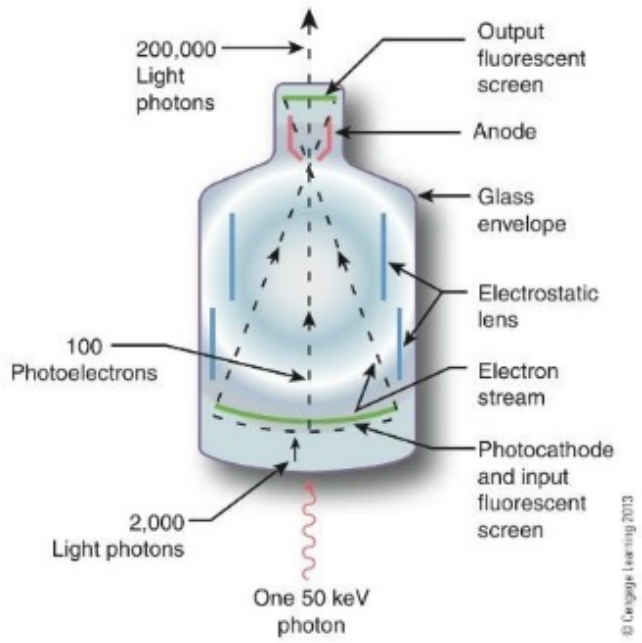

Figure 2–10. Image intensifier tube. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

The image intensifier tube is approximately 50-cm long. A potential difference of about 25,000V is maintained across the tube between the photocathode and anode. The photoelectrons are then accelerated, by the high positive voltage placed on the anode, toward the output phosphor. These accelerated photoelectrons are focused by electrostatic focusing plates or lenses located along the sides of the image intensifier. The accelerated photoelectrons travel past the anode and strike the output phosphor of the image intensifier. The electrons arrive at the output phosphor with high kinetic energy and contain the image of the input phosphor in minified form. When these high-energy electrons strike the output phosphor, they produce light many times brighter.

Figure 2–11 summarizes the entire sequence of events from initial X-ray interaction to output image. This ratio of the number of light photons at the input phosphor to the number of photons at the output phosphor is called the flux gain.

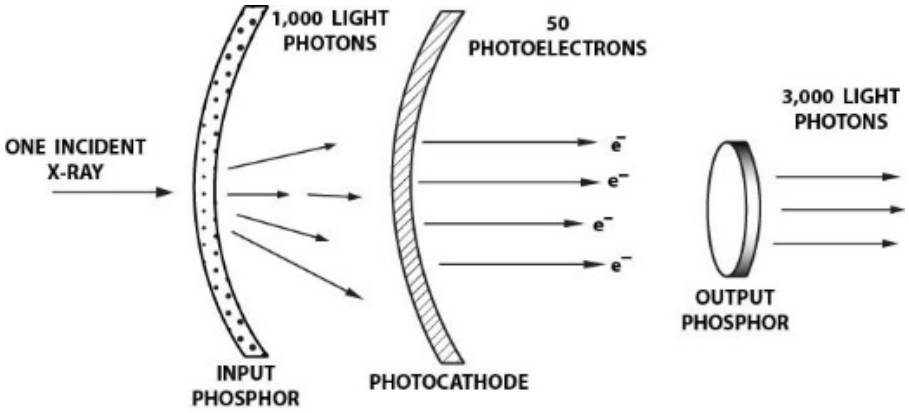

Figure 2–11. Image intensifier operation.

Flux gain is a measurement of the increase in light photons due to the conversion efficiency of the output screen. For example, if the output phosphor produces 50 light photons for each electron that

strikes it, the flux gain would be 50. Flux gain does not take into account the conversion efficiency of the input screen.

The minification gain is the ratio of the square diameter of the input phosphor to the square of the diameter of the output phosphor. Minification gain is simply an increase in brightness or intensity, not an improvement in the quality or number of photons making up the image.

The increased illumination of the image is due to the multiplication of the light photons at the output phosphor compared with those at the input phosphor, and the image minification from input to output phosphor. The ability of the image intensifier tube to increase the illumination level of the image is called its brightness gain. The brightness gain is simply the product of the minification gain and the flux gain. The primary brightness gain occurs from the acceleration and focusing of the electron beam. The acceleration of the electron beam increases its energy and its ability to emit light at the output screen.

Brightness gain deteriorates as much as 10 percent per year because, just as with intensifying-screen phosphors, the input and output screen phosphors age. Although difficult to measure, brightness gain can be evaluated by monitoring the radiation dose required to obtain a diagnostic image from a standard phantom, such as an abdomen or pelvis. Various systems with different names are used to automatically maintain satisfactory fluoroscopic image density and contrast. Automatic brightness control is the most common term, although the terms automatic dose control (ADC) and automatic brightness stabilization (ABS) achieve the same result. They maintain the brightness of the image by automatically adjusting the exposure factors as necessary according to subject density and contrast. Most ABC systems monitor the current flowing between the cathode and anode of the image intensification tube or the intensity of the output screen. In all systems, the primary beam is changed when current and intensity fall below established levels. Regulation of the primary beam can be accomplished by varying kVp, mA, and pulse time.

Input phosphor size varies from 10–35 cm and is used to identify image intensifier tubes. The brightness gain of most image intensifiers is 5,000–20,000 and decreases with tube age and use. The image intensifier tube allows for great flexibility in manipulation of fluoroscopy information.

Multi-focus tubes

Another development is the dual-focus or trifocus tube. These multifield image intensifiers provide considerably more flexibility for all fluoroscopic examinations, and are standard components in digital fluoroscopy. Dual focus tubes come in varied sizes, but perhaps the most popular is the 25 cm –17 cm (25/17) design. Trifocus tubes of 25/17/12 or 23/15/10 are also used.

These numeric dimensions refer to the diameter of the input phosphor of the image intensifier tube. For example, a 23/15 dual- focus tube has a 9" (23-cm) input screen when operating normally and uses a 6" (15-cm) area when magnifying (fig. 2–12). In a typical multifield tube, when the 23-cm mode is selected, the photoelectrons from the entire input phosphor/photocathode action are accelerated to the output phosphor. When switched to the 15-cm mode, the voltage on the electrostatic focusing lenses is increased, causing the electron focal point to move further from the output phosphor. Consequently, only electrons from the center 15-cm diameter of the input phosphor are incident on the output phosphor. The principal result of this change in focal point is to reduce the field of view and magnify the image. Use of the smaller dimension of a multifield image intensifier tube always results in a magnified image with a magnification factor in direct proportion to the ratio of the diameters. A 23/15 tube operated in the 15-cm mode produces a magnified image 1.5 times larger than the image produced in the 23-cm mode.

This magnified image comes at a price. When operating in the magnified mode, the minification gain is reduced and there are fewer photoelectrons present on the output phosphor—a dimmer image results. To maintain the same level of brightness, the X-ray tube mA is automatically increased, thus increasing the patient dose. The increase in patient dose does, however, result in better image quality.

The portion of any image resulting from the periphery (outer boundaries) of the input phosphor is inherently unfocused and suffers from vignetting, a reduction in brightness at the periphery. Because only the central region of the input phosphor is used in the magnification mode, spatial resolution is better.

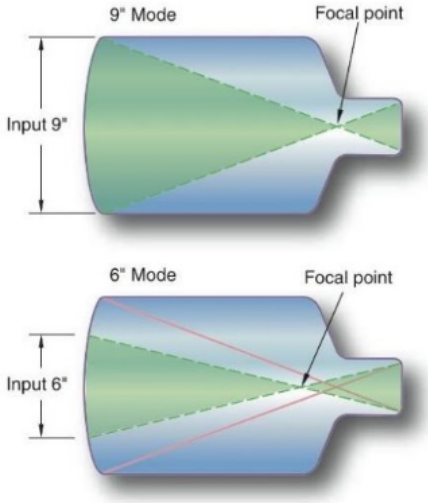

Figure 2–12. 23/15 Dual-focus image intensifier. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Image capture

There are many ways of viewing and recording the image on the output phosphor of the image intensifier. Previous versions of image intensified systems used a partially-silvered mirror, which directed light in the optical coupling box to an objective lens that focused the light image onto either a cine movie camera for video recording, or towards a television camera for real time viewing. They also featured a spot film device, which captured a single radiographic exposure before the image intensifier. In spot film devices, the unit would have a mechanical device that moved a standard X-ray cassette into position for a single exposure during a fluoroscopic examination. Cinefluoroscopy, consisted of a cine camera, driven by synchronous 60 Hz motors, shutters, and timed exposures with a pulsating X-ray beam at framing frequencies of 7.5, 15, 30, and 60 frames per second (fig. 2–13).

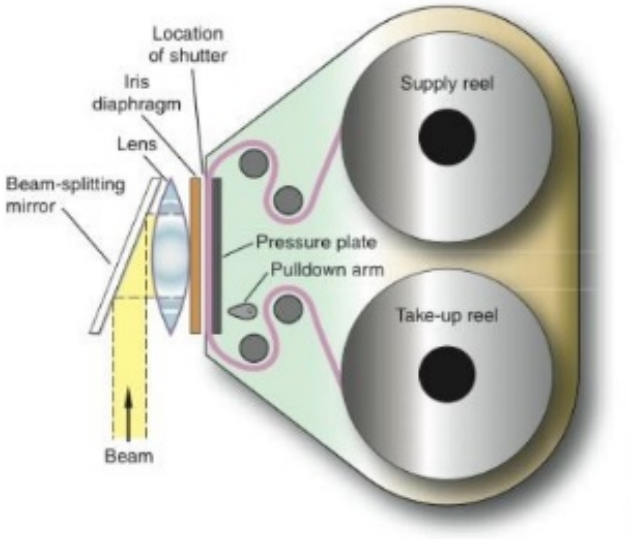

Figure 2–13. Cine camera. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

The video cameras used in image intensifier systems were originally vidicon or plumbicon analog devices borrowed from the broadcast television industry. After passing through a lens system and an aperture, the television camera tube intercepted the light image and converted the light pattern into a series of electrical signals that could be displayed on the television monitor. In modern systems, we use digital cameras based on CCD image sensors or complementary metal oxide semiconductor (CMOS) technology. The most common method to view and record the images is the use of a CCD camera. The CCD consists of crystalline silicon that is located on the output phosphor of the image intensifier (fig. 2–14).

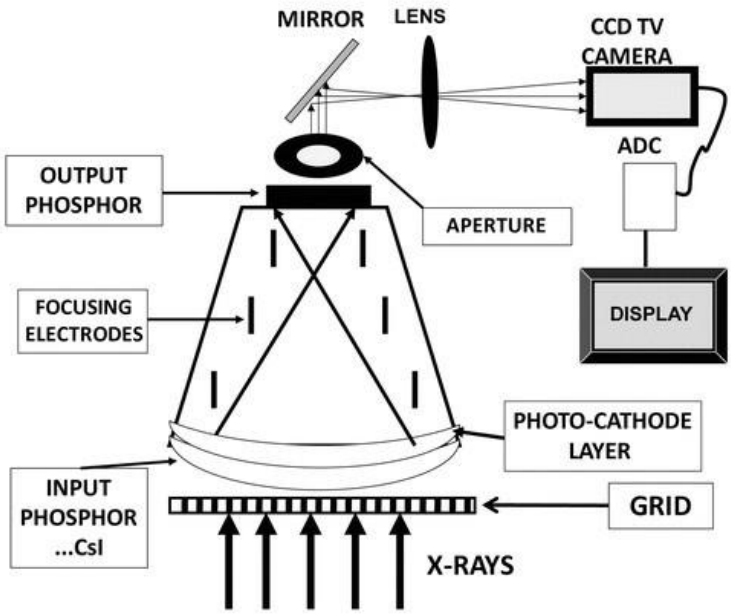

Figure 2–14. Imaging modes with image intensifier.

Digital images are obtained when the silicon is illuminated, causing generation of an electrical charge. This electrical charge is then sent as a signal to the analog-to-digital converter. The analog-to-digital converter (ADC) digitizes the signal from the CCD by converting the signal into pixels or picture elements. The ADC assigns a numerical value to each pixel corresponding to the electronic signals. The pixel numerical values are then organized (mapped) into a matrix (fig. 2–15). These images appear on monitors, which can be placed in the room or in specific viewing areas.

Figure 2–15. Digitizing an image.

The television system itself is the biggest limitation of image intensifier fluoroscopy systems. The spatial resolution that is directly measured at the output phosphor of the image intensifier is inherently good, at 5 line pairs per millimeter or more. However, the resolution of the television system is limited by the number of display raster lines and its bandwidth frequency.

Another limitation of the image intensifier fluoroscopy system is its dynamic range, the ratio of the highest input radiation level to the image receptor (without saturating) to that of the lowest level (just above the noise level) at which the unit can operate properly. If a portion of the incident X-ray beam passes through low-density anatomy, such as the lung or air external to the body, the higher X-ray

flux into the image intensifier will cause the system to saturate. Image intensifier systems have a much lower dynamic range than FPD systems.

Flat panel detector arrays

In recent years we have seen the introduction of fluoroscopic systems in which the image intensifier and video camera components are replaced by an FPD assembly. As we covered the theory of FPD technology in unit 1 of this volume, this lesson will brush up on some concepts and focus on its relevance to fluoroscopy.

FPD fluoroscopy systems utilize more modern solid-state detector arrays as the image receptor. They do not require a television camera to produce an electronic signal for the display monitor. FPDs consist of an array of individual detector elements (DEL). The elements are square, 140–200 microns per side and are fabricated using amorphous silicon thin-film technology placed onto glass substrates. By its design, the image receptor (FPD) produces a digital electronic signal, which represents the intensity of the X-rays that impinge on each detector element in the solid-state FPD array. Moreover, the entire process is digital, which reduces image noise caused by electronic components.

Detector arrays used for fluoroscopy range from about 20 x 20 cm up to 40 x 40 cm depending on the manufacturer. However, some manufacturers specify the size of the FPD by providing a diagonal measurement, while others quote the edge dimension. A single detector may contain as many as 5 million individual DELs. A CsI scintillation layer is coated onto the amorphous silicon, with thin-film photodiodes and transistors capturing the visible light signal from the scintillator to form the digital image. This image is then transferred to a computer at a frame rate selected by the user. Frame rates can be as high as 30 frames per second.

FPD receptors have a number of advantages over image intensifier fluoroscopy systems including better stability, lower patient radiation doses, and wider dynamic ranges. Many of the degradations associated with the use of combined image intensifier and television camera systems are not present in images obtained with FPD fluoroscopy systems. Images obtained with FPD systems also do not exhibit geometric deformation such the “pincushion” effect and “S” distortion because the individual DELs in the FPD array are manufactured in straight rows and columns. Consistent production techniques and appropriate software calibration ensure excellent uniformity; the veiling glare (glare extending from very bright areas) and vignetting (loss of brightness at periphery) that occurs with the use of image intensifier systems are not present in images obtained with FPD systems. In addition, because each DEL is fixed in a constant position, images obtained with FPD systems do not exhibit defocusing effects.

Flat panel detectors are also more physically compact than image intensifier video systems, allowing more flexibility in movement and patient positioning. Another advantage of FPDs is that the image receptor’s spatial resolution is defined primarily by the DEL size, and unlike the image intensifier video, is independent of the field of view. In image intensifier systems, the minification gain requires the entrance dose to vary inversely with the field-of-view to maintain a constant brightness at the output phosphor. No such constraint exists for FPDs; the entrance detector dose is independent of the field of view. FPD fluoroscopy systems have begun to dominate angiography and cardiac catheterization laboratories. Currently, only their high purchase cost is preventing their utilization with low-end fluoroscopy equipment such as gastrointestinal fluoroscopy systems and C-arm mobile units (which we will briefly cover in a later lesson).

FPD fluoroscopy systems have their own unique limitations. A challenge of the manufacturing process is to make a uniform array with few defective or degraded DELs; if there are too many defective DELs, image quality suffers. Manufacturers of FPD systems often compensate for defective DELs by using software to interpolate values for those defective elements. However, this interpolation may introduce artifacts. Moreover, FPD systems usually are temperature sensitive, and the images may be affected by changes in temperature. FPD detector arrays are also sensitive to mechanical shocks, which can permanently damage the device and are very expensive to replace.

Self-Test Questions

After you complete these questions, you may check your answers at the end of the unit.

608. Typical radiologic systems

1. Where is the radiographic tube located?

2. What is the simplest beam-limiting device?

3. What is the most common beam-limiting device?

4. What is the second stage collimator shutter leaves made of?

5. What must the collimator leaves, mirror, and lamp coincide with?

6. What is a PBL and how does it work?

7. What is the output range of the high voltage transformer?

8. The operating console usually provides control of what?

9. List the five functions of the tilt table.

10. What are the two common places you'll find a cassette holder in a radiographic room?

11. What are the two classifications of protective barriers in a typical radiographic room?

609. Typical Fluoroscopic systems

1. What is the clinical application of fluoroscopy?

2. Where is the fluoroscopic X-ray tube located?

3. Why is the image intensifier tube mounted in a metal housing?

4. What does the input phosphor of the image intensifier do?

5. What is the second stage of the image intensifier?

6. Why is the photocathode physically located so close to the input phosphor?

7. How are the photoelectrons accelerated and focused onto the output phosphor?

8. What is flux gain?

9. What is brightness gain?

10. What is the most popular dual focus tube?

11. What is the principal result when you change to the smaller focal field in a dual focus tube?

12. What must be done to maintain the same level of brightness when using the smaller focal field?

13. What is the most common method to view and record images in fluoroscopy?

14. Explain the process of obtaining a digital image.

15. What is the weakest link in image intensified fluoroscopy?

16. What component replaces the image intensifier and video camera in digital fluoroscopic systems?

17. How many individual detector elements may a single detector contain?

18. How do manufacturers of FPD systems often compensate for defective DELs?

2–2. Mobile, Mammography, and Dental X-radiation Systems

Up to this point, we have only covered fixed radiologic and fluoroscopic systems. You will, however, also encounter many other X-ray systems in your MTF, including mobile radiologic and fluoroscopic systems, mammographic, and dental X-ray systems. While you may not have all of these systems, rest assured they are in the medical community and you will likely deal with them at some time during your biomedical equipment technician (BMET) career.

610. Mobile systems

One of the more common X-ray systems in an MTF is the mobile X-ray system to include mobile radiologic and fluoroscopic. These systems are similar in many ways to fixed systems, but also have numerous differences. Because of their mobile nature, they also will take more abuse and probably require more maintenance than the fixed systems. We'll begin our discussion of mobile systems by looking at a typical mobile radiographic system.

Mobile radiographic system

Within most MTFs, you will find mobile radiographic units (fig. 2–16), which are used to perform radiographic X-rays outside the radiology department. This could occur in surgery, the intensive care unit (ICU), labor and delivery, the emergency room (ER), and wards. The unit itself is battery powered and provides limited technique factors.

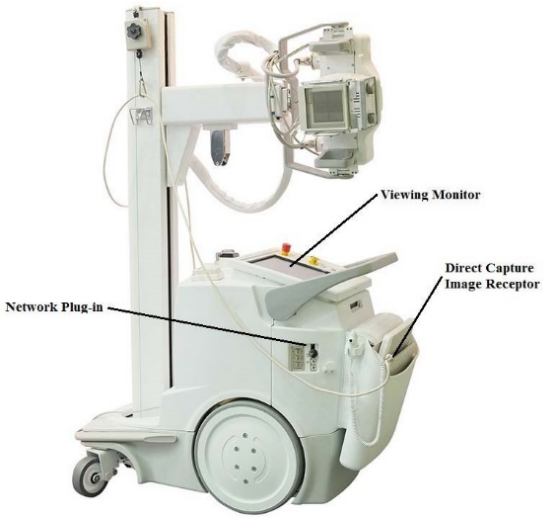

Figure 2–16. Mobile radiographic unit - digital.

Basic operation

In this battery-powered unit, the DC voltage from the batteries is converted to AC voltage in the inverter section of the unit. This AC voltage is then supplied to a high voltage transformer where it is stepped up to the correct level, rectified, and sent to the X-ray tube. As you can see, this process works just like a fixed unit, except the original voltage is supplied from battery power rather than facility voltage. Mobile units produce an X-ray beam with an average photon energy that is fairly similar to that of stationary equipment. Full-power high-frequency units that are battery powered produce a steady waveform, as shown in figure 2–17A. Because this waveform has very little ripple, it is as efficient as a stationary unit. Before high-frequency mobile units were developed, capacitor discharge units made exposures through the use of large capacitors, which allowed the proper kVp range when sufficient power was not available, (fig. 2–17B). As a result, capacitor discharge units often had serious kVp limitations.

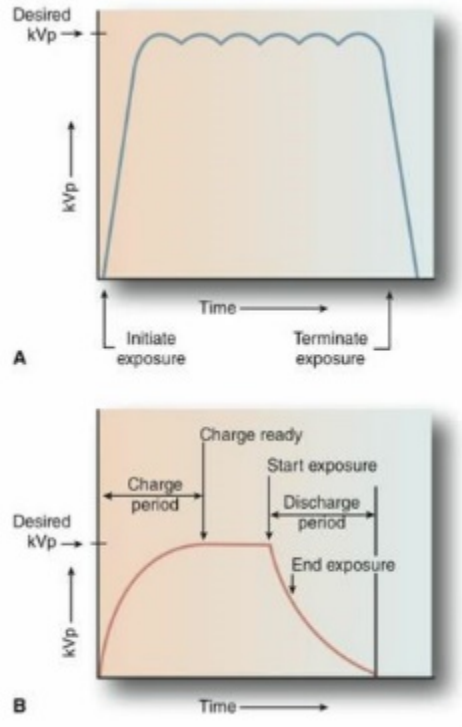

Figure 2–17. Mobile X-ray waveforms. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Improved performance

Newer mobile units come with built-in monitors and flat panel detector systems for immediate display of the image. Detectors may be tethered to the unit or use wireless technology (more common) to send the exposure data to the unit for processing and display. This allows the radiographer to view the image at the bedside and for it to be sent to a central viewing and storage system for physician review.

One advantage of a battery-powered unit is that power is available to drive the unit itself. The drive switch on a mobile unit should be of the dead-man type so that its release immediately disengages the power drive. The size and weight of a battery-powered mobile X-ray unit are considerable, and extreme care must be observed when piloting it. Patients and staff could be seriously injured by careless driving.

Mobile fluoroscopic unit

Just as with radiography, fluoroscopy has mobile units available and your MTF may have one or more units. A mobile fluoroscopy unit is used outside the X-ray department to perform fluoroscopic exams, primarily in orthopedics and surgery.

The most common model has the C-arm design (thus the common name C-arm) so patients are not required to shift or move for the examination as the C-arm can be adjusted. There is also a nonadjustable distance between the X-ray tube and the image intensifier. When the clamp or lock holding the C-arm in position is released, it permits both the tube and image receptor to be rotated to a new position. Much like its fixed counterpart, the image intensifier sports a camera and viewing monitor. It does have the capability of shooting a radiographic shot by placing a cassette in a special cassette holding device that attaches to the image intensifier input housing. Also like its fixed counterpart, newer digital fluoroscopy units will use a FPD system in place of an image intensifier, allowing for a completely digital process and eliminating the need for cassettes to attain a radiographic shot. The kVp and mA are limited since fluoroscopic techniques do not require the high technique shots of radiographic imaging. Figure 2–18 depicts a typical mobile fluoroscopy unit.

This unit differs from the mobile radiographic unit in that it does not have a drive system or battery system. The unit uses line power instead of batteries to power the unit for imaging, and good old-fashioned muscle for moving throughout the facility. One area the C-arm has in common with the mobile radiologic unit is the need for proper maintenance. Because this unit is moved around, it will receive more abuse than its fixed relative. During your maintenance, always look carefully for any damage and fully utilize the manufacturer's literature when performing preventive maintenance inspections.

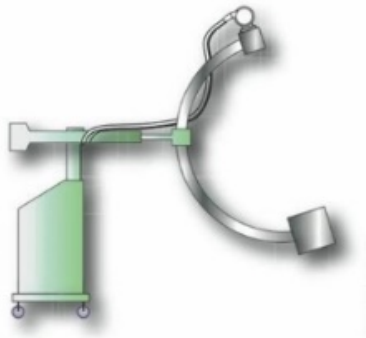

Figure 2–18. Mobile fluoroscopy unit. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

611. Mammography

Up to this point, we have discussed radiographic examinations of traditional structures; however, there is a need to examine other anatomical structures, such as soft tissue. Radiographic examination of soft tissues, called soft tissue radiography, requires selected techniques that differ greatly from conventional radiography. This is due to the substantial differences in the anatomical structures being radiographed. In conventional radiography, the subject contrast is great because of the large differences in mass density and effective Z among bone, muscle, fat, and lung tissue. In soft tissue radiography, however, only muscle and fat structures are imaged, and these tissues have similar Zs and mass densities. Consequently, in soft tissue radiography, the techniques employed are designed to enhance differential absorption in these very similar tissues. The main area of soft tissue radiography we will be concerned with is mammography.

Clinical applications

The main purpose of a mammogram is for the detection of breast cancer. Breast cancer is one of the most common forms of cancer in women and the greatest single killer of women between ages 45 and 50. Because of this, effective early detection is imperative for prevention and treatment of breast cancer. Mammography is a very effective tool used by physicians, but it requires special training, skill, and knowledge to be used properly. Mammograms must be of the highest quality to be accurately interpreted.

There are two types of mammography examinations: diagnostic (performed on at-risk patients or those with symptoms) and screening (performed on women with no symptoms). The first mammogram a woman receives is called the baseline mammographic examination and is kept on file for the life of the patient. This exam is used as a reference for all future exams.

A routine mammogram consists of four views—two views per breast. A craniocaudal projection (CC) is taken to view the breast from the top down. This view reveals the entire medial (closest to the sternum), central, and part of the lateral portions of the breast. The second view is performed with a 45-degree oblique angle. This position reveals the most amount of breast tissue—the majority being the lateral portion of the breast. The oblique angle does not visualize the medial tissue of the breast.

There may be additional or special views taken on patients with breast implants; these views can also help improve visualization of the breast tissues on patients with a tumor to see if the lesion is cancerous, benign, even non-existent. Some of these views are the "coned down view," "magnification view," "roll," "tangential," "lateral," and the "exaggerated CC."

Mammographic technique

As we stated earlier, in soft tissue such as the breast, only muscle and fat are imaged during a radiologic examination. Because of this, conventional radiographic techniques are useless. An effective mammogram requires a low-kVp technique, but this, in turn, reduces the penetrability of the X-ray beam. This reduced penetrability requires an increase in the mAs, but high mAs results in a high patient dose. Therefore, a balance between the two is struck and results in the use of technique factors between 24–28 kVp.

Another important factor in the examination is breast compression, which is used to hold the breast in position and prevent motion blur. Compression also separates underlying tissue and brings all the breast tissue closer to the image receptor. This results in a more uniform thickness and optical density of the image, and an improved contrast resolution. An additional benefit from compression is the lower radiation dosage requirement.

Principles of operation

At first, standard X-ray equipment was utilized, but was not very effective. The mammography unit was built to meet the specialized and increasing need of screening and diagnostic quality mammograms. Figure 2–19 is a mammography system with standard components.

A mammographic unit includes a generator, X-ray tube, control panel for exposure selection, a compression device, and various other components designed to produce high-quality images. Factors that play a role in image quality are technique, mA, exposure time, and geometric variables, such as focal spot size, target film distance, and subject film distance. Other factors include patient motion and size, the composition of the breast, and the type of image recording system. A complex balance of these factors is required to obtain good mammographic image quality while minimizing the radiation dose. Let's now take a look at the main components of a mammography unit.

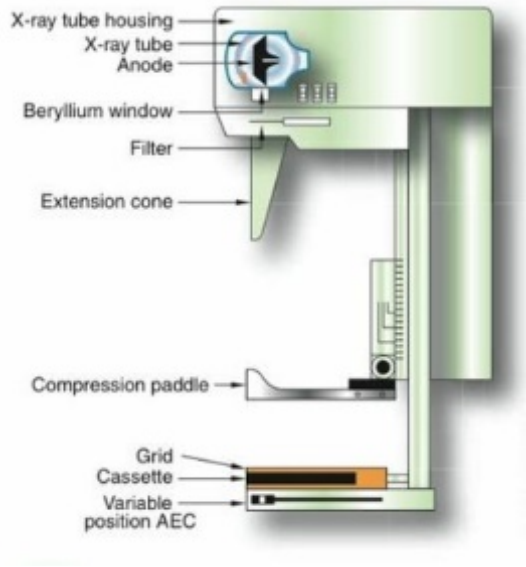

Figure 2–19. Standard mammography unit. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Generator

The majority of all mammography units manufactured today use a high frequency generator. These generators can be of the 1ϕ or 3ϕ input type. A high frequency generator converts the 50- or 60-cycle input voltage to a higher frequency, such as 20 kilohertz (kHz). Higher frequency AC helps simplify filtering, dramatically adds to the unit's compactness, and improves the effective kV output as a percentage of peak kV. The main advantage is it allows for a lower patient dose of radiation.

Tube

One of the biggest differences between a conventional X-ray unit and a mammographic unit is the X-ray tube. The tube for a mammographic unit is designed with one purpose in mind: to produce an image with high resolution and high contrast at a moderate exposure level.

The tube for a mammographic unit is produced with a rotating tungsten, Mo, or rhodium target. These materials are chosen because the X-rays produced are in the appropriate spectrum for effective mammographic imaging. Also, the focal spot size generally tends to be smaller. Another important consideration in the tube is the heel effect. This is important because the conical shape of a breast requires the radiation density be higher near the chest wall than at the nipple side. This is done by positioning the cathode toward the chest wall (fig. 2–20). Heel effect is used to maximum advantage with this orientation. Greater-intensity X-rays are placed over the chest wall area where the breast is thicker, producing a more uniformly dense image.

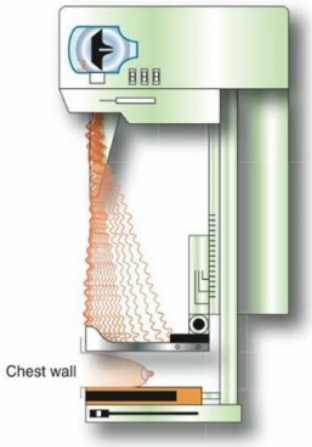

Figure 2–20. Mammography unit chest wall. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Another important factor of the tube is filtration. Because the kVp setting used in mammography is low, it is important the tube window does not attenuate the beam. Therefore, mammography tubes have a beryllium window (used because it has a low Z and allows low-energy X-rays to

exit). Filtration must be used at the tube port to attenuate the very low energy X-rays and the high-energy bremsstrahlung X-rays, which are not useful in producing the image. To accomplish this, the filter is made of the same element as the anode material.

Compression device

As previously mentioned, all mammography units have a built-in compression device. The compression device is made from a special plastic that allows transmission of the low energy X-rays. It will have a straight chest wall edge to allow the compression to grasp the breast tissues close to it. The radiographer controls the amount of compression, but the available range must be 25–45 lbs., of force. In addition to the main compression device, a smaller "spot" device is usually available to be used on areas of localized focus.

Appropriately applied compression is one of the most critical components in the production of a high-quality mammogram. The overall function of compression is to decrease the thickness of the breast, bring the breast structures as close to the image receptor as possible, and increase radiographic contrast. The specific advantages of this technique are:

- Reduced magnification—lessens geometric unsharpness and increases resolution.
- Reduced tissue thickness—requires less kVp, prompting a reduction in scattered radiation and an increase in radiographic contrast.
- Reduced radiation exposure—occurs because less exposure time is required due to the decreased tissue thickness.
- Reduced motion unsharpness—occurs be- cause the breast is completely immobilized.
- Improved visualization of breast structures—occurs because the structures are spread out over a larger area. There is also less superimposition of overlying structures.
- More uniform image receptor exposure—occurs due to the flattening effect of compression permitting optimal exposure of the entire breast.

Grids

Scatter radiation is produced from the breast during an exposure, which leads to a reduction in image contrast. To overcome scatter radiation, mammographic units use grids. The grid is placed between the breast support (what the breast rests on) and image receptor. It is made from strips of lead and the inner space material is either wood or carbon fiber. All mammography grids are the moving type and move in one direction only (they don't reciprocate).

The grid will remove about 80–90 percent of the scatter. Use of a grid in mammography greatly improves radiographic quality, and today grids are used for all mammographic images. The use of a grid will, however, increase radiation exposure to the patient by a factor of 2–3 times.

Image receptors

There are three ways the mammographic image is received today. The first method is the screen-film combination. Units utilizing this method have an intensifying screen matched with a special film specifically designed for mammography. There are various combinations, but, whatever the combination, the screen and film must be spectrally matched. The screen-film combination is placed in a specially designed cassette that has a front cover made from a material with a low Z. This ensures the maximum amount of X-rays will reach the film.

A second and third methods use modern digital techniques such as CR and DR, including CCDs and FPDs. In a CCD system, the X-ray beam interacts with an intensifying screen, which produces light. This light is captured by a fiber optic bundle or a lens system, and then directed to the CCD. The CCD then converts visible light photons to electrons. The electron signal is read in pixel-fashion to form an image. The CCD must be cooled to –25°C to reduce any electronic noise that might interfere with the image quality. Once the image is produced, it may be viewed, manipulated, and stored as previously discussed. A third system is the selenium flat panel. This system does not use a phosphor;

instead, the X-ray absorber is amorphous selenium. X-rays absorbed by this material are converted into an electrical charge by a system of electrode pads. With these digital detector systems, except in CR, it is no longer necessary to have AEC sensors under the plate.

Automatic exposure control

The AEC is located underneath the grid and image receptor. This device measures X-ray intensity and quality. Like the AEC circuitry used on conventional radiography equipment, its design and purpose is to provide consistent image receptor exposure for the various thickness and density compositions of breast tissues and for the range of kVp used.

The AEC system on most mammography machines utilizes a single radiation-sensitive detector located behind the cassette. The detector is capable of moving toward the nipple, from its primary position at the chest wall, to allow for maximum variations in breast size and for critical magnification work. Most detectors will have 10 stops between the chest wall and the nipple.

The AEC compensates for variances in distance, mA, and patient pathology. Using AECs will require regular tests with phantoms to ensure proper calibration. AECs help to provide uniform, repeatable, high-quality images, thereby reducing the number of retakes and radiation exposure to the patient.

Regulatory requirements

One very important aspect of mammography is the regulatory requirements. These rules are very stringent to ultimately ensure mammography equipment will provide consistent and reproducible images at all times. Regulatory requirements are a detailed subject with which you do not need to be an expert; however, you do need to be familiar with the basic requirements. The process is basically a two-step process, which begins with accreditation.

Accreditation

The Food and Drug Administration (FDA) adopted the Mammography Quality Standards Act (MQSA), which requires that before a mammography facility can legally perform mammography, it must be certified. Before a facility can be certified, it must become accredited. To begin the process, a facility must first contact an accreditation body (AB) and apply for accreditation. Some states have health departments approved as ABs—if your MTF is located in one of these states, the health department is your point of contact (POC). If your facility is located in a state where the health department is not an AB, your MTF must contact the American College of Radiology (ACR). In either case, it is a good idea to contact the state health department because many states have additional requirements. The AB reviews the facility's equipment, personnel (interpreting physicians [the doctors that read the radiographs], radiologic technologists, and medical physicists), and practices. The AB accredits the facility if the review establishes the mammography facility meets the quality standards established under MQSA.

Certification

The next process is certification. Certification is a process separate from accreditation and is administered by a certifying agency (the FDA or an FDA-approved certifying state). An MQSA certificate is issued after the AB notifies the certifying agency of the facility's accreditation. Only MQSA-certified facilities can lawfully provide mammography services. A full certification is good for 3 years, but provisional certificates are available for facilities not yet fully certified. It is again important to note a facility without a full or provisional certificate may not legally perform mammograms. Let's now move on to requirements of the MQSA that apply to you.

Equipment evaluations

This subject should strike a little closer to home for you as a BMET. The MQSA contains additional requirements, which state that mammography equipment (including the actual mammography unit and any related image processing equipment) must be evaluated after certain actions. These actions include installing a new unit or processor; disassembling and reassembling a unit or processor at the same or a new location; and changing or repairing major components of mammography unit or

processor equipment. These evaluations are used to determine whether the new or changed equipment meets the requirements of applicable standards. All problems must be corrected before the new or changed equipment is put into service for examinations or film processing. This equipment evaluation must be performed by a medical physicist or an individual under the direct supervision of a medical physicist.

Quality control checks

A qualified medical physicist (or someone in training under his or her direct supervision) must perform an annual quality control (QC) test. Some portions of the mammography unit tested include the AEC, focal spot, kVp output, X-ray light field, and alignment of the compression device. After the test, the physicist produces a survey report, which lists results of the test and any recommendations for corrective actions. Corrective actions must be performed before any further examinations or within 30 days, depending upon the nature of the problem.

As you can see, the requirements of the MQSA are quite detailed and we have only scratched the surface in this CDC. This should, however, give you a good working knowledge of the terms that affect you. There are no specific requirements listed in the MQSA that address the BMET, only that personnel working on the equipment should be knowledgeable.

612. Dental radiography

The radiology department is not the only place X-rays need to be taken before procedures are accomplished—the very same thing happens in the dental clinic.

Radiographs play an important role in diagnosing dental ailments. The following is a list for some uses of dental radiographs:

- To detect, confirm, or classify dental diseases.
- To detect or evaluate injuries.
- To evaluate growth and development.
- To detect extra or missing teeth.
- To document a patient's dental condition.

In short, a radiograph allows the dentist to view areas of the teeth and the surrounding structure that cannot be seen with a mouth mirror explorer examination. It is important to note, however, that a dentist will never make a diagnosis from a radiograph alone. He or she will always use a visual exam in concert with the X-rays to make a final conclusion about a patient.

There are several techniques used in dental radiography and these techniques can be further divided into two main categories: intraoral, and extraoral including panoramic, cephalometric, and cone beam computed tomography (which we will cover in unit 3 of the volume). The method with which the X-ray film is exposed determines its technique category.

Intraoral techniques

As you have probably figured out by now, this technique involves exposing the X-ray film within (intra) the mouth (oral). There are three common exams considered intraoral:

1. Periapical—this exam shows the roots of the teeth, as well as the surrounding bone (peri = around and apical = end).
2. Bitewing—this exam shows, on a single film, the upper surface of the teeth and portion of the jaw containing the tooth sockets of a given area.
3. Occlusal—this exam shows a cross-section of the upper (hard palate, upper lip, and base of the nose) and lower (tongue, floor of the mouth, and lower lip) dental arches.

Multiple techniques can be used together to perform what is called a complete mouth survey (CMX). The CMX is a series of individual periapical and bitewing X-rays that completely covers all teeth and

tooth-bearing bone structures. The number of X-rays needed for a CMX depends upon several factors, such as the radiographic technique used, number and condition of the teeth, patient age, and individual anatomical makeup; however, the usual number is around 15 periapicals and 4 bitewings.

To perform an intraoral technique, a dental X-ray film must be placed inside the patient's mouth and held in place until the radiograph is taken. There are numerous film holders manufactured to perform this duty; each is designed for a specific purpose and to fit the anatomical makeup of various individuals. An X-ray film is placed within the holder; the holder is then placed within the patient's mouth. The patient is then instructed to bite down to hold the film in place.

There is also a specific type of X-ray unit used to perform intraoral techniques—the fixed system. The fixed dental X-ray system has a wall-mounted control and tube head, and provides limited kVp and mA. The tube uses a cone for collimation instead of the elaborate collimator used on a standard radiographic system. It also utilizes a stationary anode due to the small techniques employed. Figure 2–21 is an example of a fixed dental unit.

Figure 2–21. Fixed dental intraoral unit.

There is also a mobile dental unit, which is a portable radiographic system utilizing a stationary anode and cone type collimator for use within the treatment room. This mobile unit is used to keep the patient from needing to go to another room to have radiographs performed. Its most popular use is during oral surgery or procedures requiring the patient to be sedated. When using the mobile unit, special precautions should be observed due to the hazards associated with radiation because treatment rooms are not X-ray rooms with properly lined lead walls.

Panoramic technique

A panoramic X-ray is a two-dimensional X-ray used in the dental field. It displays both jaws, the upper and lower, as well as the teeth, at the same time. Panoramic films have several uses. They are used regularly for orthodontic assessments, diagnosis of wisdom teeth impaction, diagnosis of advanced periodontal disease, assessment of the jaw joint, as well as for detecting signs of oral cancer. As you can imagine, a patient that receives a CMX is exposed to a rather large dose of radiation. To cut down on this radiation dose, the panoramic technique may be used. This technique is also utilized to overcome certain limitations of intraoral techniques, such as limited coverage of the lower jaw and facial structures. Other advantages of using the panoramic technique include a quicker

procedure, better coverage of the dental arches, and earlier detection of many dental problems. One interesting fact about a panoramic X-ray is it is used for identification of unidentifiable military members' remains.

To produce a panoramic dental X-ray, a narrow X-ray beam (much smaller than the intraoral X-ray beam) is rotated horizontally around the patient's head. The X-ray tube and film move; however, the film moves in an opposite direction of the tube. This is done to ensure the resulting radiograph is proportionally accurate. Most panoramic units have standard features that allow the patient to step up to the unit and place their chin on a chin rest. The tube is mounted at a specified unadjustable distance from the cassette and the only really selectable features are the density of the patient. Once the exposure switch is pressed, the tube head and detector traverse in a semicircle around the head. Only the area directly in front of the tube on the detector is exposed. Figure 2–22 is an example of a panoramic dental unit.

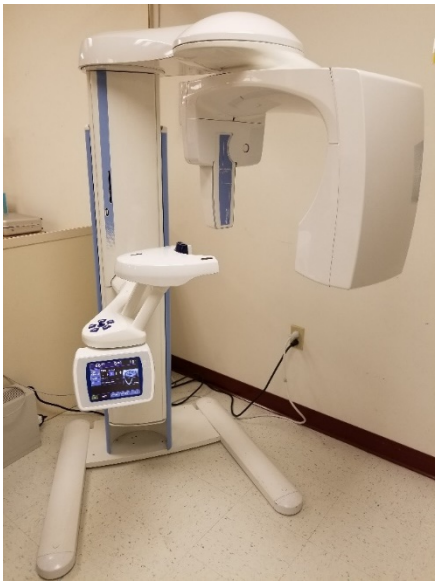

Figure 2–22. Typical panoramic dental unit.

When using conventional film, there are two kinds of film moving mechanisms, one using a sliding flat cassette which holds the film, and another using a rotating cylinder around which the film is wound. Like most other radiographic systems however, dental X-ray systems are also now incorporating digital radiography sensors and computers to capture images. One of the principal advantages compared to film based systems is the much greater exposure latitude. This means a great reduction in repeated scans, which reduces costs and also reduces patient exposure to radiation. Other significant advantages include instantly viewable images and the ability to enhance images.

Cephalometric technique

Having a cephalometric X-ray taken is very similar to having a panoramic X-ray taken, in that both X-rays focus on about the same general area with a similar orbital scanning process. Some panoramic X-ray units are dual purpose units and able to take both cephalometric and panoramic studies. These units will have additional components and software to incorporate both techniques. A cephalometric X-ray, which is also sometimes referred to simply as a ceph, is a diagnostic radiograph used primarily for orthodontic treatment planning. A cephalometric X-ray is taken during an orthodontic records appointment. Cephalometric X-rays are also used by otolaryngologists—doctors who specialize in the treatment of ear, nose and throat (ENT) disorders such as sleep apnea—because these X-rays provide a view of the patient's airways. The process of undergoing a cephalometric X-ray is pretty straightforward; concentrating on the patient's profile (or side view of the head), the X-ray technician positions the patient according to specific criteria necessary when taking the X-ray.

A cephalometric X-ray uses a slightly higher mA range (around 10 mA) compared to a panoramic X-ray (4–8 mA). The exposure usually lasts around 10 seconds and the X-ray can be developed in approximately five or six minutes. Most dental offices are equipped with the equipment necessary to take a ceph X-ray, as well as a pano. Once developed, the dentist typically uses tracing paper to "trace the ceph" in order to calculate how the patients jaw and surrounding bone will be affected by orthodontic treatment. It also provides the dentist with a look into the growth pattern of the jaw and teeth. This process can be used to determine potential courses of action and routes of treatment for dental issues.

Digital radiography has allowed the development of computerized cephalometrics, which is the process of entering cephalometric data in digital format into a computer for cephalometric analysis, as opposed to manually tracing. The process allows for the automatic measurement of landmark relationships. Depending on the software and hardware available, the incorporation of data can be performed by digitizing points on a tracing, by scanning an existing tracing of a conventional radiograph, or by originally obtaining computerized radiographic images that are already in digital format, instead of conventional radiographs. Computerized cephalometrics offers the advantages of instant analysis; readily available race-, sex- and age-related norms for comparison; as well as ease of soft tissue change and surgical predictions.

Self-Test Questions

After you complete these questions, you may check your answers at the end of the unit.

610. Mobile systems

1. What is the main difference between mobile radiologic and fixed units?

2. What are the two types of mobile radiographic units?

3. Why does the mobile radiographic unit use a dead-man switch for the motor drive circuit?

4. In what two areas of an MTF are portable fluoroscopic units primarily used?

5. How does a portable fluoroscopic unit differ from a portable X-ray unit?

611. Mammography

1. What is the main purpose of a mammogram?

2. What are the two types of mammographic examinations and why are they performed?

3. How many views are performed during a routine mammogram?

4. Why are conventional radiographic techniques useless when imaging the breast?

5. What is the purpose of compressing the breast during a mammogram?

6. What type of X-ray generator is found in most mammographic units and why is it used?

7. What range of force must be available for the compression device in a mammographic unit?

8. List the advantages of using breast compression.

9. What is used in mammographic units to overcome scatter radiation?

10. What are the three types of image receptors used in mammographic units today?

11. The FDA adopted what Act to regulate mammography?

12. How long is a facility's full certification good for?

13. What actions require a mammographic unit to be evaluated? Who performs the evaluation?

612. Dental radiography

1. List at least three uses of dental radiographs.

2. What are the two main categories of dental X-ray techniques?

3. List the three common intraoral X-ray exams and their uses.

4. What is the usual number of individual X-rays needed for a CMX?

5. List at least three advantages of the panoramic technique.

6. What is the mA range of a cephalometric technique compared to a panoramic technique?

2–3. Support Equipment

As you have seen, X-ray equipment is a very complex and involved topic. It only stands to reason that support equipment used in conjunction with X-ray procedures is also a multi-faceted topic. We will not attempt to cover all the support equipment you may see, but we will touch on some of the more common ones. We will start with some principles of contrast injectors, and then move on to other digital imaging support equipment such as CR plate readers, laser imagers, and diagnostic imaging monitors. We will conclude this section with a lesson breakdown on picture archiving and communication systems (PACS). Let's get started with contrast injectors.

613. Contrast injectors

Contrast injectors are frequently found throughout an MTF, especially those larger facilities that perform more specialized procedures. In this lesson, we will explore the various clinical applications of the contrast injector.

Clinical applications

Many special radiographic procedures require the introduction of a contrast medium as an aid for imaging internal organs. Barium and iodine compounds are the two most commonly used contrast mediums. Because these compounds have a much higher Z and mass density than soft tissue, it is much easier to see during a procedure when one of these compounds is introduced into an organ. In this lesson, we are going to concentrate on using a contrast medium to view vascular structures. The way these compounds are introduced into the vessels is through the use of a contrast injector.

Contrast medium injection devices allow for the safe delivery of a preset amount of contrast. These devices allow for large bolus injections that are too difficult to be performed by hand. Newer devices even allow for smaller bolus injections that previously could only be performed manually. A typical contrast medium injection device consists of the control panel, syringe, warming device, and pressure mechanism. During angiographic procedures, the contrast medium is injected into the circulatory system. As it enters the blood stream, the contrast medium is diluted. This dilution effect is dependent on the injection site; therefore, the site indicates the speed (rate of flow) of the injection. The larger the vessel, the greater the flow rate must be to maintain proper concentration of contrast medium so the desired anatomical features can be visualized radiographically. In some procedures, flow rates of 30–40 milliliters (ml) of contrast material per second are not uncommon. Naturally, these injections must be performed with a mechanical device. The flow rate is controlled by many factors, such as the viscosity of the contrast medium, catheter length, catheter diameter, and injection pressure. The flow rate or injection speed can be increased or decreased by varying any of these parameters.

Contrast injectors can be classified by their mode of action; they may be electro-mechanical, air pressure activated, or hydraulic. Most current injectors are designed for operation as high- or low-pressure injectors, thereby broadening their usefulness in many types of angiographic procedures. Figure 2–23 is a sample of a contrast injector. They are used more often in an angiographic (angio) room. You should familiarize yourself with several controls. They may be in a different place on your specific unit, but they should be there. Always use your manufacturer's literature prior to working on or maintaining an automatic injector system.

Controls

We are going to start with the control panel, and then work our way through the overall unit.

Control panel

The control panel displays the injection parameters, which are manipulated by the cardiovascular interventional technologist. The injection parameters include flow rate, rise, total volume, pressure, and delay. The flow rate is the rate at which the contrast medium is going to be injected. It is measured in cubic centimeters per second (cc/sec) and is dependent on factors such as catheter length, inner catheter diameter, number of catheter side holes, injection pressure, and viscosity of the contrast medium. The rise, also known as linear rise, is the time it takes to reach the desired flow rate. Rise is measured in seconds. The total volume is the desired amount of contrast to be delivered. Total volume can be calculated by multiplying the flow rate by the injection duration. Total volume is measured in cubic centimeters. The injection pressure is the force needed for a specific dose of contrast medium. Injection pressure is dependent on the size of the vascular structure and the type of catheter used for the injection. Typically, smaller vessels require lower injection pressures. Injection pressure is measured in pounds per square inch (psi). The range for injection pressures is 100–1,200 psi. The last injection parameter, delay, can be either injection or X-ray in nature. An injection delay allows for image acquisition to begin before the injection occurs. An X-ray delay allows for the injection to occur before image acquisition occurs. This is useful when performing lower extremity arteriography via a catheter in the abdominal aorta and only the vessels distal to the knee need to be visualized.

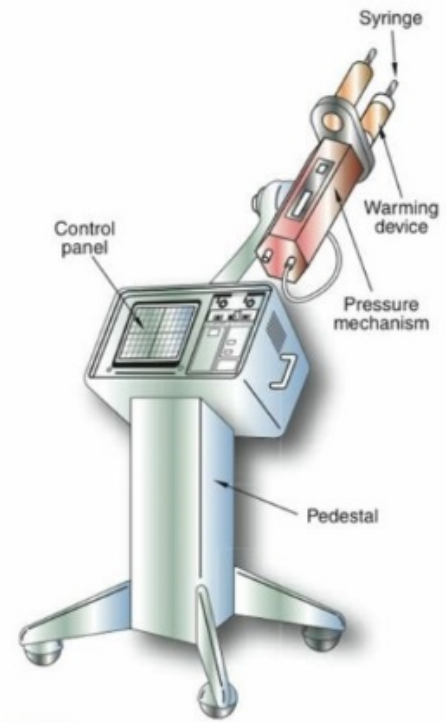

Figure 2–23. Contrast injector. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Syringe

The contrast medium injector syringe is removable and disposable. The contrast syringe capacity can range from 40 to 260 cc. A new sterile syringe is loaded in the syringe-loading assembly prior to each case.

Warming device

Iodine contrast agents become more viscous the higher the concentration of iodine, so some manufacturers offer warming systems to help lower their viscosity prior to injection. The warming device is a thermal sleeve placed over a preloaded syringe to maintain the pre-warmed contrast medium temperature at or near body temperature (37°C; 98.6° Fahrenheit [F]). The warming device maintains temperature but does not increase it from room temperature to body temperature. Therefore, the contrast medium should be warmed prior to the syringe being filled.

Pressure mechanism

The pressure mechanism is an electromechanical motor attached to a jackscrew that drives the piston into or out of the syringe. This piston movement causes the syringe plunger to move forward or backward as needed.

Base and column assembly

The base may contain the electronics cabinet or a standard base. This assembly allows the easy movement of the automatic injector.

Dual-head contrast injector

Dual-head contrast injectors uses a radio-frequency identification (RFID)-enabled system. They feature simultaneous injection and prefilled syringes, offering flexibility for radiology technicians and helping reduce the risk of infection. These units automatically capture drug information and check for expired materials, ensuring patients receive the correct prescribed dose of contrast for the examination

Syringeless injectors

Syringeless power injectors have emerged recently as a solution to reduce contrast media waste. The Joint Commission (TJC) does not allow the reuse of unused doses from single-use syringe injectors, so this option gives facilities the opportunity to use contrast media as efficiently as possible. Its syringe-free injection system is composed of a soft bag injector and associated disposables, using a hydraulic, syringe-free injector to deliver contrast media. These syringeless systems use a bulk 500 mL bolus. The injectors meter out each dose needed. It uses hydraulics rather than an electric motor to power rams, which helps eliminate some magnetic resonance imaging (MRI) artifacts. Some units also include an information reporting system module, which captures injection-related information, including patient identification. This information can be sent to a PACS, recording the amount of contrast used and any reactions observed by the patient, including any extravasation that may have occurred.

Safety

Contrast medium injectors are equipped with numerous safety mechanisms to prevent damage to the catheter and danger to the patient. Function-monitoring devices, such as a flashing light, audible tone, or written message, are used to notify the cardiovascular interventional technologist of any problems. The problem may be an injector malfunction or operational error such as an omission of an injection parameter. A volume-limiting device prevents excessive amounts of contrast from being delivered to the patient. A pressure-limiting device prevents the injection pressure from exceeding a maximum pressure set prior to the injection. Acceleration regulators allow the electromechanical drive motor to accelerate over an exact duration of time to prevent whiplash of the catheter. Finally, the rate-rise control prevents an instant surge of injection pressure by gradually increasing the psi to the preset limit.

614. Digital radiography support equipment

Because DR uses equipment different than traditional radiography, it only stands to reason there will be specialized support equipment that must be employed. In this lesson, we will discuss some of the more common equipment items that go along with DR: the CR plate reader, laser imager, and diagnostic imaging monitor. Let’s start with the CR plate reader.

CR plate reader

Hopefully, you haven’t forgotten the information you read about DR from unit 1, specifically the material on CR. With this system, you should remember special plates are used to capture the X-ray image, and the plates are placed into a CR plate reader to extract the image from the plate and send that image to a monitor or laser printer. Now, we need to focus on the CR plate reader.

As we discussed previously, the CR plate contains a phosphor layer, which is doped with trace amounts of impurities to alter its crystalline form and physical properties. When irradiated, the enhanced phosphors absorb and store the X-ray energy, as electrons, in the gaps of its altered crystal structure. This trapped energy makes up the latent image. Follow along with the graphic representation in figure 2–24 to see how the CR plate reader process works. When the CR cassette is fed into a CR plate reader, the actual plate is extracted from the cassette inside the machine. Next, a helium-neon laser beam scans the CR plate. The energy from the laser, which is of a longer wavelength than the characteristic emissions of the phosphor, is absorbed by the trapped electrons, forcing them to move from their gaps. This results in a release of energy in the form of a light photon from each electron. This light emission (or luminescence) is detected by a photomultiplier tube,

which then generates an electrical signal proportional to the amount of light received. This signal is then amplified and converted to a digital signal through the ADC, which is finally sent to a computer for processing.

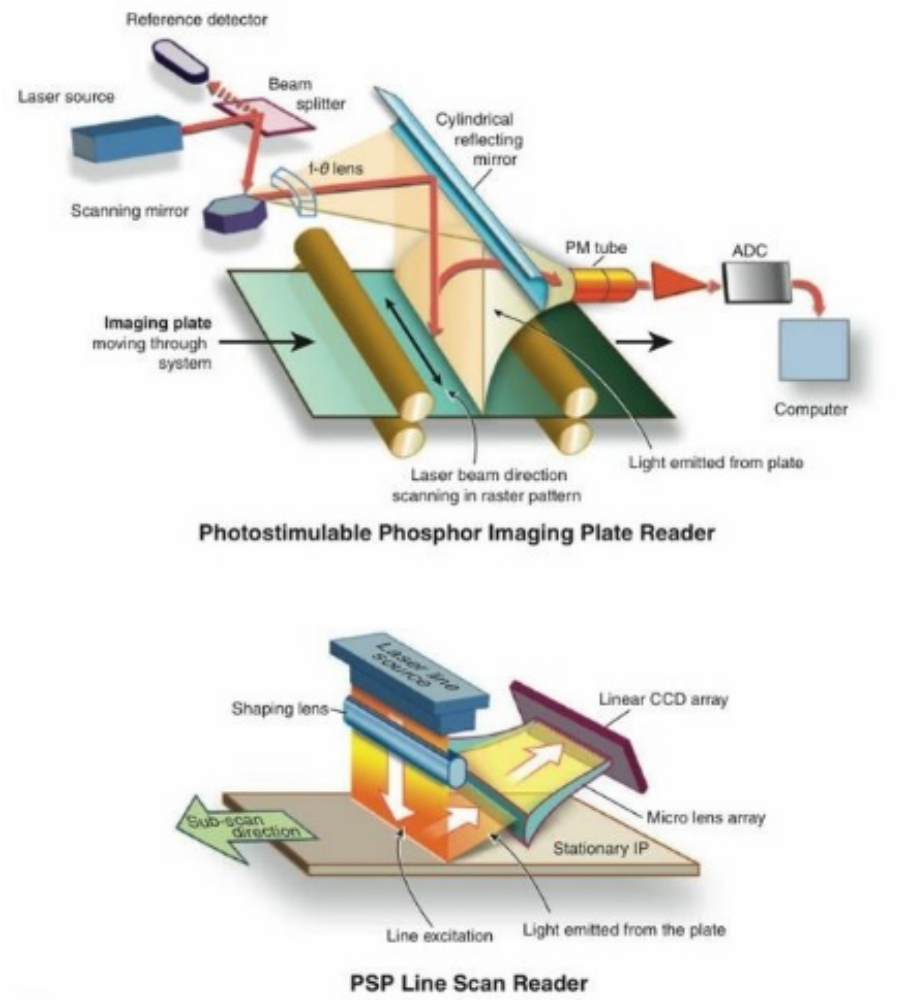

Figure 2–24. CR plate sequence of events. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

After the CR plate is read, the plate reader shines a bright white light on the imaging plate to erase any part of the image that is left. Finally, the CR plate is reloaded back into the cassette and it is ready for reuse. One of the many benefits of a CR-type X-ray system is you don't need to handle the cassettes with the same degree of care that you would with traditional X-ray cassettes. That is not to say they don't require some degree of caution, however. The imaging plate may be taken out in daylight conditions without fear of losing the image, although this is normally done within the CR plate reader. Also, it is important the cassette be read within a reasonable amount of time after exposure (usually less than an hour). This is because the quality of the image will begin to deteriorate over time; after only 2 hours, up to 20% of the image will be lost due to spontaneous luminescence within the plate. One other important note to remember is CR plates that have sat unused for several days should be erased before use. If not, there may be an increased amount of base fog in the exposure.

Many CR plate readers have a built-in preview screen that allows the radiographer to ensure all of the necessary anatomy was included in the X-ray image. There are also units that can accommodate multiple CR plates at one time to speed processing times even further. These are just two of the

advances in CR plate readers; just as CR X-ray units are employing advanced technology, CR plate readers are doing the same. Now that we have an image, let's talk about printing out a copy of the image.

Larger units typically allow stacking and loading of multiple cassettes for use in a large department, whereas a small processor will only process one plate at a time and is typically used in an office or single room. Plate readers are also used with some mobile units. Plate throughput can average anywhere from 30 plates/hour to over 200 plates/hour depending on the type of processor used. Workload and cost will determine which unit will best serve department requirements. There are also tables and upright Bucky units that contain multiple PSP plates and a reader in the unit, so no cassette handling is required. The units move and process plates after each exposure, allowing the radiographer to view images in a monitor within the control booth. Fuji offers these combined units, which can process over 200 images per hour.

Laser imager

A laser imager is basically a printer for X-ray images. There are two types of laser imagers: wet and dry. As you might imagine, their names suggest how the images are processed. In the wet imager, films are produced with chemicals much like in traditional film processing. In the dry imager, films are produced without chemicals. Most laser imagers can be used with traditional X-ray cassettes and act like daylight processors; however, these units are most often used with digital X-ray systems and that is why we will focus on them in this lesson. Let's now look at the two individual types.

Wet laser imager

In the wet laser imager, a piece of film is taken out of a supply cartridge and exposed with a laser beam in a z pattern to place the digital image on the film. After exposure, the film goes through a conventional development, fixation, and washing procedure before being delivered at the receiving tray. The one disadvantage of this type of imager is it still uses chemicals for the film processing. This means it also requires the plumbing connections and tedious cleanups associated with wet film processing.

Dry laser imager

In the dry laser imager, films are processed in a fashion similar to the wet laser imager, except it doesn't use any chemicals and the laser exposes the film in two directions instead of only one to achieve the higher level of exposure required for this type of film. Generally, a laser imager, regardless of the manufacturer, has the following components: a laser source, collimator lens or beam-shaping optics, a beam modulator, deflecting mirror-shaping lenses, and a cylindrical reflection mirror (fig. 2–25). The imager also has an unexposed film storage with a mechanical transport to move the film through the printer as it is exposed line by line in a raster scan pattern by the laser.

A constant-intensity laser source is focused by a collimator lens, modulated according to the digital value of the image pixel, and directed at the film. The F-theta/toroidal lenses and the cylindrical mirror modify the shape of the beam to minimize distortion toward the periphery of the image. In the final step, the dry laser film is exposed to a controlled heat of about 140°C for a few seconds, which transforms the latent image into a permanent image—this eliminates the need for chemicals. These imagers are often referred to as "dry-view imagers" because the film does not contain emulsion and there are no chemicals involved in the process. The image is printed with a thermal head and, because the film does not contain the traditional emulsion, it can be handled outside of a darkroom in daylight conditions.

As we said earlier, a laser imager is used in DR whenever a hardcopy image is needed. Once the image is produced, it can be viewed on a view box like a traditional X-ray image. Because these units are most often used with digital X-ray systems (CR and DR), they normally have some type of electronic input. This input can be from a network connected to any number of DR modalities. The input can also be non-networked, such as a direct connection from a CR plate reader.

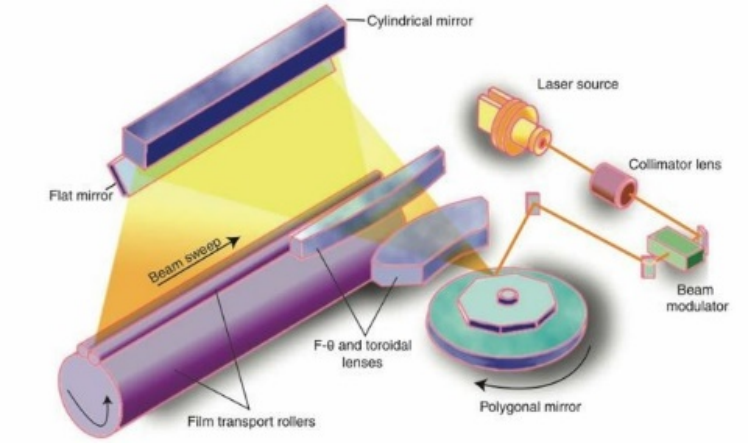

Figure 2–25. Image digitizer sequence of events. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

As you might imagine, dry laser imagers are the preferred imager and have become a standard item in radiology departments. These imagers are an important part of any digital system and generally produce excellent reproductions of an image. They are low maintenance and some can produce as many as 180 images per hour. Installation and maintenance are simple, since a water supply and drains are not necessary. Also, because it is a completely dry development process, fluid or solid chemical waste is avoided and; therefore, there are no disposal problems which reduces installation and operating costs. As with conventional films, the dry film can be recycled to extract the embedded silver, which makes this method environmentally friendly.

Diagnostic imaging monitors

Hard-copy X-ray films were generally analyzed with view boxes (or light boxes), which illuminated the image by shining a light through the back of the film. Soft-copy display refers to images visualized on diagnostic imaging monitors, usually flat panel technology, although there are still some older cathode ray tube (CRT) monitors in use.

Clinical application

A great advantage of digital imaging is the ability to manipulate the image without re-exposing the patient. Images are best manipulated when they are displayed on a monitor. Additionally, images can be viewed in multiple locations simultaneously and can be quickly distributed to multiple locations when in digital format. Monitor characteristics are determined by the requirements of the user. Technologists need to evaluate an image prior to sending it to a PACS, whereas radiologists need to use images for diagnosis. Physicians will look at images either in the hospital or at remote locations. All of these uses will not need the same display quality. Diagnostic viewing by the radiologist requires a monitor with the highest spatial and contrast resolution (along with higher cost), whereas clinical viewing stations for physicians may be acceptable with lower resolution and less cost. Radiographers require monitors that will allow adequate visualization of contrast and resolution to determine if the image is of diagnostic quality for the physician.

Monitor characteristics

There are two types of monitors on the market that are found in most imaging departments: the CRT monitor and the liquid crystal display (LCD) flat panel monitor. The resolution of a monitor is based on how many pixels can be displayed in the horizontal and vertical dimensions. A 1k CRT monitor would display 1,280 x 1,024 pixels, and a 2k CRT monitor would display 2,048 x 2,560 pixels. The higher the number of pixels, the higher the resolution. LCD monitors may have resolutions as high as 3–5 megapixels. Typically, radiographers and physicians use lower-resolution monitors, whereas the radiologist work-stations provide higher resolution. Image quality on a monitor is affected by resolution, luminance, contrast, uniformity, and glare. Because these areas can affect the technologist

and radiologist when viewing images, a QC program needs to be performed on a regular basis to ensure high quality is maintained.

All monitors in a workstation should have the same level of luminance. However, detail of a monitor with a large matrix may still be poor due to electron spot size changing or "blooming," causing pixel size to become larger than the nominal or state size. This blooming is affected by luminance and monitor age.

Contrast should be set the same on all monitors so that the same grayscale is consistent throughout the department. Ambient light reflections on a monitor will have a negative impact on contrast. Monitors have design features to reduce this effect; however, it cannot be eliminated. Therefore, monitors should not be used in rooms with bright lighting. Acceptance testing will allow an acceptable test pattern image to be stored that can be retrieved for testing annually. Contrast is also a function of luminance, and the monitor must have adequate amounts of light from all areas.

Soft-copy veiling glare, often referred to as "glare," results from the internal process of a monitor, rather than external light causing a reflection off the monitor screen. Glare will also cause a decrease in contrast, producing lower image quality. There are a number of factors that cause glare; for example, light scattering on the faceplate, light leakage, and electron backscattering. All of these actions will cause light spread, which will decrease contrast.

LCDs work by blocking light as opposed to creating light like the CRTs. Monitors have fluorescent tubes behind the LCD panels, these are called backlights (fig. 2–26). The back lights deteriorate over time but can be replaced more cheaply than by purchasing a new monitor. The LCD monitor regulates the image by using a light source behind the screen shining on individual pixels. This controls the amount of light transmitted through at varying levels. Liquid crystal and hydrogenated amorphous silicon (a-Si:H) TFTs are contained between glass plates on the front of the monitor, which regulate pixel transparency. This allows rapid changes in voltage to alter a pixel from black (no light transmitted) to transparent (full light transmission).

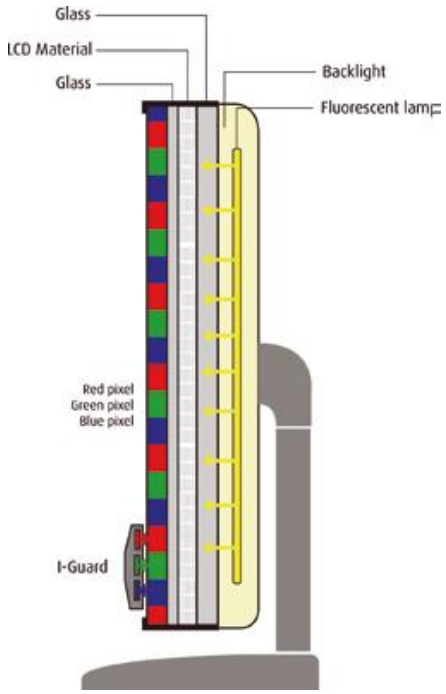

Figure 2–26. LCD monitor.

The polarized glass polarizes the light from the backlight as it passes through. The common electrode plane is a transparent conductive surface divided into pixel sized areas. The layer of liquid crystal

substance that will either pass or block light depends on the charge between the common electrode plane and the top electrode. The top layer of the polarized glass is oriented 90 degrees out of phase with the bottom polarized glass layer.

Light enters the first layer of polarized glass which polarizes the light so that all wavelengths are now in the same plane. Polarized light strikes the first of the liquid crystals in the LCD matrix and the crystals are aligned in a spiral configuration between the transparent electrodes.

Each pixel is made up of three subpixels that is tinted to create different colors or shades. In a color monitor these pixels are red, green and blue. In a monochrome monitor the three subpixels are three varying shades of gray.

LCD monitors have restricted fields of view. This is the angle from which the display screen can be viewed. A special problem of LCD displays is that the field of view gradually deteriorates as the viewer moves away from the center of the screen. The LCD requires quality testing on a regular basis, starting with acceptance testing and continuing periodically throughout the life of the monitor. Acceptance testing of new equipment will establish a baseline for QC measurements that will be a reference for annual testing.

Test patterns

A number of test patterns are required to evaluate the performance of display devices. Some test patterns are just for visual checks, others can be used to make optical measurements. To make optical measurements, you can connect an optical sensor. However, in testing a display device, it is preferred that the patterns be viewed using the display application that is used clinically. When displaying these patterns, no special processing functions should be applied. Furthermore, for most patterns, it is essential to have a one-on-one relationship between the image pixels and the display pixels. Routine visual evaluations of performance are conveniently done using a single comprehensive test pattern. The TG18-QC pattern is recommended for an overall display quality assessment (fig. 2–27).

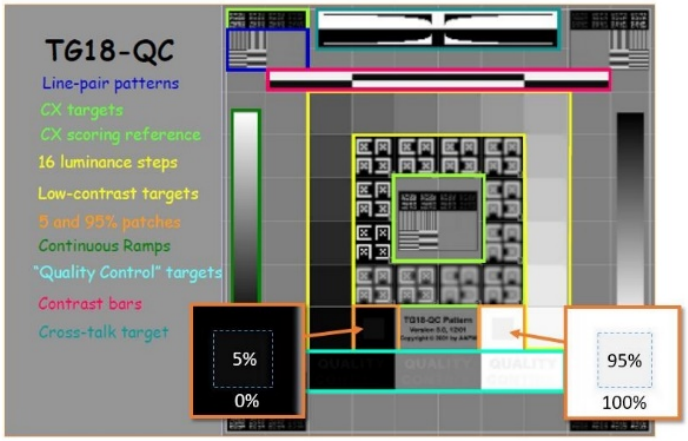

Figure 2–27. TG18-QC test pattern.

The TG18-QC test pattern consists of multiple inserts embedded in a mid-pixel value background. The inserts include the following:

1. Line-pair patterns at the center and four corners at Nyquist and half-Nyquist frequencies for resolution evaluation, having pixel values at 0–255 and 128–130.
2. “Cx” patterns at the center and four corners with pixel values of 100, 75, 50, and 25% of maximum pixel values against a zero pixel value background, for resolution evaluation in reference to a set of 12 embedded scoring reference with various amounts of Gaussian blurring applied.

3. Sixteen 102 x 102 (1k version) luminance patches with pixel values varying from 8 to 248 (in 8-bit version) and 128 to 3968 (in 12-bit version) for luminance response evaluation. Each patch contains four small 10 x 10 corner patches (1k version) at ±4 of pixel value difference from the background, +4 in upper-left and lower-right, -4 in lower-left and upper-right. The small patches are used for visual assessment of luminance response. Additionally, two patches with minimum and maximum pixel value are embedded containing 13, and 242 pixel value internal patches, similar to 5% and 95% areas in the Society of Motion Picture and Television Engineers (SMPTE) test pattern.
4. Grid lines (one pixel) with thicker lines (three pixels) along periphery and around central region, for the evaluation of geometric distortions.
5. Contrast-detail "QUALITY CONTROL" letters with various contrasts at minimum, mid-point, and maximum pixel values for user-friendly low-contrast detectability at three luminance levels.
6. Two vertical bars with continuous pixel value variation for evaluating bit-depth and contouring artifacts.
7. White and black bars for evaluating video signal artifacts, similar to those in the SMPTE pattern.
8. A horizontal area at the top-center of the pattern for visual characterization of crosstalk in flat panel displays.

Prior to evaluation, the display device should be warmed up for approximately 30 minutes. The pattern should be evaluated for distinct visibility of the 16 luminance steps. You should be able to clearly identify the presence of the 5% patch inside the 0% patch, as well as the 95% patch inside of the 100% patch. You should also evaluate the continuity of the continuous luminance bars at the right and left of the pattern, the absence of gross artifacts, and the proper size and positioning of the active display area. Any adjustments to vertical and horizontal size must be made prior to performing the luminance measurements. Dust and smudges on the face of the display will absorb, reflect, or refract emitted light possibly resulting in erroneous test results. In addition, newly installed displays are sometimes covered with a protective plastic layer, which upon removal can leave residual marks on the faceplate. Before testing a display device, the cleanliness of the faceplate should be verified. If the faceplate is not clean, it should be cleaned following the manufacturer's recommendations.

The geometric distortion of a display system is ascertained visually using the TG18-QC. The patterns should be maximized to fill the entire usable display area. For displays with rectangular display areas, the patterns should cover at least the narrower dimension of the display area and be placed at the center of the area used for image viewing. The patterns should be examined from a viewing distance of 30 cm. The patterns should appear straight without significant geometric distortions, and should be properly scaled to the aspect ratio of the video source pixel format so that the grid of the TG18-QC pattern appears square. The lines should appear straight indicative of proper linearity without any curvature or waviness.

Auto calibration

Some diagnostic imaging monitors have built in real-time sensors similar to the I-Guard sensor from the Barco monitor in figure 2–26. These sensors are embedded optical precision photometers positioned at the front of the LCD screen. Contrary to backlight sensors, they measure values from the point of view that the user sees. These sensors perform periodic auto calibrations. The sensor will measure the output from the monitor and relay the feedback for calibrated corrections and image stabilization.

Photometer

The photometer, or "puck," is used to measure the luminance and illuminance of diagnostic imaging monitors. The intensity of light emitted by a display, or luminance, is typically expressed as candela

per square meter (cd/m2). The relationship between cd/m2 and the older luminance unit foot-lamberts (fL) is given by 1 cd/m2 = 0.292 fL. Because LCDs are non-Lambertian surfaces, luminance is typically measured with a photometer, which has a narrow light acceptance angle (≤5°). Some photometers have a display to provide luminance measurements to the operator. Others interface with a computer, which can provide electronic data capture and display the luminance values. Often, medical imaging grade LCDs utilize a calibration software package that supports a limited number of photometers. For calibrating medical imaging grade displays, a photometer should be accurate to within 5% except for very low luminance levels (0.5 cd/m2) where an accuracy of 10% may suffice. Note that photometers used for calibrating color LCDs have independent red, green, and blue (RGB) measurement channels. This provides for calibration of the color tone as well as the luminance of color displays.

A photometer is an appropriate device to measure the luminance emitted (or reflected) from a display. Environmental lighting surrounding a display produces ambient light, otherwise known as illuminance, which by diffuse or specular reflection into the observer's eye, may influence the physical contrast and the perceived contrast of a display. The typical unit of illuminance is the lux (lx). As opposed to the directionally oriented light emitted from a display, ambient light incident upon any object may come from any direction. Therefore, it cannot be measured directly with a standard narrow-angle photometer. Rather, it is measured using a photoreceptor behind an optical cosine diffuser. Whereas ambient light cannot be measured directly using a narrow-angle photometer, the ambient light reflected from a display itself can be measured with a narrow-angle photometer.

The luminance response and the luminance uniformity quantitative tests require a calibrated photometer to measure the luminance of the display device. A luminance uniformity test measures the light from the center and outer corners of the monitor. This will pick up any deviation across the display. There are two types of such devices available in the market. For the near-range type of device, the photometer is held at a close distance from the faceplate of the display (fig. 2–28A). In the telescopic type of photometer, the photometer is aimed toward the display from a distance of about a meter. The measured luminance values vary slightly depending on the type of photometer used, primarily due to the contributions of stray light to the measurements. Otherwise, either type will be acceptable for display assessment as long as the measurements are performed in a consistent manner, which is particularly important for repeated quality control measurements. To maintain consistency, particular attention should be paid to the ambient light level and the use of light blocking devices. For near-range photometers, a stopper ring should be used to block the ambient lighting (fig 2–28B). For telescopic photometers, a baffled cone (frustum) or funnel covered with a black light absorbing coating may be used.

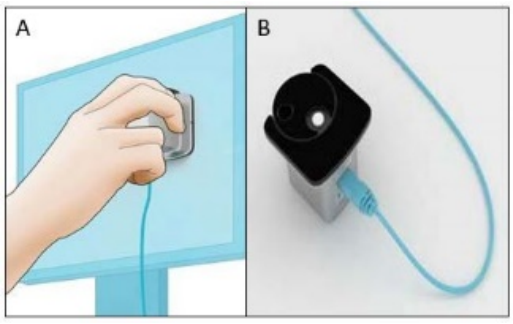

Figure 2–28. Photometer (Courtesy of Fluke/Raysafe).

A complete assessment of luminance response for display systems requires luminance measurements at a large number of signal levels. To automate this process, some manufacturers offer photometers with direct interface to the monitor. The luminance values at multiple signal levels are automatically recorded and subsequently used to calibrate the display.

615. Picture archiving and communication systems

The PACS is a computerized storage and transmission system for handling digital images. The benefit of the PACS is it allows various types of images to be acquired, handled, transmitted, and stored digitally. The necessity of the PACS should be evident to any military member, who over the course of his/her career will travel to several different MTFs. In fact, the early development of the PACS was due to the United States military, as a means to transmit images between various Department of Veteran Affairs (VA) hospitals, and get battlefield images sent to long term treatment facilities. The basic concept of a PACS system is to transmit images across a network, either from within a facility, or across the world.

Digital Imaging and Communications in Medicine standard

One important aspect of the PACS is the Digital Imaging and Communications in Medicine (DICOM) standard. DICOM is a detailed specification that describes a means of formatting and exchanging images and is the standard protocol for transmitting medical images across networked devices. The standard applies to the operation of the interface, which is used to transfer data in and out of an imaging device. DICOM relies on computer industry standard network connections and media devices that address the communication and storage of digital images from diagnostic modalities (i.e., MRI, X-ray, ultrasound, etc.). It also supports the connection of networked printers, such as laser imagers.

A manufacturer uses this standard to design and build a product. The DICOM standard describes all of the detailed functional specifications a device with a communications interface must meet. The standard provides a common reference for all developers, but does not impose a single implementation, thus allowing manufacturers to develop new innovations.

The interface uses a set of computer software, which operates in the imaging equipment or accessory box (or both), and executes the DICOM protocol. The programs also format the data for transmission. This allows easy connection of DICOM-conformant imaging equipment to exist in clinic networks supported with off-the shelf networking hardware and telecommunications services, covering a broad range of networking configurations.

PACS Components

A PACS, as the name hints, is a connection of multiple systems working together to provide communications and image access of medical images. The four basic components of a PACS are the acquisition device, display workstation, network, and storage system. This is obviously the bare bones of a system and with these limited components, you can only shoot an image, display it to the techs or radiologist, and archive the study to an internal or external source. This simple conception of a system does not take into consideration the workflow of a radiology department and in order to understand the purpose of a PACS in radiology, we must learn the workflow! Our standard architecture for systems deployments in the AF include the PACS, the voice dictation system and PACS broker which integrates with the radiology information system (RIS), currently the Composite Healthcare Computer System (CHCS). CHCS transmits and receives information using the Health Level 7 (HL7) language standard, which for our purpose is used to transmit radiology order information and updates into PACS and the dictation system. HL7 refers to a set of international standards for transfer of clinical and administrative data between software applications used by various healthcare providers. If you remember the concepts from Healthcare Information Technology (volume 4 in your 4A251A set), these standards focus on the application layer, which is “layer 7” in the OSI model. CHCS also receives radiology reports from the dictation system.

Acquisition device

An acquisition device is any source that sends images to the PACS. Image acquisition can come from a variety of sources, all of which are required to transmit information using the DICOM standard protocol. Using units that communicate via the DICOM standard protocol is critical to maintaining a connection across multiple modalities (which can come from a variety of vendors/manufacturers).

Display workstation

Any computer monitor that allows digital radiographic images to be viewed is considered a display system. The workstation is used to select images and data for review; view the images and manipulate them as needed; and document findings. Image manipulation includes (but is not limited to) window and level adjustments, image stitching, image annotation, magnification, and pan, scroll, and zoom image visualization features. There are three types of workstations available:

1. Diagnostic workstations—have high-resolution displays and allow radiologists to interpret diagnostic images.
2. Clinical review workstations—have lower resolution and are used outside radiology, typically by referring physicians for viewing images and data. (Many facilities now use personal computers (PC) in place of clinical review workstations.)
3. Specialty clinical review workstations—offer additional specialized functions for use in specific clinical applications.

Each workstation is controlled by a computer and connected to the network.

Network

A PACS system works across a computer network, which is defined as: (1) two or more objects sharing resources and information, or (2) computers, terminals, and servers that are interconnected by communication channels sharing data and program resources. This allows us to take full advantage of the digital nature of the X-ray images. In the PACS network, there may be many different nodes, including workstations, remote workstations, servers, storage devices, and so forth (fig. 2–29).

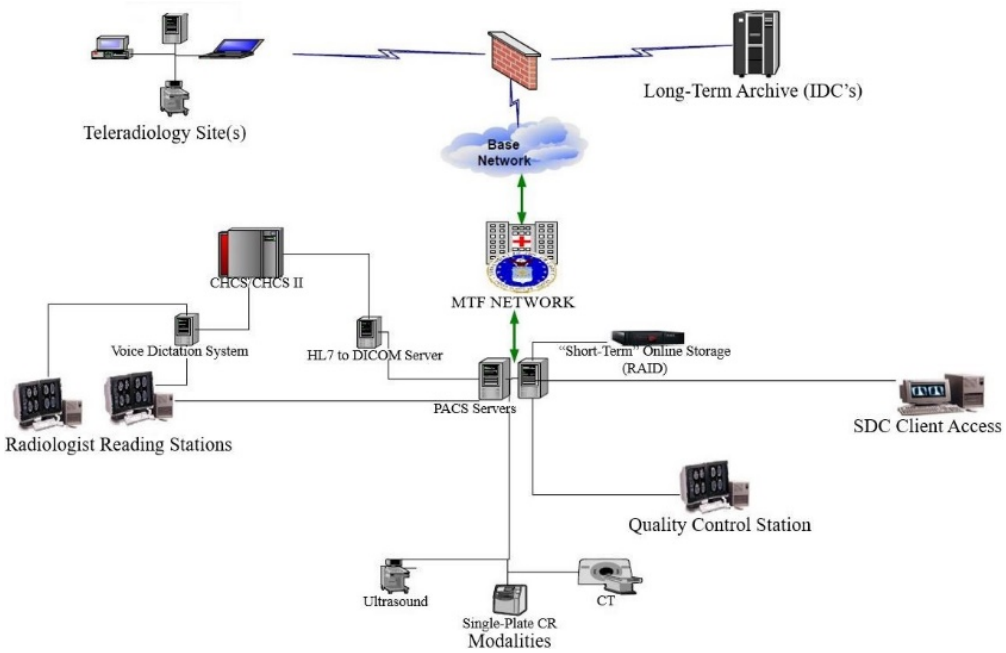

Figure 2–29. PACS network layout.

A PACS system will either be on a local area network (LAN) or a wide area network (WAN). A LAN is what you have within your work facility. Once you complete an X-ray on a patient, let's say on Mr. Dan Marino, his images are available on his ordering provider's computer, the radiologist's reading station or the QC computer for review. These are all local devices. The best example of a WAN is when your facility transmits the images to another MTF to be read by a radiologist at that location. This is known as teleradiology.

Understanding the network infrastructure utilized in a PACS environment is crucial for the proper maintenance and health of the system. Without this copper and fiber backbone, PACS would not be

possible. Every device on the PACS has a network interface card (NIC), which acts as the gateway to other devices and allows for sharing of data. These devices send information to various other gateways like switches and routers. The differences between these two devices are in the way they handle and route data. PACS uses three pieces of information to interconnect properly; Internet Protocol (IP) address, port, and application entity (AE) title. Each server and client within PACS has a unique IP address, which identifies each device to ensure data is sent to the proper destination. The port number is relevant to the protocol that is used to communicate with each device in PACS. The best known protocol is the reserved DICOM port 104 which is used for transmitting DICOM object data. Other protocols can include HL7, Structured Query Language (SQL) database, Hypertext Transfer Protocol (HTTP) 80, and Hypertext Transfer Protocol Secure (HTTPS) 443. The AE title is a case sensitive alpha numeric name unique to a DICOM device. The AE title for sending DICOM images and receiving modality work lists can be different on some vendors for the same type of device. These three pieces of data are minimum requirements for connecting devices into PACS and it is very important to have a list of all your devices and keep track of any address changes.

Teleradiology

Aside from the ability to view images from various locations within a facility on a LAN, teleradiology allows digital images to be transmitted from MTF to MTF using a WAN connection. Teleradiology is the concept of sending digital radiographic images to another location for a radiologist to interpret. This concept enables a radiologist sitting in one location to see images acquired at another different physical location. This innovation has allowed civilian and military radiology facilities alike to increase the amount of exams captured at their facility, even though a radiologist is not on site. In addition, money is saved as many small facilities no longer have to employ a dedicated radiologist to read images daily.

At larger AF MTFs, radiologists provide a teleradiology service, acting as a hub for smaller MTFs and even various deployed locations around the world that do not have a radiologist on site. A good example of this usage is when images are taken in a deployed setting and an injured service member is transferred to the next level MTF with all of his or her radiographic images (CR, DR, and MRI). Having this capability reduces the patient dose exposure levels and saves treatment time since similar exams do not need to be repeated every time the patient is transferred to a different level MTF.

Security Concerns

AF medical and standard Department of Defense (DOD) systems all communicate using the same network back bone, and when it comes to securing the network the security accreditation and certification rules apply in similar form. The threat to our network is very real and if we become complacent by not securing every system that touches our network by hardening or applying the latest security updates (patches) from the manufacturer, it will be left vulnerable to an intruder. This could lead to treats such as an intruder potentially loading a logic bomb, which can trigger a system corruption or shutdown. Hardening an operating system means to apply several security measures like changing default passwords, removing non-essential services or programs, and encrypting files. This is routinely done by the vendor under the guidance of the AF Information Assurance team.

There is an entire team at the Air Force Medical Operations Agency (AFMOA) dedicated to ensuring our systems are safe before they touch our network and stay safe as updates are available. PACS administrators need to ensure that approved patches are being applied on a regular basis. The word approved is very important because this is what separates our medical devices from the rest of the DOD systems. Because our devices are backed by the FDA, they need to comply with certain standards in order to maintain the functionalities necessary to perform procedures on patients.

Storage system

One of the most substantial advantages of PACS is the ability to archive digital information. With a PACS, computer memory replaces the file room where we used to store hard-copy films. This seems simple, but the storage of digital images is the most problematic part of the PACS. Because X-ray

images take up a large amount of memory, storage and memory backup are imperative to a PACS. This is one area that requires continual technological advances to make the PACS effective.

Cache versus long-term storage

The image manager of a PACS is the master-filing system of everything within the storage system, or archival system. It is responsible for receiving, fetching, and distributing the stored images throughout the archival system as well as controlling all the archival system's DICOM processes. Most PACS in MTFs are very large, requiring gigabytes (GB) and terabytes (TB) of capacity.

Once the radiologist reports a study and the report is transferred back to PACS from the CHCS, the study is considered finalized and fully cached. Cache is the PACS' short-term storage and can average from 500GB–7TB of storage space, depending on the workload of the MTF. From the cache the study is sent to long-term storage (or the archive). At this point, two copies of the study exist—one in the cache and one in the archive. The cache follows a simple rule for making room for more studies when the space gets low. The cache starts to delete the oldest accessed study in a manner known as "first-in, first-out." The archives themselves are backed up to a secondary location for disaster recovery purposes.

Good stewardship of image storage

When looking at the long-term storage or the short-term cache, it is not uncommon to use TB in terms of total storage. Just because there are TBs available doesn't mean you should be careless about what is sent into the PACS storage. A computerized tomography (CT), MRI, or ultrasound scan (without additional reformats or cine clips) can easily take up several GBs. Depending on the size of the long-term storage and the total number of exams performed in your MTF, filling up storage within your archival system can happen rather quickly.

The image file size is dependent on the modality type. An average single chest X-ray can be 15 megabytes (MB)–25MB, compared to a CT chest study, which has hundreds of images starting at around 500MB. With this comparison, it is easy to understand how one modality can use up storage space a lot quicker than another modality. It is important to follow the storage policies as outlined by the PACS administrator at your MTF.

Storage redundancies

Within the PACS, there should be a certain amount of redundancy. If, for example, you have a long-term archive that can hold 500TB of digital information, the PACS should be able to make a duplicate of every image. Making a duplicate of every image will cut your storage in half and protect your data if a drive/disk array ever fails.

A redundant system is a system that has failover mechanisms, which prevents it from completely shutting down services. Within the various storage levels in the archival system, redundancy is built into the system, allowing for the copying of data across several other hard drives. This aids in data restoration if one of the drives fails or becomes corrupted. This configuration redundancy is called redundant arrays of independent disks (RAID). While you have already learned the RAID concept in depth in 4A251A, volume 4, let's review some basic concepts as they relate to PACS. New data is stored on the RAID for quick access, while older data is maintained only in long-term storage, from which access is slower (several seconds to a few minutes, compared with less than a second to a few seconds for RAID access). Many archives now use data compression, which reduces an image file's size, as a means of reducing storage costs.

There are two main PACS configurations (levels) used: RAID 1 and RAID 5. RAID 1 requires a minimum of two disks and the data is mirrored to all drives. With this type of RAID setup, an exact system duplicate exists; therefore, if one drive were to go down, the other drive would immediately take over. RAID 5 uses a minimum of three disks. One disk acts as the parity controller that sends bits of information across the other drives. This information is then used to restore a failed hard drive. RAID 5 increases performance when reading and writing to the drives. In addition, if one drive were

to fail, you would notice decreased PACS performance. When a system failure occurs, it will automatically route clients to the working server for access. The user does not know the difference and continues to work without hiccups.

Workflow

Now that you understand the various components of a PACS, let's look at how they work together to create the study workflow. Workflow consists of all the materials, services, and information that are systematically organized to perform a repeatable pattern for doing a task from beginning to end. The radiology PACS workflow begins when the referring or requesting physician enters a radiology order into the RIS for a patient. At this point an HL7 message is sent to the PACS broker, which in turn converts the message into the DICOM format and sends the request to the PACS to start prefetching relevant prior studies for this patient from our regional archives. The patient then checks into radiology for the study, and the receptionist or technologist must verify that an order has been placed in CHCS. PACS uses the HL7 message, which contains the patient demographics, procedure and order history, and makes it available to all of the acquisition devices (modalities). This is known as a modality work list. The technologist pulls up the patient order at the modality and proceeds with the patient exam. Once image and data quality have been verified the study is then sent to PACS. At this point, the current and relevant prior studies are available to the radiologist for interpretation using the voice recognition dictation system. Upon completion of the report, the dictation system uses the radiology order information that was initially served and uploads the final report into CHCS for distribution back to the requesting physician. The patient workflow ends here but the image workflow continues on to the regional archives which makes the study available to the entire AF.

When it comes to workflow, your PACS administrator plays an important role in the flow of information in and out of your digital radiography department. They must understand all of the acquisition devices at your MTF. The type of device will determine how he or she configures the PACS. As an example, let's say a new mammography device is installed at your facility. The administrator cannot simply associate the device to the PACS using an AE title, IP, and port. Yes, the device will communicate and you will see an image appear on the PACS, but the workflow piece of the puzzle cannot be forgotten.

How will the image appear at the radiologist's reading station—like hanging protocols or demographic overlay specific to mammography? Will these images require any type of compression when storing to the archive or sharing through a web server? Do you need to increase your short-term storage space to accommodate the workload increase? These are all questions a PACS administrator has to answer prior to installing any new piece of imaging equipment. Understanding image workflow from beginning to end is the easiest way to see that every PACS user is taken care of.

Self-Test Questions

After you complete these questions, you may check your answers at the end of the unit.

613. Contrast injectors

1. What are the two most common compounds used as contrast mediums?

2. How is the contrast medium introduced during angiographic procedures?

3. How are automatic injectors classified?

4. What is the capacity range of the contrast syringe?

5. Describe the warming device and its purpose.

6. Describe the components of a syringeless injector system.

614. Digital radiography support equipment

1. What scans the CR plate inside a CR plate reader?

2. Briefly explain how light is produced from a latent image stored in a CR plate.

3. What is used to erase a CR plate?

4. How soon after exposure should a CR plate be read? Why?

5. What procedure should be performed on CR plates that have been unused for several days? Why?

6. List the types of laser imagers and the primary difference between them.

7. What is the main disadvantage with a wet laser imager?

8. In a dry laser imager, what transforms the latent image into a permanent image?

9. What are the two types of diagnostic imaging monitors?

10. What is the resolution range of an LCD monitor?

11. What can cause poor image detail of a monitor with a large matrix?

12. What is contained between glass plates on the front of the LCD imaging monitor?

13. How is the top layer of polarized glass orientated?

14. What test pattern is recommended for an overall display quality assessment of diagnostic imaging monitors?

15. What affects will dust or smudges have on a display?

16. What is the function of a photometer?

17. What are the photometer accuracy requirements for calibrating medical imaging grade displays?

18. What does the luminance uniformity test measure?

615. Picture archiving and communication systems

1. What does DICOM stand for?

2. What does the DICOM standard describe?

3. What are the four main components to the PACS?

4. Describe HL7 and its uses.

5. List the three types of PACS workstations available and briefly describe each.

6. List the two definitions of a network.

7. List the three pieces of information that PACS uses to interconnect properly.

8. Which port is used for transmitting DICOM object data?

9. Provide an example of teleradiology.

10. What does it mean to harden an operating system?

11. What is the most problematic part of a PACS?

12. How does the PACS determine the order for deleting images when space is low?

13. How much memory does the average single chest X-ray image use?

14. List the two main RAID storage configurations used in a PACS.

2–4. Equipment Maintenance

We've covered a lot of ground concerning X-ray equipment and its related support equipment; however, you still need some information on the most important aspect of X-ray equipment to the BMET—maintenance. We discuss maintenance at the end of this unit in order for you to have a chance to understand how some of the support elements work together through the X-ray process. This will give you a better comprehension of some of the calibration processes in the following lessons. In this section, you will learn about basic preventive maintenance on fixed and mobile X-ray systems, non-invasive X-ray test equipment, and the post calibration radiation inspection (PCRI). The following information will not make you an expert X-ray maintenance technician, however it will prepare you to go out and "get your hands dirty" the next time an X-ray unit needs to be maintained or repaired. Let's get started with some maintenance procedures.

616. Maintenance procedures for fixed systems

If your MTF has X-ray equipment maintained by the local BMET, you will likely find these systems take up a great deal of your time. Therefore, it is important you perform timely and efficient maintenance so you aren't spending unnecessary time making numerous unscheduled calls during the month. This lesson will cover the operational inspection and proper preventive maintenance for fixed X-ray systems. Although these two inspections (operational and preventive maintenance) are often listed separately, they will be grouped together in this lesson for simplicity and ease of understanding. You will normally perform both inspections at the same time.

The inspection for an X-ray system is much like that for any other piece of medical equipment, except there is much more to check out. It is important to always check every feature and function of an X-ray system during your inspection to ensure they are working properly. Before you begin your inspection, talk with the radiology staff to see if they have any particular complaints or areas of concern that need special attention. Because there are so many items to check on an X-ray system, it will help to break these items down into smaller components.

Ceiling rails and upper tube support

Probably the best place to start your inspection is at the ceiling rails and upper tube support. If you properly inspect and clean this area first, you won't need to go back and reclean the rest of the unit after all the dust and dirt settles. Start off by checking the mechanical items—tighten any loose hardware and replace any screws, nuts, bolts, and so forth that may be missing. Inspect the counterweight assembly and tube mounting hardware. Move the tube assembly through its full range of motion, paying particular attention to any binding. Properly lubricate the mechanical paths or other moving parts per the manufacturer's instructions. Thoroughly inspect all wires and cables for good connections, as well as for signs of deterioration or damage. Ensure all safety interlock switches are working properly. Verify that movement stops at applicable detent positions and detent lights or identifiers are working. Check the source-to-image distance (SID) indicator for accuracy. Last, use a vacuum cleaner to thoroughly clean all parts of the rails and tube support (remember, this will probably be the only time this part of the unit gets cleaned, so you should do a good job).

X-ray table

Now that you have performed the maintenance on the rails and tube support, the next area to work on is the X-ray table. Begin by moving the tabletop through its full range of motion. Next, slide the bucky along the table—it should not bind anywhere along the path. The bucky tray should travel in and out of the bucky smoothly. Check, adjust, or tighten the pin, which activates the size-sensing arm in the bucky. Check the motor drive assembly, including any belts or chains, and lubricate the system per the manufacturer's instructions. While you are lubricating the drive assembly, you should inspect and lubricate any bearings or other mechanical paths for the table. Check all the safety interlocks and table travel limit switches. Lastly, check the table collision sensor by moving the table and activating the collision switches.

Wall stand

The wall stand is used to enhance the number of arrangements available for patient positioning and allows X-rays to be taken with the patient in the standing position. It is a relatively simple portion of the unit, which contains a moveable bucky and bucky tray, as well as a counterweight assembly (if manually operated) or drive assembly (if automatically operated) that require inspection and possible adjustment. Always check to ensure the bucky moves smoothly and in the full range.

High voltage transformer

One of the most important items to check in this portion of the X-ray unit is the high voltage cables. During your inspection, check the cables carefully for any cuts or other damage in the insulating cover. Ensure the power is turned OFF before removing any high-voltage cables. Next, remove the cable ends from the X-ray tube, clean them, and inspect for any signs of damage. Also inspect and clean the tube wells. After inspecting, apply new dielectric compound to the cable connections before reinserting into the tube (properly applied compound should completely surround the connector after it is plugged into the tube well [fig. 2–30]). The dielectric compound forces air out of the tube well, eliminating the chance of arcing. After reinserting the connector, tighten the retaining ring until the rubber gasket is tightly compressed with the appropriate cable wrench.

NOTE: Ensure you properly discharge the high-voltage cables before handling by repeatedly touching the conductors to ground as soon as they are removed from the well.

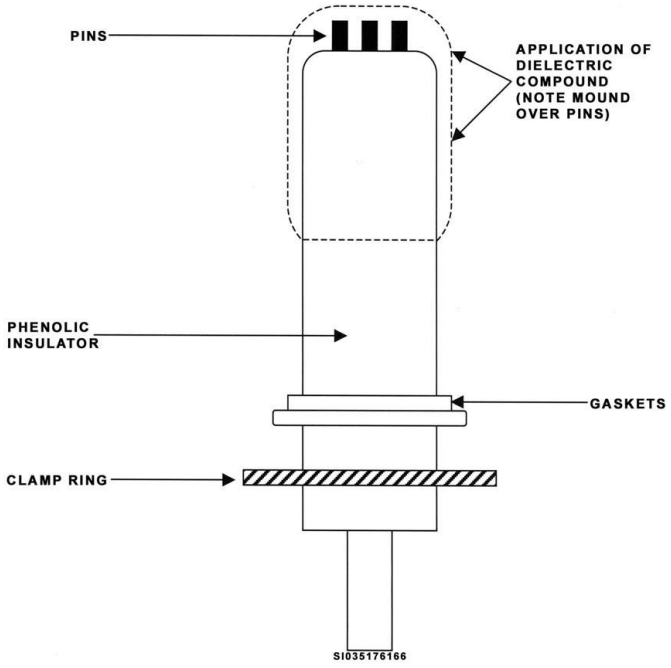

Figure 2–30. Proper application of dielectric compound.

The next step in cable maintenance is to remove the cables from the high voltage transformer. This end of the cable sits in the transformer oil and dielectric compound is therefore not used on this end. Inspect the cable connectors from this end as you did previously. When reinserting the connectors, a small amount of transformer oil should be added to the cable wells.

The last part of inspecting this component is checking all the interface connections and cleaning up after yourself. The oil and dielectric compound can be rather messy, so always clean up after you complete this step.

X-ray tube

Unless you are calibrating the unit, this inspection is fairly simple. You cleaned and checked the connectors in the previous step; now, you just need to inspect for loose screws or connections in the tube assembly and ensure everything is secure. Next, check the bearings in the rotor by performing an operational inspection and listening for proper sound (any unusual sounds should be investigated and may require tube replacement). Next, ensure X-rays are inhibited until the anode is up-to-speed. Finally, ensure the brake on the high-speed stator (if equipped) is operating correctly and annotate the anode coast down time after an exposure is completed.

Collimator

Check the field lamp and laser pointers to ensure they are illuminating. Next, verify the tape measure is present and accurately measures the SID indicator. Verify the collimator shutter blades varied during collimator control knobs operation. Load a cassette into the bucky tray, and ensure the automatic system is working properly and the collimator blades move freely.

X-ray control unit

From here, you can accomplish an operational inspection. The majority of your work on the control unit is visual in nature. You need to carefully check all connections within the unit to ensure their security. Next, you should inspect all the indicators and any lights/lamps, along with all the switches, buttons, and knobs. You should accomplish a thorough operational inspection to ensure all features

and functions meet the manufacturer's specifications. Lastly, access the Error Logs and document errors or problems identified by the system maintenance programs.

617. Maintenance procedures for mobile systems

Mobile X-ray units travel throughout the hospital, bumping into walls, elevators, and fixed objects. With that amount of abuse, many malfunctions can occur. You must ensure your preventive maintenance inspections are extremely thorough, looking for all possible problems even before they occur. Be sure to always refer to your manufacturer's literature for specific maintenance procedures. For now, here are some general guidelines that are specific to mobile X-ray systems.

Battery maintenance

Because all power for the portable unit is supplied from the batteries, proper battery maintenance is imperative. The batteries in the unit must be kept in top condition at all times. Most units will have a battery status indicator on the front panel so operators can easily see the condition of the batteries. There is circuitry that monitors battery strength, and normal maintenance procedures including the periodic replacement of old batteries.

To ensure the batteries are correctly maintained and maximum life is obtained from them (they are quite expensive), it is important to establish some type of battery charging protocol for the portable units in your MTF. The battery charger should be plugged in whenever the unit is not being used. The charger circuit has a charge complete cycle and a charger cut-off for protection of the batteries. The batteries require monitoring and lots of exercise so they will not develop a memory, an inherent trait of nickel-cadmium (NiCad) batteries. Equalization cycles should be performed annually. If you find that you must perform them sooner (every six months), then the unit is not being exercised enough. This is a very simple problem to solve—just have the technicians take the unit into one of the X-ray rooms daily and shoot at least ten consecutive shots. This exercises the batteries and keeps them from developing a memory.

Mechanical and electrical inspection

Look for any loose panels or sections. Pay particular attention to the mounting of the collimator. With a screwdriver, check for possible loose screws, particularly with the tube support arm and the vertical bearing tracks. With the X-ray tube set to minimum height, check the vertical suspension wire rope for possible broken strands.

> **CAUTION**: Do not test with bare fingers. Instead perform the test by rubbing the cables up and down using a piece of cloth.

Check the action of the tube-stand bearings. Are there any visible gaps between the bearings and the track surface? Also are there any 'clunking' noises or 'jerking' when moved, which can indicate damaged bearings? You should also check for possible loose lock handles, and ensure manually operated locks have an adequate range of adjustment.

Pay particular attention to the cabling from the X-ray tube and tube stand. All movements of the system should not cause any stress or pulling of the cables. Inspect the HT cables for any sign of damage to the safety earth shield at the X-ray tube cable ends. You will want to ensure the cable ends are firmly inserted into the X-ray tube, and the securing ring nut is not loose. Where there is evidence of twisting or pulling on the HT cables, particularly at the X-ray tube end, investigate means of providing additional support. As with most electrical equipment, be sure to carefully examine all plugs and sockets attached to cable ends. The outer insulation of cables should not be pulled out from the cable clamp.

Two commonly damaged items of primary concern during your inspections are the AC power cord and exposure hand switch. The power cord may be run over by the unit, or technicians may move the unit without unplugging it and damage the pins—so pay close attention to this component. The

exposure hand switch often may be pulled, dropped, or otherwise receive rough handling, and should also be carefully inspected for damage.

Operational inspection

One other important consideration within this unit is the drive mechanism, which assists in moving the mobile unit from the X-ray department to the site. The drive system can be adjusted, so when you perform the calibrations, ensure the revolution of the wheels per minute is completed. Adjusting the speed too high will cause an X-ray technician to lose control of a fairly heavy and expensive piece of equipment. Ensure the mobile brakes operate in a positive fashion when the hand is released from the handle, and that they are fully released while the unit is travelling. Where there is an anti-crash bumper, manually operate the bumper. This should stop the motor drive

CAUTION: Do *not* test by standing in front while the unit is moving forward.

Check all indicator lamps and displays. If necessary, operate different selection techniques to ensure all required status indicators operate correctly. Where a digital readout of radiograph settings is provided, select a number of different values to ensure there are no display errors or missing segments. Test the operation of the electromagnetic locks; there should be no hesitation in operation, nor should the lock 'stick on.' In some cases the surface of the lock may require cleaning, to obtain a better 'grip'.

618. Non-invasive X-ray test equipment

Now that you have learned some of the basic maintenance procedures of radiographic systems, you now need to become familiar with the test equipment used during X-ray equipment maintenance. You may be asking yourself, what's the difference between non-invasive and invasive types of test equipment? Well, an invasive piece of test equipment is actually introduced into the X-ray circuit and requires connections to be broken and reconnected. Non-invasive test equipment is simpler to use and does not require any break in connection of the X-ray circuitry. In general, invasive test equipment was considered to be more accurate; however, as technology has improved, that is not as much of a concern as in the past. Also, in some cases, invasive test equipment cannot be used because the introduction of test equipment into the circuitry will negatively affect the operation of the X-ray unit. General testing of X-ray output devices now uses non-invasive test equipment as the preferred method of testing. There are a couple models of non-invasive testers that you might see in AF BMET shops, most notably the RaySafe Unfors units (fig. 2–31) and RTI Barracudas®. Whichever type or manufacturer you use, the basic concepts and functions should be the very similar. We'll start this lesson with descriptions of various detectors and then discuss a range of measurement functions.

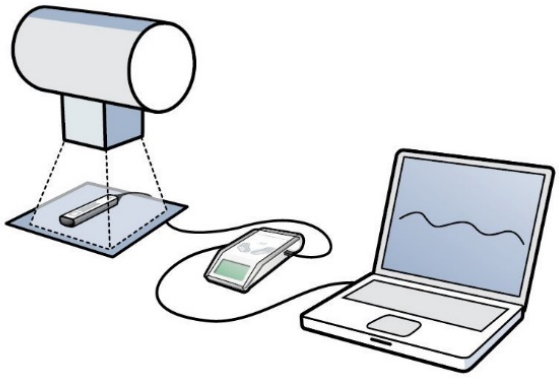

Figure 2–31. Non-invasive X-ray test equipment setup (Courtesy of Fluke/Raysafe).

Detectors

Your tester should contain a base unit that you can use for all testing functions as well as a variety of interchangeable external detectors, each with different functionality (fig. 2–32). Communication between detector and the base unit is purely digital, thereby minimizing sensitivity to mechanical or electrical stress. The base unit may also be equipped with an optional integrated tube current meter (mA/mAs). The base unit automatically identifies the connected detector and displays the settings and parameters specific to that detector. There is a built-in active compensation that automatically applies corrections for different beam qualities, filtrations, and temperatures. During fluoroscopy, survey (leakage), or light measurements, the displayed values are continuously updated as these measurements are of longer duration and considered real-time values.

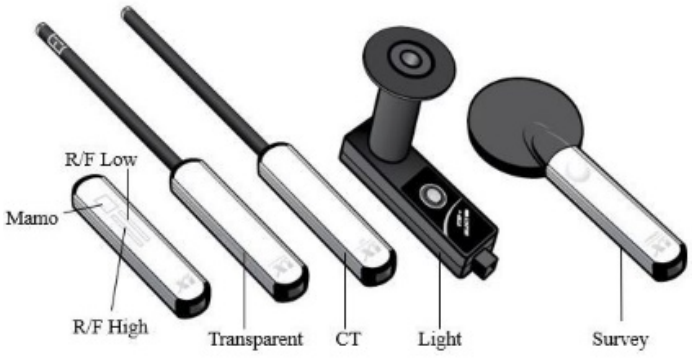

Figure 2–32. Non-invasive X-ray detectors (Courtesy of Fluke/Raysafe).

Radiography/fluoroscopy

The radiography/fluoroscopy (R/F) detector has two sensors: R/F high is designed for conventional, high dose rate measurements normally generated without a phantom between the detector and the X-ray source. This sensor is used for values higher than 1 mGy(gray)/s (7 R/min). R/F low is designed for low dose rate measurements normally generated with a phantom between the detector and the X-ray source with measurements lower than 1 mGy/s (7 R/min).

Mammography

The mammography sensor is used for both low and high dose rates generated in mammography applications. Mammography applications have a dedicated holder designed for use with the mammography unit to ensure proper placement of the sensor and protection from damage caused by the compression device.

Transparent

The transparent detector is a solid-state detector with a very small radiologically visible footprint. It is intended to be used for applications where the R/F detector would influence the AEC on X-ray equipment and serves as a complement to the R/F detector. The transparent detector is designed to mimic the response of a pancake ion chamber detector and therefore it has no backscatter protection.

CT

The CT detector is an ionization chamber, designed to measure CT dose for applications such as dose length product (DLP) and computed tomography dose index (CTDI).

Light

The light detector is designed to measure luminance on LCD and CRT monitors, as well as light boxes, and illuminance for diagnostic X-ray applications. This detector attachment holds the same characteristics as the “puck,” as previously explained in the diagnostic imaging monitors section.

Survey

The survey detector is designed for measuring leakage or scattered radiation from X-ray tubes or in examination rooms and leakage radiation from γ-emitting isotopes. This probe measures radiation at very low levels.

Measurement values

Non-invasive test equipment measures a range of different parameters concerning the output of radiation devices. From the intensity of radiation, to duration of exposures, as well as dose to the patient, these values are critical to determining the proper function of an X-ray unit. This should give you a better working knowledge of the concepts used for calibration.

Dose Rate

Dose is the energy absorbed per unit mass, usually expressed in terms of Gy. Non-invasive devices measure radiation (dose) with an air filled ionization chamber or solid-state radiation detector.

Ionization chamber

The ionization chamber consists of an air-filled chamber between two collecting electrodes (anode and cathode) with a difference of potential bias of approximately 300V. This type of detector works when X-rays react with the air in the chamber, releasing free electrons and ions. The electric field created by the potential difference between the anode and cathode causes the negatively charged electrons to move to the anode, while the positively charged ions are drawn to the cathode. This movement of electrons and ions causes a small electrical current, which is directly proportional to the X-ray exposure. This small electrical current provides the energy to drive an indicator. The CT detector is an ionization chamber with a carbon fiber housing. It has a flat energy dependence curve and automatic pressure and temperature correction.

Solid-state detector

The solid-state detector is made of a specially doped silicon material. When the detector absorbs X-rays, electron hole pairs are created. Of course, the number of electron hole pairs created is directly proportional to the level of X-ray energy. The electron hole pairs then move to opposite sides within the detector and create a small electrical current relative to the X-ray exposure. This small current provides the energy to drive an indicator. The R/F, mammography, and transparent detectors measure dose with a multi-segment solid-state sensor, which use active compensation features to automatically correct the displayed dose (and dose rate) for beam qualities with varying levels of filtration.

kV/kVp

The peak voltage applied across the anode and cathode of an X-ray tube during an exposure is called kVp. During X-ray generation, the surface electrons released from the cathode by thermionic emission accelerate towards the anode target using an applied voltage (kVp), ultimately producing X-rays. Thus, the kVp corresponds to the kinetic energy imparted on the electrons which are accelerated through the tube, and is proportional to the peak energy of the resulting X-ray emission spectrum. An oscilloscope can measure the kVp waveform which illustrates how the applied kVp varies over time during an exposure. If the X-ray tester is connected to a computer, you can also see the waveform represented on screen for the measured exposure.

The tester calculates kVp on the R/F high sensor if the signal level is high enough, otherwise kV average will be displayed. If no kVp value is displayed, try increasing mA. Calculations for kVp are conducted within a kVp measuring window. The kVp measuring window begins after a trigger delay and kVp delay and is approximately 160 ms wide.

For high frequency or DC like waveforms, kVp matches kV. If there is a ripple on the waveform, a kV value will be lower than kVp. Use the kVp delay function on machines with a slow rising output; 150 ms on dental single-phase machines and 1000 ms on a fluoroscopy machine is recommended.

Time

Time is measured from the starting trigger of an exposure until the signal falls below 25 % of the peak (50 or 75 % if adjusted with a trigger level setting). At low dose rates (about 1 % of the max dose rate for the active sensor), the 25 % end level is changed to a low level (about the lowest measurable dose rate for the active sensor).

If the radiation has a pulsed characteristic, the time is measured until the last pulse ends. The dead time interval between pulses must, however, be less than the delay time (0.5, 2, 4, 6 or 7 s). Single phase generators (normally dental applications) may have slowly increasing amplitude characteristics. As with the delay function, it is recommended to activate a higher trigger level after the first exposure.

Pulse

Pulses are counted from 1 to 9999. The trigger to increment the pulse counter occurs when the dose rate has a negative slope and the amplitude falls below 25 % of the peak amplitude for the exposure. At peak dose rates of approximately < 3 µGy/s (for R/F low) the signal to noise ratio is too low to count pulses. If possible, increase the mA and/or kVp to increase the radiation output.

Frame rate

The unit calculates frame rate as: (number of pulses -1)/(exposure time in seconds). If pulses cannot be counted, the frame rate cannot be calculated.

Dose per frame

Dose per Frame is calculated by (accumulated dose)/(number of pulses). Like frame rate, if pulses cannot be counted, the Dose per Frame cannot be calculated.

Half value layer

Half value layer (HVL) is an indication of the beam quality and is defined as the amount of aluminum (Al) filtration, measured in mm, needed to reduce the dose in half. HVL is kVp dependent. HVL should not be confused with total filtration, but used to estimate the total filtration of the X-ray beam.

Total filtration

Total filtration is a calculation of the amount of filtration between the X-ray source and the patient, expressed in equivalent amount of millimeters of aluminum (mm Al). Total filtration is the combined effect of inherent and added filtration. A minimum total filtration of 2.5 mm Al is required for X-ray tubes operating above 70 kVp. The total filtration value measured may differ from the filtration stated by the X-ray manufacturer, as there might be additional filtration not specified by the manufacturer.

mA and mAs

The measure of mAs is the product of the electrical current (mA) used to produce the X-rays and the exposure time (s), and is related to the total number of X-ray photons produced. While this is an invasive test process, some non-invasive X-ray measurement devices can be equipped with adaptors and accessories to measure mA and mAs.

Measurement techniques

The primary difference in the measuring technique for the various X-ray modalities is the tube/filter combination. For example, for unit measure standard radiographic exposures, it will generally be setup to measure a unit with a tungsten tube and a total filtration of 2.5 mm AL with a kVp range of 60 – 150. Again, the way this technique is achieved will vary depending on the operation of the unit. For example, if the unit is self-contained, you will probably select from a series of buttons or touch keys to achieve the proper set up. If the unit requires the selection of various filter combinations, you will need to choose the combination that matches the filtration of the X-ray unit. Some of these units come with preselected “filter packs” that contain the exact filtration combinations needed for various

X-ray modalities. You may also need to select the appropriate radiation detector for the X-ray modality being measured.

Filtration is the main concern with non-invasive equipment, because kVp readings can be affected by filtration of the X-ray beam and sometimes requires correction to account for this filtration. Some units come with software that enables the unit to perform this correction automatically, while others require manual calculation. Either way, the manufacturer's literature should address this topic and tell you how to handle necessary corrections.

619. Post Calibration Radiation Inspection

A PCRI ensures the radiation output of an X-ray unit is within acceptable safety standards, and must be performed annually by the local BMET shop or regional MERC on all X-ray systems within the MTF. Results of the inspection are recorded on a PCRI form, which can be found on the AFMOA website. There are numerous PCRI templates available, such as one for radiographic equipment, fluoroscopic equipment, mobile radiographic equipment, and so forth. In this lesson, we are only going to cover the PCRI for a fixed radiographic unit.

The spreadsheet form is designed for use on a laptop computer and saves time by performing the necessary mathematical calculations required during testing, as opposed to manually calculating the values and equations in the past. You can add a printed or digital copy of the form into the units equipment data file (EDF) once you enter all appropriate information. We will break the form up into various sections and go through each one of the tests explaining some of the theory, test set ups, and measurements as appropriate. The form that your local shop or MERC uses, and the testing set values might be slightly different from what we will cover, but the purpose of this lesson is to familiarize you with the concepts and reasons behind the various tests.

Equipment information

These sections ensure your data quality information is accurate and lists identifying characteristics about who is performing maintenance, the equipment specifications, and test equipment used to validate calibration.

Facility information

The first section is self–explanatory (fig 2–33). It includes the name of the facility where the inspection is taking place, the date, time, index number/equipment control number (ECN) of the system under test, whether the unit is under contract, the site ID, and name of the inspecting agency (your regional MERC for instance).

System identification

This is where you identify the manufacturer of the system (i.e., General Electric, Phillips, or Siemens) and the maximum tube potential for kVp and mA (i.e., 150 kVp, 1000 mA). You will also identify the system configuration (fixed radiographic system, mobile radiographic system, or part of a radiographic/fluoroscopic system), whether the unit is a mAs delivery unit (a mAs delivery unit will not allow for the separate selection of mA or time by the user), and finally the tube serial number.

Test equipment data

This section ensures that you identify the test equipment used and list the calibration dates to validate that your testing sources are arcuate and in compliance. Here you will list the quality assurance (QA) meter, mAs meter, and light meter's ECN, make, model, and calibration dates.

POST-CALIBRATION RADIATION INSPECTION RECORD
RADIOGRAPHIC
Form Current as of: 2 Mar 2015 [Derived from AF IMT 2025, 198207011, V1]

FACILITY: 99th MDG, Nellis AFB, NV (Site of Equipment) Under Contract or Warranty? (YES/NO) Yes
DATE 22-Mar-2018 TIME: 1730 ECN: 25440 SITE ID:
INSPECTING AGENCY: MEDICAL EQUIPMENT REPAIR CENTER, TRAVIS AFB, CA

I. EQUIPMENT IDENTIFICATION

A. CONTROL MANUFACTURER: GE Discovery XR656 (Make and Model of Equipment)
B. MAX TUBE POTENTIAL 150 kVp 800 mA
C. X FIXED RADIOGRAPHIC ___ MOBILE RADIOGRAPHIC
D. DISPLAY ONLY mAs No
E. TUBE SN: 105671BI2

II. QUALITATIVE MEASUREMENTS

TEST EQUPMENT DATA

QA DEVICE:	ECN:	44217	Mfr:	Unfors	Model:	Xi	Cal Date: 17-Nov-17
mAs METER:	ECN:		Mfr:		Model:		Cal Date:
LIGHT METER:	ECN:	40957	Mfr:	Extech	Model:	Easyview	Cal Date: 1-Sep-17

Figure 2–33. PCRI, facility and system identification.

Timer accuracy

This is generally the first test performed on a PCRI. It verifies the accuracy of the exposure timer from the shortest duration available to the longest selection available. To perform this test, select several settings on the radiographic timer and enter these in the appropriate “SETTING” blocks. Be sure to select a range of times from the unit’s timing selector, and select both the shortest and the longest available times as two of the test points. Ensure that you collimate your X-ray field to your detector and set your meter up to read exposure time, displayed in milliseconds (ms) (fig 2–34). Initiate an exposure at each of the selected settings and record actual exposure times indicated on the test equipment in the appropriate blocks of the PCRI form (fig 2–35).

EXAMPLE: If your set time value is 250 ms, with a 5% tolerance, your measured results should be between 237.5 and 262.5 ms (12.5 ms deviation from the set value).

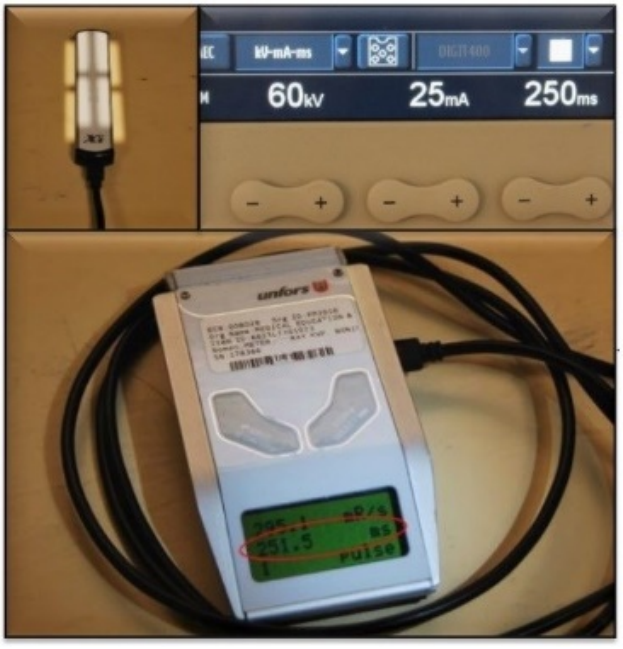

Figure 2–34. PCRI, timer accuracy setup.

1. TIMER ACCURACY:			Tolerance ±:	5%				
	(If mAs driven, leave blank)							
Setting:	2	100	250	500	1000	2000	N/A	N/A
Actual:	2.1	100	251.5	500.3	1001	2002	N/A	N/A
% Result:	5.00%	0.00%	0.60%	0.06%	0.10%	0.10%	#VALUE!	#VALUE!	Sat

Figure 2–35. PCRI, timer accuracy results.

NOTE: This test will not be performed on a mAs generator, since you cannot select time separately from the mA setting. If unit is mAs driven, leave the section blank and annotate the word "NOTE" in the final results block. Be sure to annotate this in the Actions block at the end of the form.

Radiographic leakage

The leakage test measures radiation leakage from areas of the tube other than the output port. It is based on the manufacturer's stated leakage technique factor and is used to locate shielding problems of the tubehead or collimator. This test is only performed during the initial acceptance of a unit, or if the X-ray tube has been replaced since the last PCRI.

With the collimator shutters closed, and the beam blocked with at least 1/8 inch lead shielding, check the tubehead assembly for compliance with 21 CFR 1020.30 *(k).* The maximum permissible leakage is 100 mR in one hour while the tube is operated at its maximum continuous rated current (mA) for the maximum rated tube potential (kVp). For this test, complete the following steps:

1. Select the unit's maximum kVp or highest value allowed by manufacturer.
2. Select a time station approximately equal to the response time of the survey meter you are using (*usually 1 to 3 seconds)*.
3. Select the lowest mA station. We use the lowest mA station to eliminate a deliberate miscalibration of the system to actual leakage technique factors. Ensure that the mA corresponds to the preset kVp and time such that the heating rate does not exceed the tube anode's cooling capacity.
4. Position the round flat scatter (or survey) probe of the radiation meter approximately one meter (about 3 feet) from any direction of the tube (fig 2–36). Make sufficient exposures around the entire area surrounding the tube and average the results. You may bring the probe closer to the tube to locate "Hot Spots," but be aware that at very small distances, the radiation measured, does not reflect actual leakage radiation.

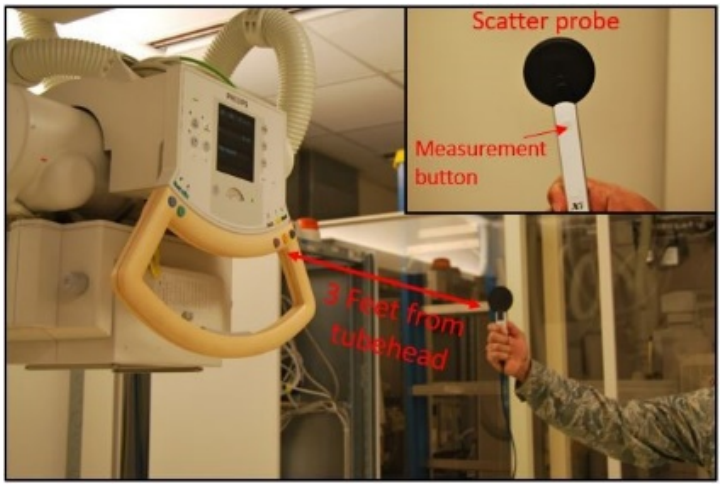

Figure 2–36. PCRI, radiographic leakage setup.

5. If the average value obtained in step 4 above is less than 100 mR/hr, the radiation leakage is satisfactory, and there is no need to proceed further with the test. If the value exceeds 100 mR/hr, you will have to calculate the actual mR/hr using the following procedure:

Divide the mA from the leakage technique factors (listed by the manufacturer) by the mA setting used for this test and multiply that by the average mR obtained in step 5.

$$\frac{\textit{(Manufacturer's Leakage mA technique factor)}}{\textit{(mA used at Leakage kVp technique factor)}} \times \textit{Averaged mR/hr value measured} = \textit{Average mR/hr leakage value}$$

EXAMPLE: If an average of 300 mR/hr is obtained at 25 mA, 150 kVp, and the specified leakage technique factor is 3.33 mA. The extrapolated value for the leakage factor would be:

$$\frac{\textit{3.33 mA}}{\textit{25 mA}} \times \textit{300 mR/hr} = \textit{39.9 mR/hr}$$

Since this value is below the maximum permissible leakage of 100 mR/hr, it is satisfactory. Record the values on the PCRI form (fig. 2–37).

2. RADIOGRAPHIC LEAKAGE: (100 mR/hr max. Perform only if tubehead removed or replaced since last test.)

150	kVp	25	mA	39.90	Avg mR/hr	Sat

Figure 2–37. PCRI, radiographic leakage results.

SID detent and scale accuracy

You must verify that the distance scale on the collimator-to-image receptor agrees with the actual focal spot location of the X-ray tube to the image receptor; this is your SID. Note that if the SID is incorrect, the unit will believe that the tubehead is either closer or further from the patient than it actually is and possibly over or under expose that patient to unnecessary radiation.

The detent is a mechanism that tends to stop a moving part in a specific location. In this instance, the detent stops the tubehead from gliding along its track at specific distances. Generally, you will see detent at the 40-inch mark when the unit is in the horizontal table position and at the 72-inch mark in the vertical wall stand position. Therefore, when adjusting the tube head height over the horizontal table, the movement will stop and “catch” when you reach 40 inches. This reduces the estimation of the tube positioning for standard distance measurements.

The focal spot of the X-ray tube should have a marking on the tube housing, either a sticker, tape, or permanent marker. This indicates the exterior measurement reference point for your SID. If the focal spot is not marked, be sure to refer to your manufacture’s literature to determine the proper way to measure and mark the housing. You will verify the accuracy of all Source-Image-Distance indicators, both tapes and detents, for each tube receptor combination available. Typical combinations are Tube-Table Horizontal (fig. 2–38), Tube-Table Vertical, and Tube-Chest Receptor (fig. 2–39).

Use the following steps to perform this test:

1. Using the focal spot marked on the X-ray tube (fig. 2–38A), measure and record indicated versus actual SID for each tubehead-receptor combination. Check at a commonly used SID such as 40 inches, 48 inches, or 72 inches.
2. Be sure to measure the SID to the open image receptor tray or bucky (fig. 2–38B), not the table.

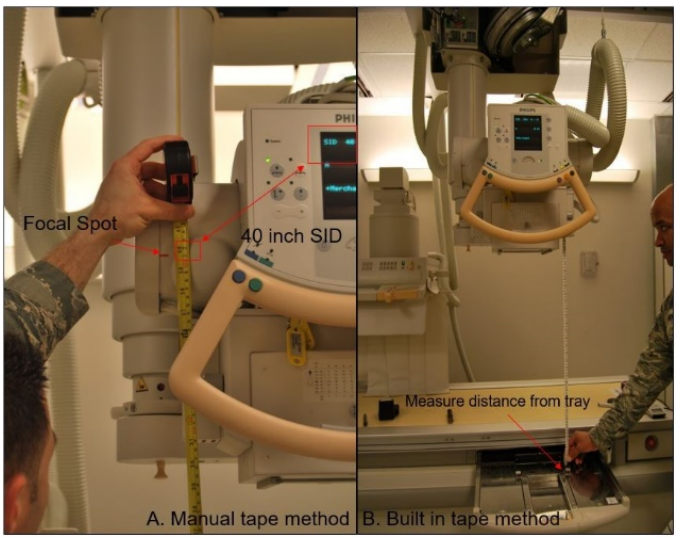

Figure 2–38. PCRI, SID table at 40 inches.

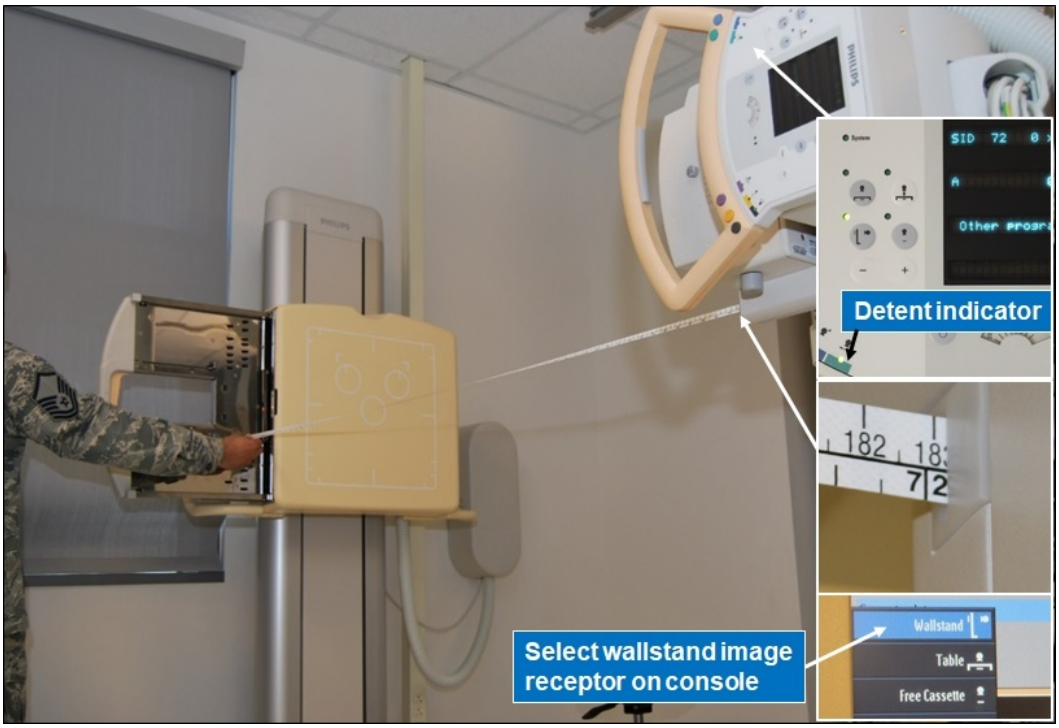

Figure 2–39. PCRI, SID chest stand at 72 inches.

3. Record the values on the PCRI form (fig. 2–40). If the measured values are out of tolerance, you will have to adjust the unit's perceived SID or detent position.

3. SID DETENT AND SCALE ACCURACY:		Tolerance ±: 5% of Indicated		
	Indicated	Actual	%	
Tube to table (horizontal):	40	39.90	-0.25%	Sat
Tube to table (vertical):	40	39.90	-0.25%	Sat
Tube to Chest Receptor:	72	72.00	0.00%	Sat

Figure 2–40. PCRI, SID detent results.

Illumination

This test verifies that the light field (collimator light) provides at least 15-footcandles (FC) of illuminance at a distance of one meter or at the maximum SID, whichever is less, above the ambient light in the X-ray suite. The collimated light is broken down into 4 quadrants (fig. 2–41).

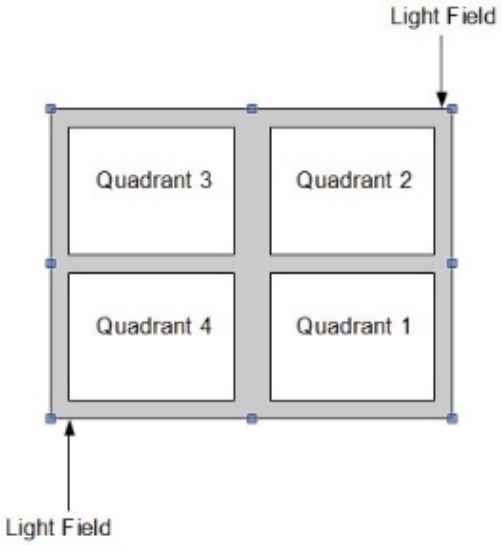

Figure 2–41. PCRI, light field quadrants (Courtesy of Fluke/Raysafe).

To complete this test:

1. Using a light meter, measure the ambient room light as a point of reference.
2. Next, measure the illumination in the center of each quadrant at a distance of 100 cm (39.5") or the maximum SID, whichever is less, as seen in figure 2–42.

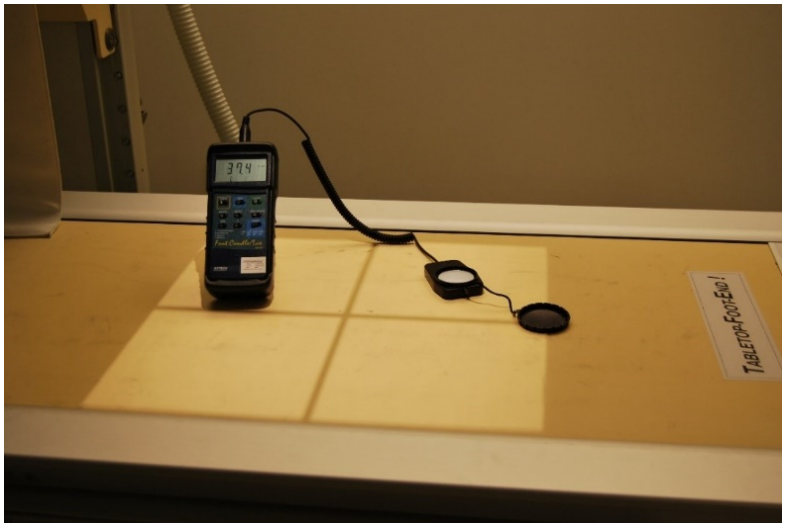

Figure 2–42. PCRI, illuminance measurement.

3. Average the four results and subtract the ambient room light. The minimum requirement is 15 FC brighter than the ambient room light. You may need to replace the collimator lamp if results do not meet the standard. Usually the collimator template just has dust build-up on the inside of the collimator package. Cleaning it may fix this issue.

4. Indicate the ambient and average values on the PCRI form to determine total illuminance (fig. 2–43).

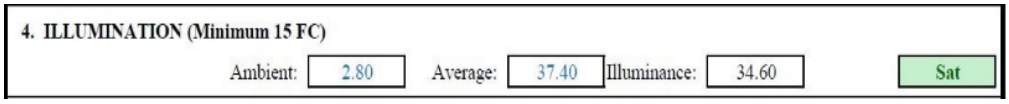

Figure 2–43. PCRI, illuminance results.

Light field offset

The light field offset test is also known as radiation field versus light field. This test verifies that the collimated light field accurately indicates the location that radiation will strike on the film during an exposure. There are two ways to perform this test: direct measurement using an X-ray ruler and manually using markers and exposures.

Direct method

The direct measurement method is relatively simple. Set the SID to 40" and collimate the light field to a common size. The direct X-ray ruler will have a zero point in the center. Position that zero line directly across the border of the light field, as shown in figure 2–44.

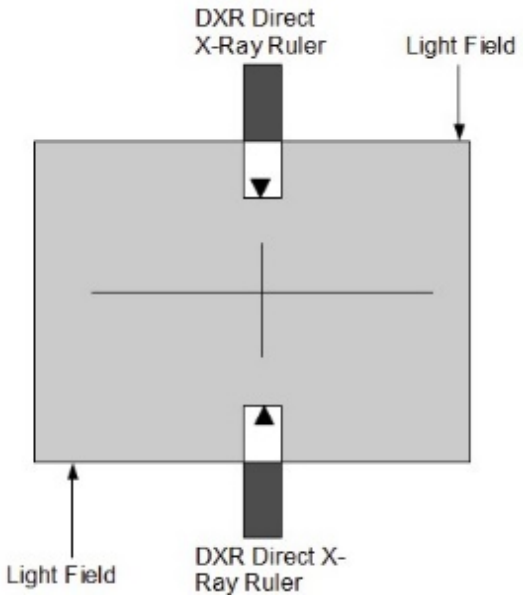

Figure 2–44. PCRI, light field ruler alignment (Courtesy of Fluke/Raysafe).

Create an exposure and measure the deviation for each side of the light field (fig. 2–45). The ruler display should indicate the presence of radiation along gradual marks. You will take four total exposures: two for the total length deviation (L1 + L2) and two for the total width deviation (W1 + W2). This will give the left and right versus the top and bottom offset of the actual exposure area to the light field.

Figure 2–45. PCRI, light field direct exposure (Courtesy of Fluke/Raysafe).

Manual method

If you do not have an X-ray ruler, the manual method allows you to measure the offset by placing markers on the edges of the light field and measuring the offset on the exposure.

To perform the manual light field offset test:

1. Manually adjust the collimator shutters so the light field, when measured on an unexposed cassette at a SID of 40 inches or 48 inches, is set to a commonly used field size. (i.e., 8" x 10" or 10" x 12").

   NOTE: *If the cassette is placed on the tabletop the indicated SID will not be correct and the tube will have to be raised.*

2. Mark the edges of the light field so they will be visible on the exposed film (fig. 2–46). Most BMETs use hex wrench or quarters, but hex wrenches provide a long flat edge for better alignment with the edge of the light. You can place the flat edge of the wrench either inside or outside of the light field; the most important detail is making sure that they are aligned as level with the light beam as possible.

Figure 2–46. PCRI, light field Allen wrench setup.

3. Expose and develop the film and pull it up on the QC workstation for measurement.
4. As you can see in figure 2–47, the Allen wrenches are clearly visible on the exposure. Using the measurement software, measure the offset between the indicated and actual edges of the radiation field on the exposed film. Sum the two edges of the film and enter the deviation on the form.

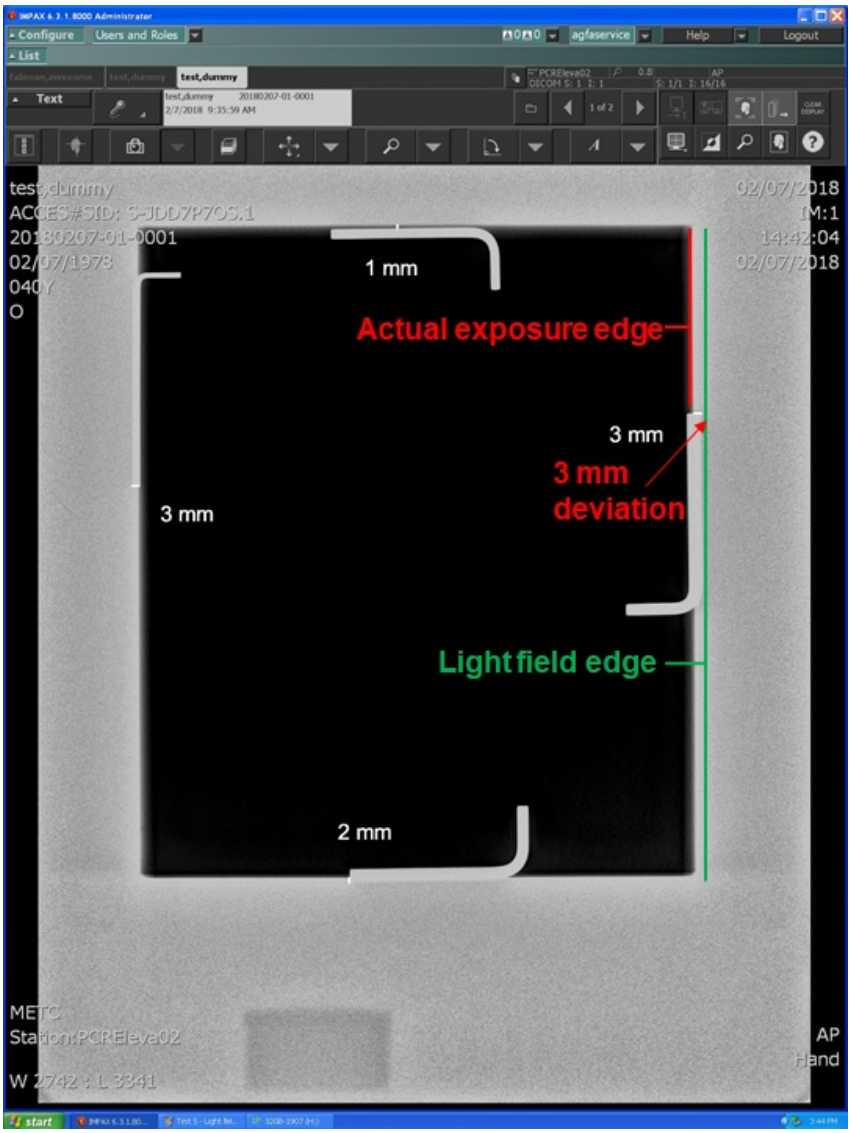

Figure 2–47. PCRI, light field Allen wrench measurement.

5. Calculate the allowable errors in either dimension by multiplying the SID by 0.02 (2%).
6. If either the width or length exceeds allowable values, indicate unsatisfactory on the inspection form (fig. 2–48).

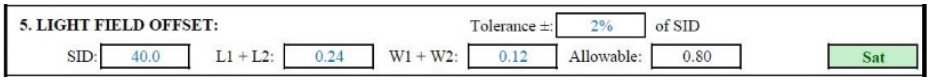

Figure 2–48. PCRI, light field offset results.

NOTE: If possible, try to correct errors unless unit is under warranty or service contract.

Field size versus indicator

Field size versus indicator verifies that when you adjust the collimator to a specific size, you are radiographing the correct sized field. In this test you measure and compare the length and width of the X-ray field after an exposure and compare it to the size selected. For instance, if you collimate down to a 10" x 10", is the X-ray image truly 10" x 10"?

Use the following steps to perform this test:

1. Place a cassette in the image receptor tray or on top of the table and select a commonly used SID (fig. 2–49).

   NOTE: You may need to readjust the SID if the cassette is placed on the table top.

2. In manual mode, adjust the collimator indicators to a commonly used field size (i.e., 8" x 10" or 10" x 12"). Use a larger cassette and collimate the field slightly smaller. This is so you can fully measure the outer dimensions of the selected field.

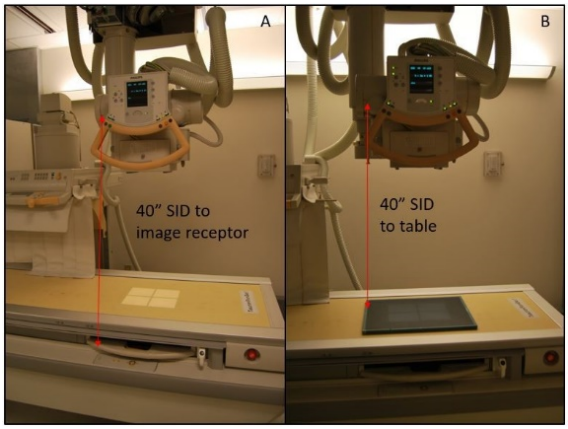

Figure 2–49. PCRI, field size versus indicator setup.

3. Make an exposure and develop the film.

4. Measure the actual length and width of the radiation field on the exposed film. Subtract those values from the indicated dimensions and enter the length and width differences.

   EXAMPLE: Figure 2–50 shows an exposure of a 10" x 12" collimated field. The long edge measurement of 299 mm, will convert to 11.77 inches (conversion factor is mm/25.4). The short side measurement of 250 mm is equal to 9.84 inches. Therefore, your actual exposure field is 9.84" x 11.77".

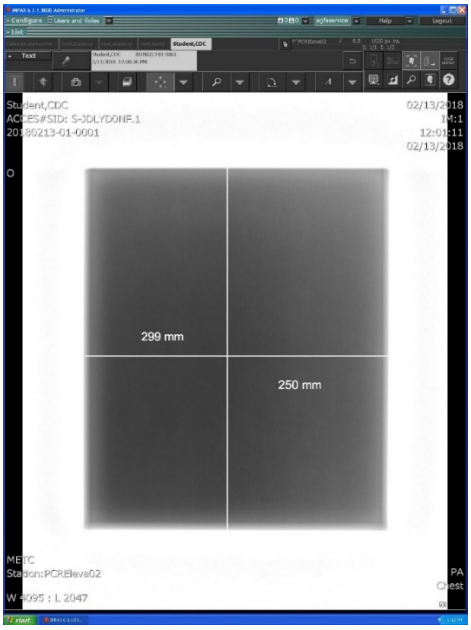

Figure 2–50. PCRI, field size versus indicator measurements.

5. Calculate the percent of error by multiplying 2% tolerance times the SID used. Therefore, 2% of a 40" SID leaves you with an error tolerance of 0.8". Indicate whether satisfactory or unsatisfactory on the PCRI form (fig. 2–51).

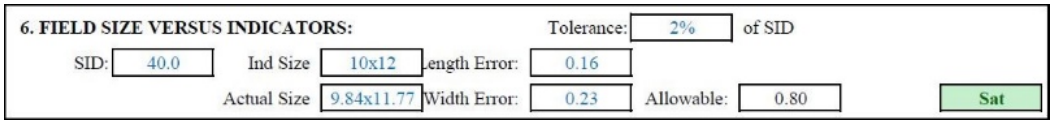

Figure 2–51. PCRI, field size versus indicator results.

Field versus receptor offset

The purpose of this test is to verify that when you center the X-ray tube on the image plane, the center of the film is also the center of the radiographed object. You are measuring the distance between the radiated area versus overall image to determine the offset of the center points (fig 2–52). Be sure to perform this test on all image receptor-tubehead combinations. Table-horizontal and table-vertical are considered two different image receptors for this test.

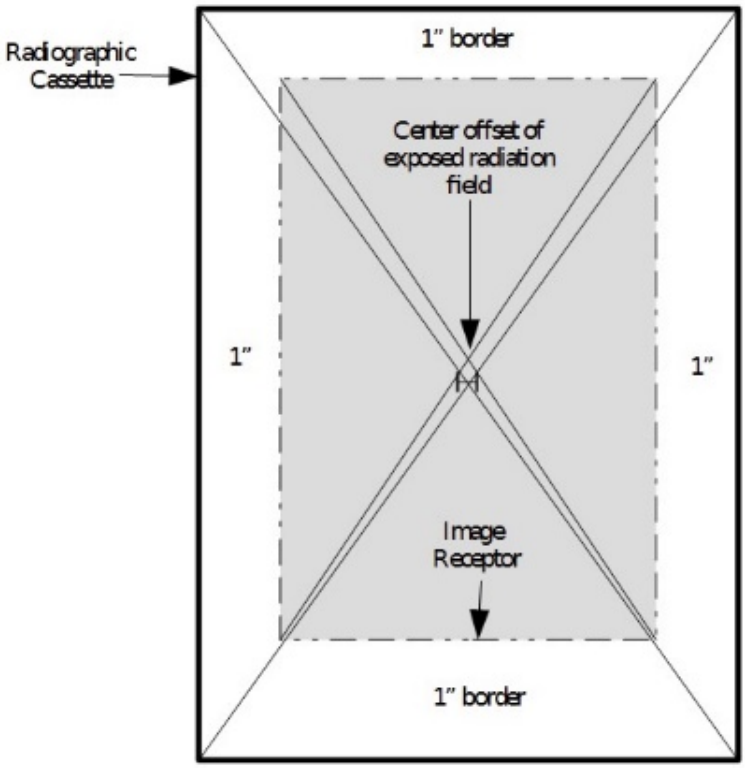

Figure 2–52. PCRI, center offset description.

To perform the center offset verification:

1. Place a cassette in the cassette tray.
2. Using applicable detents, angulation indicators and centering lights, center the tubehead and collimator to the receptor at a commonly used SID.
3. Manually adjust the collimator to a field smaller than the cassette loaded in the receptor (fig. 2–53).

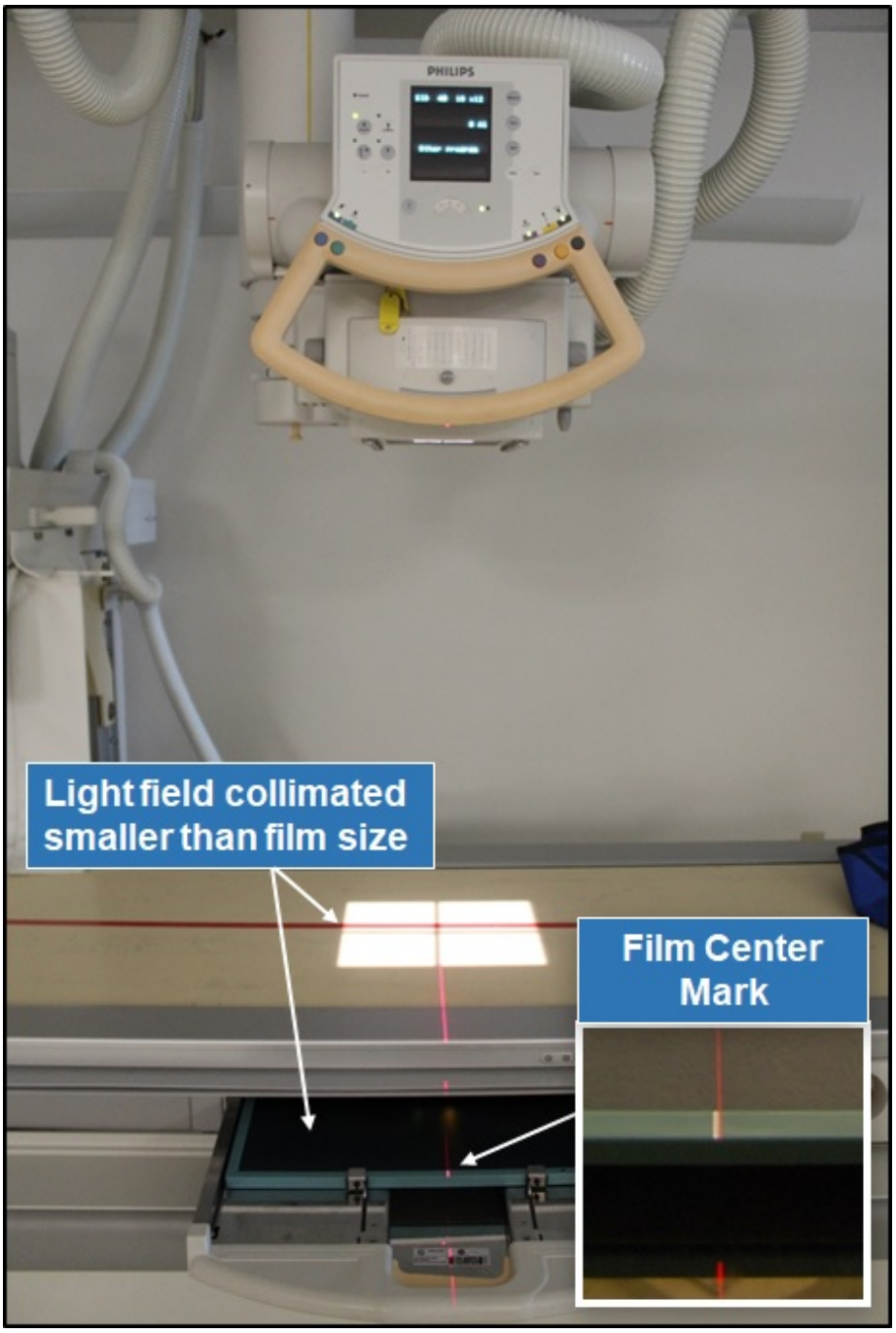

Figure 2–53. PCRI, center offset setup.

4. Expose and process the film then bring the image up on the PACS workstation.
5. Draw two sets of crossed lines on the film, one set connecting the corners of the film and one set connecting the corners of the exposed X-ray field.
6. Measure the offset between the film center and the exposed X-ray field center as shown in figure 2–54.

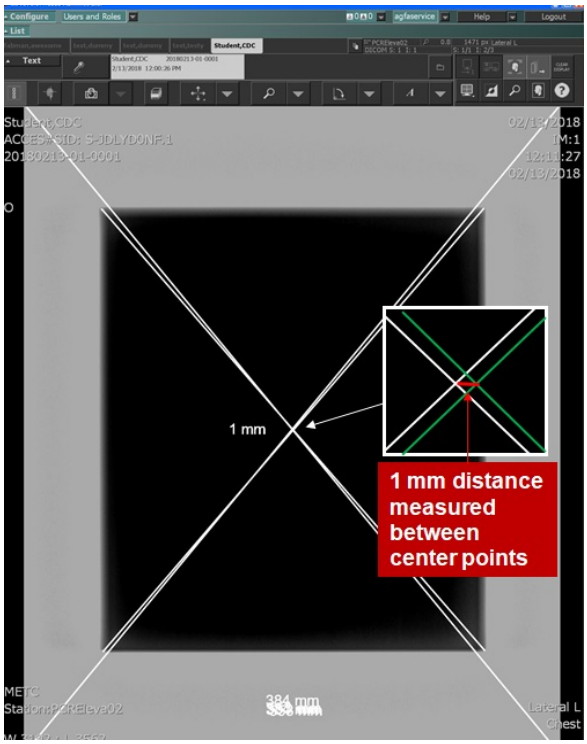

Figure 2–54. PCRI, center offset measurement.

7. Calculate the allowable offset by multiplying 2% times the SID used, and record results on the PCRI form (fig. 2–55).

   Example: A 2% tolerance times a 40 inch SID equals an allowable deviation of 0.8 inches. Using our example from figure 2–54, a 1mm distance between center points equals 0.04 inch deviation (1mm/25.4 conversion factor = 0.0393 inches).

NOTE: Complete the above test for each tubehead-receptor combination.

NOTE: Use metric size film as standard size cassettes may introduce errors in the center offset test.

7. FIELD VERSUS RECEPTOR OFFSET: Tolerance ±: 2% of SID

	SID	Center Offset	Allowable	
Table Horizontal	40.0	0.04	0.80	Sat
Table Vertical	40	0.04	0.80	Sat
Chest Receptor	72	0.2	1.44	Sat

Figure 2–55. PCRI, center offset results.

Field size (auto)

Most collimators can automatically adjust the collimator to the size of the cassette inserted into the bucky. This is known as PBL. This test verifies the collimated radiation field dimensions agree with the physical dimensions of the image receptor at a particular SID. Evaluation of the test image will identify any misalignments in the radiation field caused by the collimator shutters being out of alignment. Failure of the PBL test could excessively expose the patient to radiation or lose important diagnostic information due to image cut-off.

NOTE: This test is not applicable to purely digital systems as they will not have an autocollimation function for a digital receptor.

The easiest way to perform this test is by using the collimator rotation method. To perform this test, use the following steps:

1. Place a fourteen by seventeen (14" x 17") inch cassette in the receptor. Align, center, and lock the X-ray tube to the center of the image receptor at a commonly used SID (i.e., 40 inches, 44 inches, or 72 inches).
2. Rotate the collimator approximately 45 degrees (fig. 2–56). The collimator should lock in place at 45 degrees.

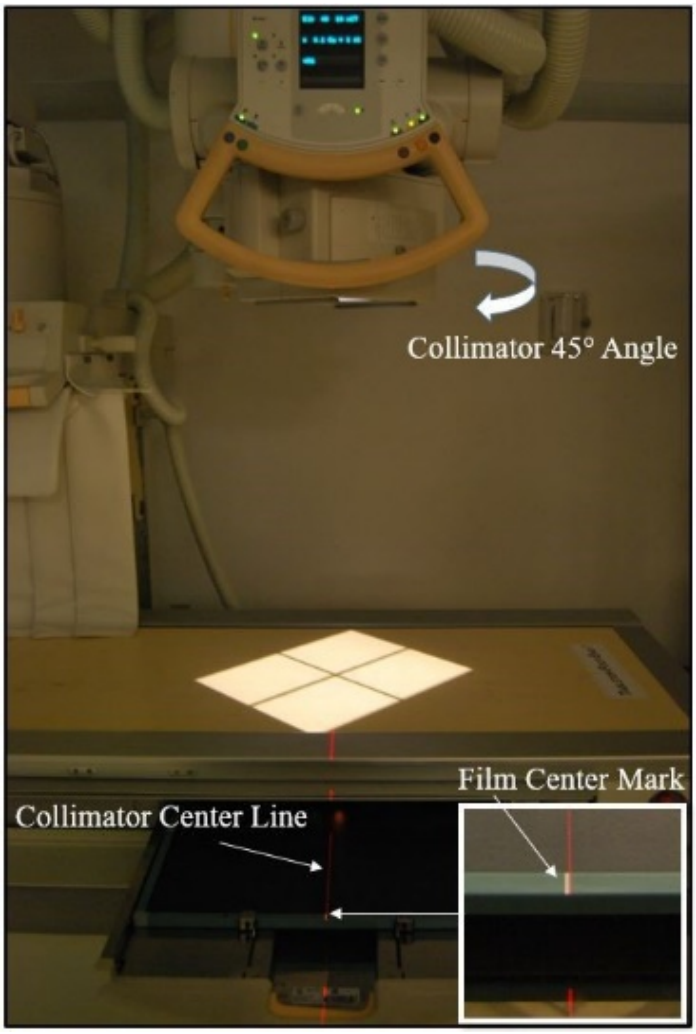

Figure 2–56. PCRI, field size auto setup.

3. Take an exposure at the following technique: 70 kVp, 2 mAs. Phototiming is suggested for most systems.
4. After processing the film, bring the image up on the PACS workstation.
5. Measure the length and width of the exposed area. At this point you can see why we rotate the image by 45 degrees in figure 2–57. It allows you to measure the exact dimensions where the auto-collimated exposure cuts off. In previous tests, we collimated the field to slightly smaller than the film size to see the exposure edges, but since the field size should be automatically collimated to the same size as the film size, you would not see the borders without rotation. Annotate the difference of the length and width.

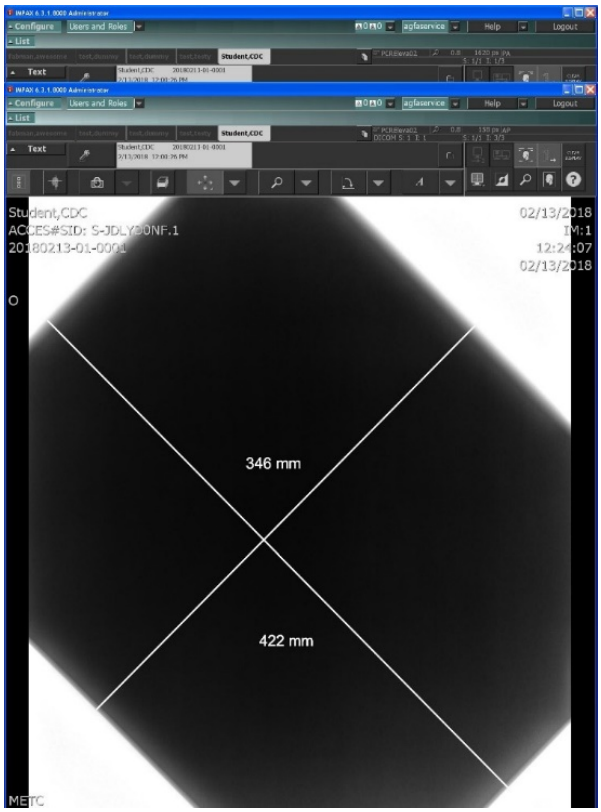

Figure 2–57. PCRI, field size auto measurement.

6. Calculate the allowable offset and measured values and record the results on the PCRI form (fig. 2–58). For any one dimension (length or width) the radiation field cannot exceed three percent (3%) of the SID. When the errors of both length and width are added together, the total error cannot exceed 4% of the SID

   EXAMPLE: A 4% tolerance times a 40 inch SID equals an allowable deviation of 1.2 inches. Using our example from figure 2–57, the long edge measurement of 422 mm, will convert to 16.61 inches using the conversion factor; mm/25.4. The short side measurement of 346 mm is equal to 13.62 inches. Therefore, your actual collimated exposure size is 13.62" X 16.61" (0.38 and 0.39 deviation respectively).

8. FIELD SIZE (AUTO):				Tolerance ±:	3%	of SID per dimension
A: Table Horizontal:	SID:	40.00		Tolerance ±:	4%	of SID for sum of errors
Length Error:	0.38	Width Error:	0.39	Allowable per dimension:	1.20	
Sum of Errors:	0.77	llowable for sum of errors:	1.60			Sat
B: Chest Receptor:	SID:	72.00				
Length Error:	0.72	Width Error:	0.73	Allowable per dimension:	2.16	
Sum of Errors:	1.45	llowable for sum of errors:	2.88			Sat

Figure 2–58. PCRI, field size auto results.

NOTE: This test is performed on each tubehead-receptor combination that has auto collimation. The table image receptor (cassette holder) is considered as one receptor regardless of vertical or horizontal position.

If the radiation field exposure is greater or smaller in physical dimension than the size of the test cassette, adjustment is required.

Reproducibility

This test confirms the same amount of radiation (mR) is produced when the same exposure parameters (mA, time, and kVp) are selected. This verifies how consistently a unit will provide the same readings when repeatedly taking the same shot. Four consecutive exposures, with identical parameters, are taken one minute apart.

To perform the reproducibility test, use the following steps:

1. Set the tubehead at a 27 inch SID and collimate the beam to the X-ray tester (fig. 2–59).

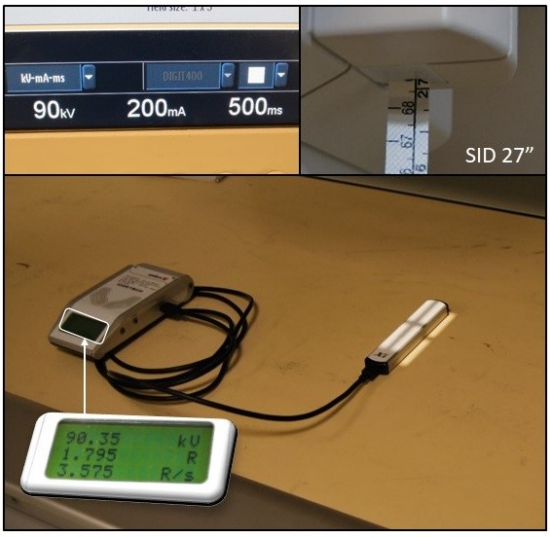

Figure 2–59. PCRI, reproducibility setup.

2. Take four consecutive exposures at the following technique: 90 kVp, 200 mA, 0.5 sec. Wait 1 minute between shots to allow time for the unit to cool.
3. Record the mR values obtained on the PCRI form (fig. 2–60). If all values are within 5% of the average, reproducibility is satisfactory.

9. REPRODUCIBILITY:					Tolerance ±:	5%	of Average	
SID	27.00	kVp	90	mA	200	Time	0.50	
mR1	1795.00	mR2	1796.00	mR3	1794.00	mR4	1795.00	
				Min	Max			
Standard Deviation	0.71			1705.25	1884.75	Average	1795.00	Sat

Figure 2–60. PCRI, reproducibility results.

Tube current output and linearity

This test requires the most amount of radiographic shots of any of the PCRI tests. The tube current output and linearity test validates the calibration performed on the mA stations and ensures each station produces an acceptable amount of radiation. Exposures are taken at variety of mA stations, for a set time and a set kVp using both the large (L) and small (S) focal spots. This verifies accuracy of the kVp and also validates that the selected mA is accurate and remains constant throughout the kVp range. Radiation readings are then calculated by dividing the mR by the time selected, which provides a given amount of radiation per unit of time (mR/time).

You will perform the test at three different kVp settings, such as one at 60 kVp (low), one at a neutral kVp setting (the kVp at which basic mA calibration is performed, usually 80–90 kVp), and one at 120 kVp (high).

To perform this test:

1. Continue to use the same equipment set up from the reproducibility test (fig. 2–61).

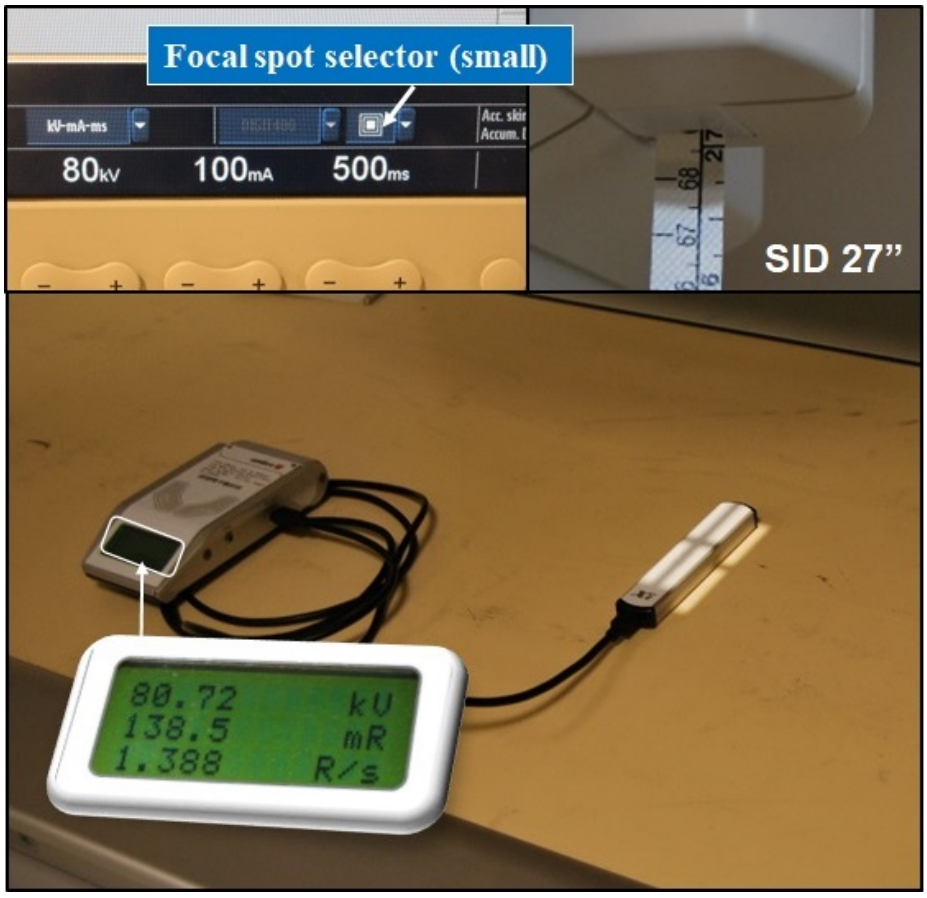

Figure 2–61. PCRI, tube current and linearity setup.

2. For each kVp tested (low, neutral, and high), make an exposure at representative mA stations, including at a minimum the following; 25, 50, 100*(S)*, 100*(L)*, 200*(S)*, 200*(L)*, 300*(S)*, 300*(L)*, or 320, 400, 500, 600 or 640, 700 or 720, 800, 900, and 1000mA. Exposure durations used should be in the range of 0.1 *(1/10)* second for mA stations above 100mA to .5 *(1/2)* second for mA stations below 100mA. Time selections can be varied between mA stations since this will be compensated for in the conversion to mAs.

   NOTE: If the X-ray system has a mAs selector rather than independent mA and time selections, record the settings in the mAs blocks on the worksheet.

   TIP: Start at the highest kVp setting and allow adequate cool down between exposures. You wouldn’t want to start taking shots at your highest kVp ranges after your tubehead is already near its max heat limits.

3. Record the mA, time selected, and mR obtained in the appropriate blocks of the inspection form (fig. 2–62). The computerized PCRI form will calculate the radiation per unit of time.

4. Verify the values recorded are within the upper and lower limits (+/- 10% of this average). Determine whether any value is outside the range either above or below the limits, and readjust the mA stations to obtain satisfactory results (mA adjustment cannot exceed manufacturer’s calibration tolerances for the particular mA station under test).

NOTE: This linearity limit is more stringent than addressed in 21 CFR 1020.30 (c).

10. TUBE CURRENT OUTPUT AND LINEARITY:

Small Focal

SCD			Low kVp		Neutral kVp		High kVp	
27			60.00		80.00		120	
mA Sta	Time	mAs	mR	mR/mAs	mR	mR/mAs	mR	mR/mAs
25.0	0.50	13	53.08	4.25	165.20	13.22	218.30	17.46
50.0	0.50	25	106.20	4.25	335.70	13.43	435.90	17.44
100.0	0.10	10	42.53	4.25	138.50	13.85	173.80	17.38
200.0	0.10	20	85.38	4.27	263.60	13.18	345.50	17.28
320.0	0.10	32	106.50	3.33	420.69	13.15	431.60	13.49

Large Focal

SCD			Low kVp		Neutral kVp		High kVp	
27			60.00		80.00		120	
mA Sta	Time	mAs	mR	mR/mAs	mR	mR/mAs	mR	mR/mAs
320.0	0.10	32	106.10	3.32	198.20	6.19	428.70	13.40
400.0	0.10	40	136.40	3.41	253.40	6.34	550.80	13.77
500.0	0.10	50	171.50	3.43	318.90	6.38	692.80	13.86
640.0	0.10	64	214.00	3.34	398.50	6.23	867.00	13.55
800.0	0.10	80	271.30	3.39	504.80	6.31		0.00
			10		10		9	
			AVERAGE	3.72		9.83		15.29
Tolerance ±:	10%	of average	3.35	4.10	8.84	10.81	13.76	16.82 Sat

Figure 2–62. PCRI, tube current and linearity results.

Beam quality

X-rays produced are not all of the same intensity. Low-level X-rays will not add any detail to an image, but will increase radiation exposure to the patient. To reduce the amount of undesirable radiation that reaches the patient, we use a filter in the beam. This test verifies there is sufficient filtration in the beam path. The HVL of an X-ray beam is the thickness of absorbing material necessary to reduce the X-ray intensity to half of its original value. Federal law requires an HVL of 2.5 millimeters of aluminum (mm Al) or aluminum equivalent for 1ϕ X-ray systems, and an HVL of 3.2 mm Al for 3ϕ systems (both at 90 kVp).

Follow these steps to perform the beam quality test:

1. Select a 90 kVp, 200 mA, and 0.5 sec exposure (100 mAs). Keep kVp, mA, time, and distance constant for each exposure.
2. Remove all selectable filters and initiate an exposure. You should be able to select added filtration options from either the exposure settings on the control unit or the collimator controls (fig. 2–63). Record the mR value attained in the “0mm” block of the PCRI form.

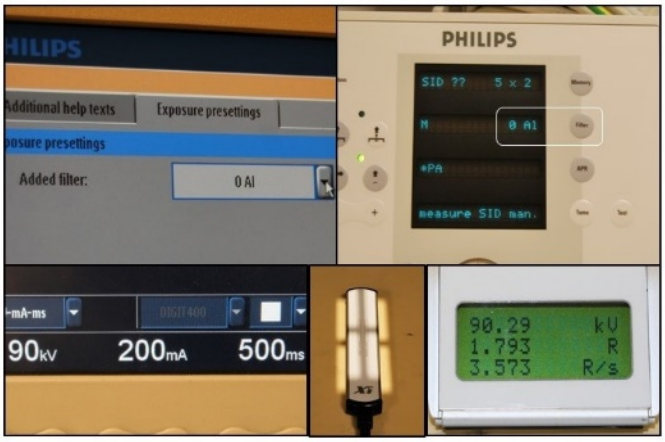

Figure 2–63. PCRI, beam quality 0 mm Al.

3. Insert 2.5mm of aluminum between the tubehead and detector. You can either select the added filtration from the unit, add 2.5 mm Al by placing aluminum plates between the tubehead and the detector, or use a combination of both (fig. 2–64). Just ensure that the total filtration in the beam path, between the source and the detector, is equal to 2.5 mm Al. Initiate an exposure and record the mR value attained in the "2.5mm" block on the PCRI form.

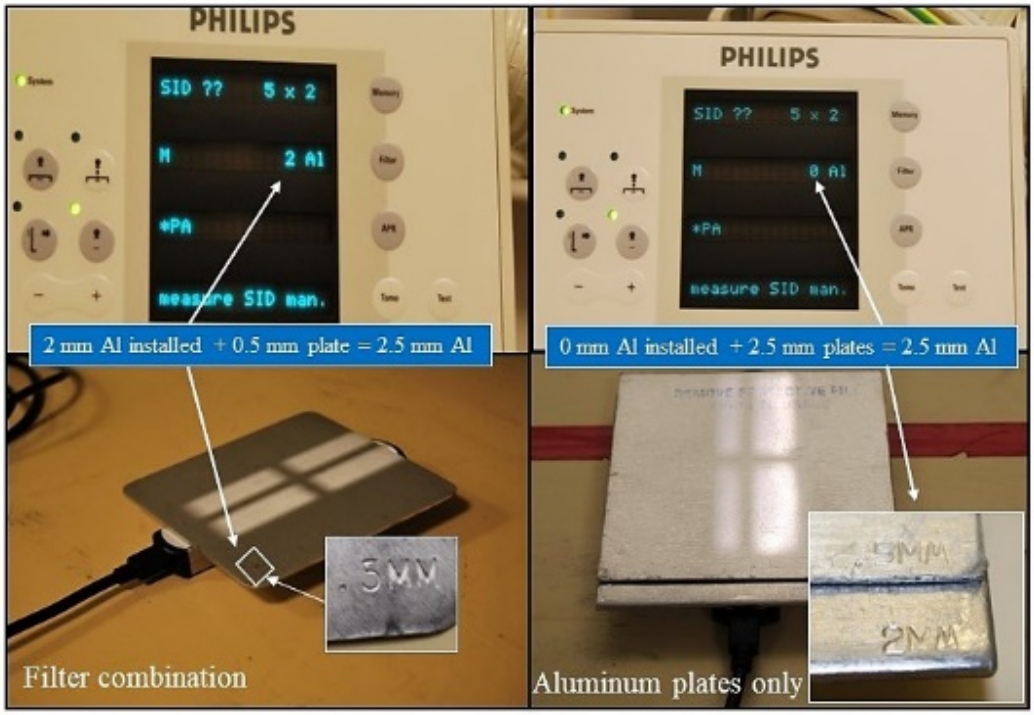

Figure 2–64. PCRI, beam quality 2.5 mm Al.

4. Repeat step 3 using a total filtration of 4.5 mm Al. Initiate an exposure and record the mR value attained in the "4.5mm" block on the PCRI form (fig. 2–65).

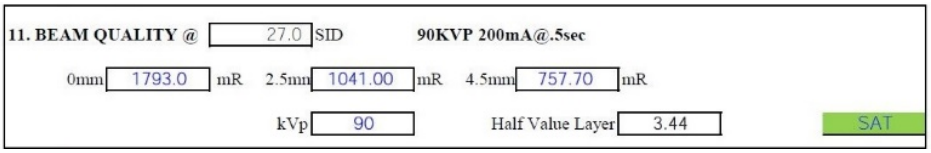

Figure 2–65. PCRI, beam quality results.

HVL must be at least 3.2 mm Al at 90 kVp. As you can see, the mR values significantly reduce with each added layer of filtration. Some versions of the PCRI form (as different bases use various versions) will auto calculate the HVL with the given mR values. If you are using the Unfors non-invasive X-ray tester, however, you may not need to add any aluminum at all. This unit can calculate the HVL in one shot (fig. 2–66).

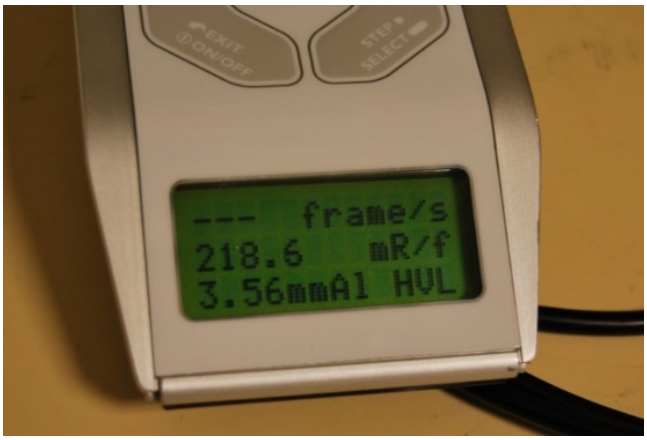

Figure 2–66. PCRI, beam quality using Unfors.

Notes section

This is the area where any action items or unsatisfactory results should be explained, and any corrective actions documented. For instance, if you N/A any test for a valid reason, explain the justification in this block (fig. 2–67).

NOTES
ACTION REQUIRED (YES/NO): No EXAMPLES: No Timer Accuracy settings: 50 KV, 200 mA, 2 ms min, 2000 ms max Reproducibility/Linearity: SID setting 36.5 Test 2: Tubehead has not been removed or replaced in the last year (Radiographic Leakage Test not needed) Test 7/8: N/A due to system being a Digital Unit
TECHNICIAN(S): MSgt Don Shula, TSgt Dan Marino, SrA Larry Csonka

Figure 2–67. PCRI, action required block.

Finally, the name of the technicians performing the inspection are placed in the last block. This completes a radiographic PCRI. For information on performing the tests, contact your MERC or refer to the BMET advanced courses.

Self-Test Questions

After you complete these questions, you may check your answers at the end of the unit.

616. Maintenance procedures for fixed systems

1. Where is the best place to start the preventive maintenance inspection on a fixed X-ray unit and why?

2. List five items to check on the ceiling rails and upper tube support sections.

3. When inspecting the X-ray table, what should you look for in the bucky and bucky tray?

4. What would be the result of not putting enough dielectric compound on the high-voltage cable connections to the X-ray tube?

5. List three actions to complete when inspecting the X-ray control unit.

617. Maintenance procedures for mobile systems

1. What should be established in your MTF to ensure batteries in mobile X-ray units are correctly maintained?

2. What should be done if you find you must perform battery equalization cycles more than once per year?

3. What do "clunking" noises or "jerking" motions indicate when checking the tube-stand bearings?

4. List two areas of primary concern during a preventive maintenance inspection of a portable X-ray unit.

618. Non-invasive X-ray test equipment

1. Briefly describe the difference between invasive and non-invasive X-ray test equipment.

2. Which measurements are considered real-time values?

3. Match each definition in column A with the appropriate term in column B. Items in column B may be used once or not at all.

Column A	*Column B*
____ (1) Dose.	a. Used to estimate the total filtration of the X-ray beam.
____ (2) Time.	b. An oscilloscope waveform that illustrates how the applied kVp varies over time during an exposure.
____ (3) HVL.	c. Measured from the starting trigger of an exposure until the signal falls below 25 % of the peak.
____ (4) kVp.	d. The energy absorbed per unit mass, usually expressed in terms of Gys.
____ (5) kVp waveform.	e. The product of the electrical current used to produce the X-rays and exposure time.
____ (6) mAs.	f. The peak voltage applied across the electrodes of an X-ray tube during an exposure.

4. What are the two methods used by non-invasive X-ray test equipment to measure radiation (dose)?

5. What is the primary difference in the measuring technique used by non-invasive X-ray test equipment for different X-ray modalities?

6. What is the primary concern with non-invasive X-ray test equipment?

619. Post Calibration Radiation Inspection

1. The form for PCRIs is designed for use with what?

2. Why do you list the test equipment calibration dates on the PCRI form?

3. What does the timer accuracy test verify?

4. Why would you not perform the timer accuracy test on a mAs generator unit?

5. When is the radiographic leakage test performed?

6. What is the maximum permissible radiographic leakage value?

7. Calculate the radiographic leakage value with an average measurement of 250 mR/hr at 25 mA, 150 kVp, and a specified leakage technique factor of 3.72 mA.

8. Explain a potential hazard of an inaccurate SID.

9. What is the purpose of the detent mechanism?

10. The light field must be calibrated to at least how many FCs?

11. How do you determine the illumination level of the collimator lamp?

12. List the two ways to perform the light field offset test.

13. What common objects can you use to mark the light field during the manual light field offset test?

14. What does the field size versus indicator test verify?

15. How do you determine the field versus receptor offset?

16. The failure of which test could excessively expose a patient to radiation or lose important diagnostic information?

17. How far should you rotate the collimator during the field size PLB test? Why?

18. What is the purpose of the reproducibility test?

19. Which test requires the most amount of radiographic shots?

20. What is the HVL of an X-ray beam?

Answers to Self-Test Questions

608

1. Over the tabletop.
2. A cone.
3. A collimator or variable aperture collimator.
4. Usually 3-mm thick lead.
5. The X-ray beam.
6. Positive beam-limiting device. When film is loaded in the bucky tray and clamped into place, sensing devices in the tray identify the size and alignment of the cassette. An electronic signal is transmitted to the collimator housing and actuates synchronous motors that drive the collimator leaves to a precalibrated position, so the beam is restricted to the size of the film being used.

7. 40,000–150,000V.
8. Line compensation, kVp, mA, and exposure time.
9. (1) A means to position the patient automatically by the use of a tabletop, motor-drive assembly.
   (2) Can be rotated from 0° horizontal to a 90° vertical position for special-purpose techniques.
   (3) Houses several special components, such as the cassette holder and bucky device.
   (4) May contain an AEC circuit that automatically terminates the exposure to the film, depending on patient density, for consistent radiographs throughout a wide range of patient densities.
   (5) May contain an extra X-ray tube for special procedures to include a spot film tower and an image intensifier.
10. (1) Under the tabletop.
    (2) On a wall stand.
11. (1) Primary barriers.
    (2) Secondary barriers.

609

1. To visualize organs in motion, insertion of catheters into blood vessels of the heart, introduction of opaque substances into organs, and different parts of the body in different planes and projections.
2. Inside the tilt table.
3. To protect it from rough handling and possible breakage.
4. Converts X-rays into visible light.
5. The photocathode.
6. If the photocathode were farther away from the input phosphor, the light would lose its intensity by the time it reached the photocathode, thus reducing the number of photoelectrons and the final brightness of the output phosphor.
7. Accelerated by the high positive voltage on the anode; focused by electrostatic focusing plates or lenses located along the sides of the image intensifier.
8. A measurement of the increase in light photons due to the conversion efficiency of the output screen.
9. Ability of the image intensifier tube to increase the illumination level of the image; a product of the minification gain and flux gain.
10. The 25cm–17cm (25/17) design.
11. Reduce the field of view, thereby magnifying the image.
12. The X-ray tube mA must be increased (thus increasing patient dose).
13. A CCD camera system.
14. Digital images are obtained when the silicon is illuminated, causing generation of an electrical charge. This electrical charge is then sent as a signal to the analog-to-digital converter. The ADC digitizes the signal from the CCD by converting the signal into pixels or picture elements. The ADC assigns a numerical value to each pixel corresponding to the electronic signals. The pixel numerical values are then organized (mapped) into a matrix. These images appear on monitors, which can be placed in the room or in specific viewing areas.
15. The television system.
16. An FPD assembly.
17. 5 million.
18. By using software to interpolate values for those defective elements.

610

1. Original voltage for the mobile unit is supplied from batteries rather than facility voltage as for a fixed unit.
2. (1) Full-power high-frequency units.
   (2) Capacitor discharge units.
3. So that upon release it immediately disengages the power drive functions to prevent harm.

4. (1) Orthopedics.
   (2) Surgery.
5. It does not have drive or battery systems.

611

1. The detection of breast cancer.
2. (1) Diagnostic (performed on at risk patients or those with symptoms).
   (2) Screening (performed on patients with no symptoms).
3. 4 (2 per breast).
4. Because only muscle and fat are imaged during the exam.
5. It is used to hold the breast in position and prevents motion blur; separates underlying tissue and brings all the breast tissue closer to the image receptor; gives a more uniform thickness, optical density, and improved contrast resolution; and lowers the radiation dose required.
6. High-frequency; it allows for a lower patient dose of radiation.
7. 25–45 lbs.
8. Reduced magnification, reduced tissue thickness, reduced radiation exposure, reduced motion unsharpness, improved visualization of breast structures, and more uniform image receptor exposure.
9. Grids.
10. (1) Screen-film combination.
    (2) Digital techniques using a CCD.
    (3) Digital techniques using a FPD
11. The MQSA.
12. 3 years.
13. Installing a new unit or processor, disassembling and reassembling a unit or processor, or changing or repairing major components of a unit or processor; a medical physicist or an individual under the direct supervision of a medical physicist.

612

1. Any three of the following:
   (1) To detect, confirm, or classify dental diseases.
   (2) To detect or evaluate injuries.
   (3) To evaluate growth and development.
   (4) To detect extra or missing teeth.
   (5) To document a patient's dental condition.
2. (1) Intraoral.
   (2) Extraoral including panoramic, cephalometric, and cone beam computed tomography.
3. (1) Periapical—shows teeth roots and surrounding bone.
   (2) Bitewing—shows on a single film the upper surface of the teeth and portion of the jaw containing the tooth sockets of a given area.
   (3) Occlusal—shows a cross-section of the upper and lower dental arches.
4. Around 15 periapicals and 4 bitewings.
5. Any three of the following:
   (1) Lower radiation dose.
   (2) Overcomes certain limitations of the intraoral techniques.
   (3) Quicker.
   (4) Better coverage of the dental arches.
   (5) Earlier detection dental problems.
6. 10 mA in cephalometric compared to 4–8 mA in panoramic.

613

1. (1) Barium.
   (2) Iodine compounds.
2. It is injected into the circulatory system.
3. By their mode of action (electro-mechanical, air pressure activated, or hydraulic).
4. 40 to 260 cc.
5. It is a thermal sleeve placed over a preloaded syringe to maintain the pre-warmed contrast medium temperature at or near body temperature (37°C; 98.6°F); it maintains temperature but does not increase it from room temperature to body temperature.
6. It is composed of a soft bag injector and associated disposables, using a hydraulic, syringe-free injector to deliver contrast media.

614

1. A helium-neon laser within the CR plate reader.
2. The trapped electrons absorb the energy from the scanning laser. This energy is a longer wavelength than the characteristic emissions of the CR plate phosphor, causing the trapped electrons to move from their gaps within the phosphor layer. When this happens, the electrons release energy in the form of a light photon.
3. A bright white light.
4. Usually within less than an hour; any longer and the quality of the image will begin to deteriorate.
5. They should be erased; without erasure, there may be an increased amount of base fog in the exposure.
6. Wet and dry; the primary difference between the two is the way the image is processed. In the wet imager, processing chemicals are used and in the dry imager, no chemicals are used.
7. Processing chemicals are still used, requiring the plumbing connections and tedious cleanups associated with wet film processing.
8. Exposure to a controlled heat of about 140°C for a few seconds.
9. (1) CRT.
   (2) LCD.
10. 3–5 megapixels.
11. Electron spot size changing or “blooming,” causing pixel size to become larger than the nominal or state size. This blooming is affected by luminance and monitor age.
12. Liquid crystal and hydrogenated amorphous silicon (a-Si:H) thin film transistors (TFTs).
13. 90 degrees out of phase with the bottom polarized glass layer.
14. The TG18-QC pattern.
15. They will absorb, reflect, or refract emitted light possibly resulting in erroneous test results.
16. To measure the luminance and illuminance of diagnostic imaging monitors.
17. Within 5% except for very low luminance levels (0.5 cd/m2) where an accuracy of 10% may suffice.
18. The light from the center and outer corners of the monitor.

615

1. Digital Imaging and Communications in Medicine.
2. A means of formatting and exchanging images and is the standard protocol for transmitting medical images across networked devices.
3. (1) Acquisition device.
   (2) Display workstation.
   (3) Network.
   (4) Storage system.
4. HL7 language standard refers to a set of international standards for transfer of clinical and administrative data between software applications used by various healthcare providers. It is used to transmit radiology order information and updates into PACS and the dictation system.

5. (1) Diagnostic workstations have high-resolution displays and allow radiologists to interpret images.
   (2) Clinical review workstations have lower resolution and are used outside radiology, typically by referring physicians for viewing images and data. (Many facilities now use PCs in place of clinical review workstations.)
   (3) Specialty clinical review workstations offer additional specialized functions for use in specific clinical applications.
6. (1) Two or more objects sharing resources and information.
   (2) Computers, terminals, and servers that are interconnected by communication channels sharing data and program resources.
7. (1) Internet Protocol (IP) address.
   (2) Port.
   (3) Application entity (AE) title.
8. DICOM port 104.
9. When images are taken in a deployed setting and an injured service member is transferred to the next level MTF as well as all of his or her radiographic images (CR, DR, and MRI).
10. To apply several security measures like changing default passwords, removing non-essential services or programs, and encrypting files.
11. Storage of digital images.
12. The cache starts to delete the oldest accessed study in a manner known as “first-in, first-out.”
13. 15 to 25MBs.
14. (1) RAID 1.
    (2) RAID 5.

616

1. The ceiling rails and upper tube support. Because after cleaning and inspecting this area, you won’t need to go back and reclean the rest of the unit from dust and dirt that settles.
2. (1) Tighten any loose hardware and replace any screws, nuts, bolts, and so forth that may be missing.
   (2) Inspect the counterweight assembly and tube mounting hardware.
   (3) Inspect all movement and properly lubricate the mechanical paths.
   (4) Inspect all wires and cables for good connections, as well as signs of deterioration or damage.
   (5) Ensure all safety interlock switches are working properly.
3. Smooth travel along the table, and the bucky tray should travel in and out of the bucky smoothly. Also check, adjust, or tighten the pin, which activates the size-sensing arm in the bucky.
4. Possible arcing within the tube wells.
5. (1) Check all connections within the unit to ensure their security.
   (2) Inspect all indicators and any lamps/lights and switches, buttons, or knobs.
   (3) Complete a thorough operational inspection to ensure all features and functions meet the manufacturer’s specifications.

617

1. Some type of battery charging protocol.
2. Exercise the unit by having the technicians take the units into an X-ray room daily and shoot at least 10 consecutive shots.
3. Damaged bearings.
4. (1) AC power cord.
   (2) Exposure hand switch.

618

1. Invasive equipment is actually introduced into the X-ray circuit and requires that connections be broken and reconnected; non-invasive test equipment is simpler to use.
2. Fluoroscopy, survey, and light.

3. (1) d.
   (2) c.
   (3) a.
   (4) f.
   (5) b.
   (6) e.
4. (1) Ionization chamber.
   (2) Solid-state detector.
5. The tube/filter combination.
6. Filtration of the X-ray beam.

619

1. Laptop computer.
2. To validate that your testing sources are arcuate and in compliance.
3. The accuracy of the exposure timer from the shortest duration available to the longest selection available.
4. You cannot select time separately from the mA setting.
5. Only during the initial acceptance of a unit, or if the X-ray tube has been replaced since the last PCRI.
6. 100 mR in one hour while the tube is operated at its maximum continuous rated current (mA) for the maximum rated tube potential (kVp).
7. 37.2 mR/hr.
8. The unit will believe that the tubehead is either closer or further from the patient than it actually is and possibly over or under expose that patient to unnecessary radiation.
9. It stops the tubehead from gliding along its track at specific distances; generally 40-inches in the horizontal table position and 72-inches in the vertical wall stand position.
10. 15.
11. Measure the illumination in the center of each light quadrant at a distance of 100 centimeters (39.5") or the maximum SID, whichever is less, then average the results, and subtract the ambient room light reading.
12. (1) Direct measurement using an X-ray ruler.
    (2) Manually using markers and exposures.
13. Hex wrenches or quarters.
14. That when you adjust the collimator to a specific size, you are radiographing the correct sized field.
15. By measuring the distance between the center point of the radiated area versus the center point of the overall image to determine the offset of the two center points.
16. Field size (PBL).
17. 45 degrees; it allows you to measure the exact dimensions where the auto-collimated exposure cuts off. Since the field size should be automatically collimated to the same size as the film size, you would not see the borders without rotation.
18. To confirm the same amount of radiation (mR) is produced when the same exposure parameters (mA, time, and kVp) are selected. This verifies how consistently a unit will provide the same readings when repeatedly taking the same shot over four consecutive exposures.
19. Tube current output and linearity.
20. The thickness of absorbing material necessary to reduce the X-ray intensity to half its original value.

Complete the unit review exercises before going to the next unit.

Unit Review Exercises

Note to Student: Consider all choices carefully, select the *best* answer to each question, and *circle* the corresponding letter. When you have completed all unit review exercises, transfer your answers to the Field-Scoring Answer Sheet.

Do not return your answer sheet to AFCDA.

35. (608) Inside an X-ray tube housing, what draws the heat away from the X-ray tube?
    a. Circulating fan.
    b. Rotating anode.
    c. Vacuum.
    d. Pure oil.

36. (608) Where can you find a cassette holder in a *standard* radiography room?
    a. Under the table.
    b. On top of the table.
    c. Mounted on the wall.
    d. Under the table or mounted on the wall.

37. (608) The *typical* X-ray room should be lined with
    a. ¼" lead or lead equivalent.
    b. ⅛" lead of lead equivalent.
    c. $^{1}/_{16}$" lead or lead equivalent.
    d. $^{1}/_{32}$" lead or lead equivalent.

38. (609) What reaction occurs at the input phosphor of the image intensifier?
    a. X-rays are changed to photoelectrons.
    b. Light is converted to photoelectrons.
    c. Electrons are converted to light.
    d. X-rays are changed to light.

39. (609) The photocathode in an image intensifier takes visible light and
    a. accelerates it toward the output phosphor.
    b. converts it to photoelectrons.
    c. focuses it for viewing.
    d. amplifies it.

40. (609) Minification gain in an image intensifier is the
    a. ratio of the square diameter of the input phosphor to the square of the diameter of the output phosphor.
    b. ability of the image intensifier tube to increase the illumination level of the image.
    c. smallest size that can be selected for viewing on the image intensifier.
    d. ratio of input light photons to output light photons.

41. (609) What is the *most common* viewing method used during fluoroscopy?
    a. Cine camera.
    b. Mirror-optics.
    c. Spot film camera.
    d. Charge-coupled device (CCD) camera.

42. (610) What are the two *primary* areas within a medical treatment facility where a mobile fluoroscopy unit is used?
    a. Orthopedics and surgery.
    b. Pediatrics and primary care.
    c. Aerospace medicine and cardiopulmonary lab.
    d. Otorhinolaryngology and occupational therapy.

43. (610) A mobile fluoroscopy unit differs from a mobile X-ray unit because the fluoroscopy unit
    a. has limited techniques factors.
    b. has a motor drive system.
    c. operates on line power.
    d. operates on batteries.

44. (611) What is *not* a benefit of using breast compression during mammography?
    a. Brings breast tissue closer to image receptor.
    b. Decreased contrast resolution.
    c. Prevents motion blur.
    d. Lower radiation dose.

45. (611) Because mammographic techniques use a low kilovolt peak and cannot afford attenuation of the X-ray beam, the window of the X-ray tube is made from
    a. tungsten.
    b. rhodium.
    c. beryllium.
    d. molybdenum.

46. (611) Under the Mammography Quality Standards Act (MQSA), how long is a *full* certification good for?
    a. 1 year.
    b. 3 years.
    c. 5 years.
    d. Indefinitely.

47. (612) Which dental X-ray technique shows the roots of the teeth, as well as the surrounding bone?
    a. Occlusal.
    b. Bitewing.
    c. Periapical.
    d. Diagnosal.

48. (612) What dental X-ray unit requires special precautions due to the radiation hazards presented?
    a. Fixed.
    b. Mobile.
    c. Panoramic.
    d. Full mouth.

49. (612) In the panoramic dental X-ray unit, what is done to ensure the radiograph is proportionally accurate?
    a. The film and X-ray tube move in opposite directions.
    b. The film and X-ray tube move in the same direction.
    c. The film moves, but the X-ray tube is stationary.
    d. The film is stationary, but the X-ray tube moves.

50. (613) Which is *not* a classification of contrast injectors?
    a. Air pressure activated.
    b. Electro-mechanical.
    c. Gravity fed.
    d. Hydraulic.

51. (614) In a computed radiography (CR) plate reader, what detects the luminescence from a CR plate after it has been scanned by a laser?
    a. Thin-film transistor array.
    b. Charge-coupled device.
    c. Photomultiplier tube.
    d. Scintillator.

52. (614) What type of light is used within a computed radiography (CR) plate reader to erase a CR plate?
    a. Bright white.
    b. Ultraviolet.
    c. Infrared.
    d. Black.

53. (614) In a dry laser imager, the end of the processing cycle exposes the film to a controlled heat to
    a. remove excess chemicals.
    b. transform the image.
    c. shrink the image.
    d. dry the film.

54. (614) The resolution of a diagnostic imaging monitor is based on
    a. how many pixels can be displayed in the horizontal and vertical dimensions.
    b. the number of gray scale variations of the imaging software.
    c. the contrast depth of the imaging monitor.
    d. the luminance capacity of the backlight.

55. (614) In a monochrome imaging monitor the three subpixels are three varying shades of
    a. red.
    b. blue.
    c. gray.
    d. green.

56. (615) What is *not* one of the four main components of a picture archiving and communications system?
    a. Acquisition device.
    b. Imaging phantom.
    c. Storage system.
    d. Network.

57. (615) What is *not* a type of picture archiving and communications system workstation?
    a. Diagnostic.
    b. Clinical review.
    c. Network clinical review.
    d. Specialty clinical review.

58. (615) Which is *not* one of the three pieces of information that a picture archiving and communications system uses to interface properly?
    a. Internet Protocol (IP) address.
    b. Application entity (AE) title.
    c. Domain name.
    d. Port.

59. (615) What are the two *main* redundant arrays of independent disks (RAID) storage configurations used in a Picture Archiving and Communications System (PACS)?
    a. RAID 0 and RAID 1.
    b. RAID 0 and RAID 5.
    c. RAID 1 and RAID 2.
    d. RAID 1 and RAID 5.

60. (616) When performing scheduled maintenance on a standard X-ray room, where is the *best* place to start your inspection?
    a. X-ray tube housing.
    b. X-ray generator.
    c. Ceiling rails.
    d. Wall stand.

61. (616) What prevents arcing inside the tube wells of an X-ray tube?
    a. Dielectric compound.
    b. Heat sink compound.
    c. Insulating oil.
    d. Air.

62. (616) How would you *normally* check the bearings of an X-ray tube rotor?
    a. By removing the X-ray tube from the housing and manually spinning the rotor.
    b. By listening to the rotor spin during the operational inspection.
    c. By visually inspecting the bearings.
    d. The bearings cannot be checked.

63. (617) *Normally*, how often should equalization cycles on mobile X-ray unit batteries be performed?
    a. Weekly.
    b. Monthly.
    c. Semiannually.
    d. Annually.

64. (618) Patient dose of X-rays is the energy absorbed per unit mass and is usually expressed in terms of
    a. grays.
    b. roentgen.
    c. milliamps per second.
    d. millimeters of aluminum.

65. (618) Which non-invasive method of measuring radiation (dose) uses an air-filled chamber between two collecting electrodes with a difference of potential of approximately 300 volts?
    a. Solid-state detector.
    b. Ionization chamber.
    c. Dual detector.
    d. Voltage ratio.

66. (619) What is the *minimum* collimator light illumination requirement in a post calibration radiation inspection (PCRI)?
    a. 5 footcandles (FC) below ambient room light.
    b. 5 FC above ambient room light.
    c. 15 FC below ambient room light.
    d. 15 FC above ambient room light.

67. (619) Which post calibration radiation inspection (PCRI) test verifies how consistently a unit will provide the same readings when repeating the same radiographic exposure?
    a. Tube current output and linearity.
    b. Positive beam limitation (PBL).
    c. Half value layer (HVL).
    d. Reproducibility.

68. (619) What is the half value layer (HVL) aluminum (Al) equivalent requirement for the beam quality test in a post calibration radiation inspection (PCRI)?
    a. 0.5 mm Al for single-phase systems, and 3.2 mm Al for 3-phase systems.
    b. 1.5 mm Al for single-phase systems, and 2.2 mm Al for 3-phase systems.
    c. 2.5 mm Al for single-phase systems, and 3.2 mm Al for 3-phase systems.
    d. 3.5 mm Al for single-phase systems, and 2.2 mm Al for 3-phase systems.

Please read the unit menu for unit 3 and continue ➔

Student Notes

Unit 3. Special Equipment and Procedures

THE DISCOVERY of advanced specialty imaging techniques has enhanced the way we diagnose diseases and potential clinical issues in healthcare. The responsibilities of the radiology department, as well as BMETs, continue to grow as we incorporate new technology across diagnostic imaging modalities. The equipment, as well as training required for these platforms, can be very expensive; however, it has a place in all MTFs.

In this unit, we will discuss multiple forms of tomography and nuclear medicine, both of which take full advantage of ionizing radiation. We will also discuss newer technology that does not require irradiation of the patient. MRI and ultrasound have found their way into the MTF and are proving they are extremely valuable technologies. We will look at these technologies to give you more insight into how they work and the equipment that makes it happen.

3–1. Conventional and Computerized Tomography

Conventional tomography is a radiographic technique that was first developed in 1921 and for a long time was the premier investigative procedure for removing superimposing structures before the invention of CT and MRI. Tomography has been called planigraphy, stratigraphy, and laminography over the years, and it is still sometimes called body section radiography.

CT is perhaps the most significant discovery in this field since Roentgen discovered X-rays in 1895. In the past, it has been called computerized axial tomography (CAT), computerized transverse axial tomography (CTAT), reconstructive tomography (RT), and computerized or computed tomography. CT is the term that has been accepted to identify this diagnostic tool.

CT has not been in use very long. The first CT head scanner was installed at Atkinson Morley's Hospital, London, England in 1971. The Mayo Clinic and Massachusetts General Hospital obtained the first units for the US in 1973. The first whole-body scanner was installed at Georgetown University Medical Center in 1974. Since then, CT has gone through many improvements; however, the fundamentals have remained the same. Over the course of this section, you will learn the theory and equipment behind tomography, CT, and cone beam computed tomography (CBCT).

620. Tomography

With conventional methods of radiography, images are often superimposed and obscure the structure under study. Sometimes stereoscopy, tube or film angles, along with additional projections, may be used to alleviate this condition. However, under some conditions, tomography may be required to eliminate superimposition.

Our study of tomography begins with a review of terminology and the types of tomographic systems. After that, we'll explain the principles involved and finish up with some specific tomographic examinations.

Terminology

Tomography describes all body sectioning techniques using similar principles according to the International Commission on Radiation Units and Measurements (ICRU). A radiograph produced by these techniques is called a tomogram.

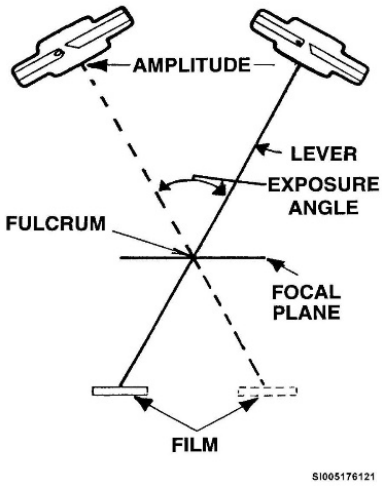

Figure 3–1. Terminology diagram

A lever is a connecting rod that couples the tube and image receptor so the tube and film movements are proportional (fig. 3–1). The fulcrum is the point about which the lever pivots. When a tomogram is performed, the fulcrum is adjusted to correspond to the layer or plane to be radiographed.

The focal plane is the layer of the body that appears sharpest on the tomogram. Blurring, in tomography, represents the unsharp body area outside (above or below) the focal plane.

Amplitude is the distance (expressed in inches) the tube travels during exposure. Rate is the speed of tube travel (usually expressed in inches per second).

Exposure angle is the angle (in degrees) of the tube travel during exposure. Zonography is tomography with a small exposure angle (less than 10°).

Tomographic tube movements

The more a tomographic motion differs from the shape of the object being radiographed, the more uniform the image. Tomography was first achieved with a linear motion. Although numerous motions and patterns to minimize section thickness were developed, only the linear type motions (linear and curvilinear) remain in common use (fig. 3–2). The simplest and most common is rectilinear (or linear) tomography. In linear tomography, the tube and image receptor move in a straight line along the axis of the table. Linear motion has a major quality problem however, because the SID and object to image distance (OID) change as the tube moves. Both SID and OID are greater at the extreme left and right positions than at the center position.

Linear tomography remains popular because it is an inexpensive addition to a diagnostic X-ray unit. It requires a rod to attach the X-ray tube and the bucky tray, an adjustable fulcrum through which the rod may pivot, and a motor to drive the X-ray tube stand. All other tomographic motions require specialized units designed to provide the desired motion through complicated tracks, gears, pivots, and so forth.

Curvilinear tomography improves on the linear motion by maintaining SID and OID and reducing magnification differences.

Principles of tomographic operation

If you have seen a radiograph of a chest, the first thing you notice is the ribs superimposed over the lung tissue. The ribs are the easiest part to visualize! What if the radiologist was looking for a lesion on the lung tissue—would it be as easily viewed on this radiograph? No. That is why tomography was invented—so we can eliminate the ribs and only record the lung tissue.

Movement

Not only can the ribs be removed, but also any anatomical part that might be superimposed on a structure that needs to be radiographed. Figure 3–3 shows three structures in a vertical plane. Assume the triangle is the suspected lung lesion, and the circle and square are the ribs. Using the tomographic concept, the circle is projected to the right of the triangle as the exposure begins, but is blurred completely across the triangle to appear at the left by the end of the exposure. Also note that the square goes through the same blurring, but in the opposite direction.

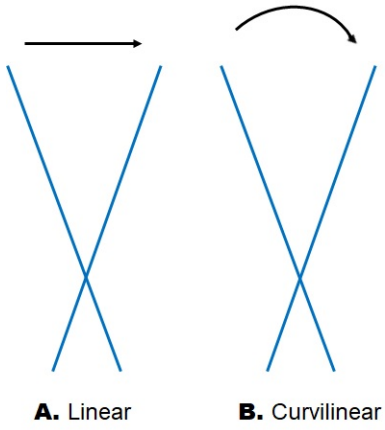

Figure 3–2. Tomographic tube movements.

This change of the projected positions of the shapes blurs the images and basically eliminates the unwanted objects from the radiograph. The projected triangle stays in the center of the film throughout the movement, and therefore, is not blurred. Consequently, the lung lesion is visualized without the superimposed shadows of the ribs.

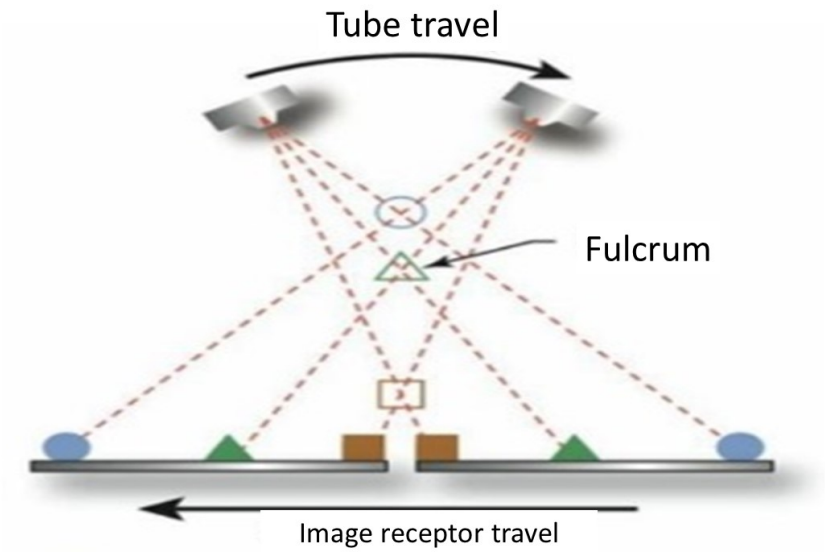

Figure 3–3. Tomographic movement and fulcrum position. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

You may wonder why the structures in the focal plane are not blurred. Remember, we stated the focal plane is the layer of the body that appears the sharpest on a tomogram. The longer the blurring, the less opportunity to create a sharp image. The shorter the blurring, the sharper the image. Therefore, there is a direct relationship between distance from the fulcrum and blurring. The greater the distance from the fulcrum, the greater the blurring, and vice versa.

Those images that lie in the plane of the fulcrum will be projected onto exactly the same location on the image receptor because the image receptor is moving at exactly the proper rate to maintain their location. Objects located above and below the fulcrum will be projected onto varying locations on the image receptor as it moves, thus blurring their images.

Section thickness

When looking at a tomogram, you immediately see some areas are blurred and cannot be read. You will generally find, in the center of the image, a section considered sharp enough for diagnostic purposes. This sharp area is sandwiched between areas of blurring. Figure 3–4 shows a schematic of a tomogram. There is a blurred area leading into a 10-mm section preceding another blurred area. The 10-mm section is the focal plane or thickness of the section. Now, 10 mm is not the only size section you can get. The section thickness depends upon the exposure angle used during the tomogram. Exposure angle is inversely proportional to section thickness. As the exposure angle increases, section thickness decreases.

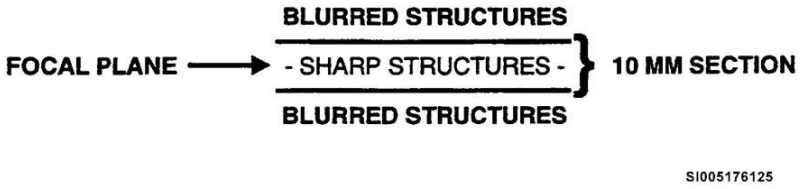

Figure 3-4. Schematic of a tomogram.

Exposure angle

Figure 3–5 illustrates tube movements through two different exposure angles. In detail A, the projected dots on the film move considerably less than the dots in detail B. This should lead you to the conclusion that detail A, with the less movement, will have less blurring above and below the focal plane. Therefore, more of the radiograph is readable—thus, a thicker section or slice. In detail B, the projected dots move over a larger distance. More movement means more blurring; this means less of a readable area in the tomogram—hence, a smaller or thinner slice.

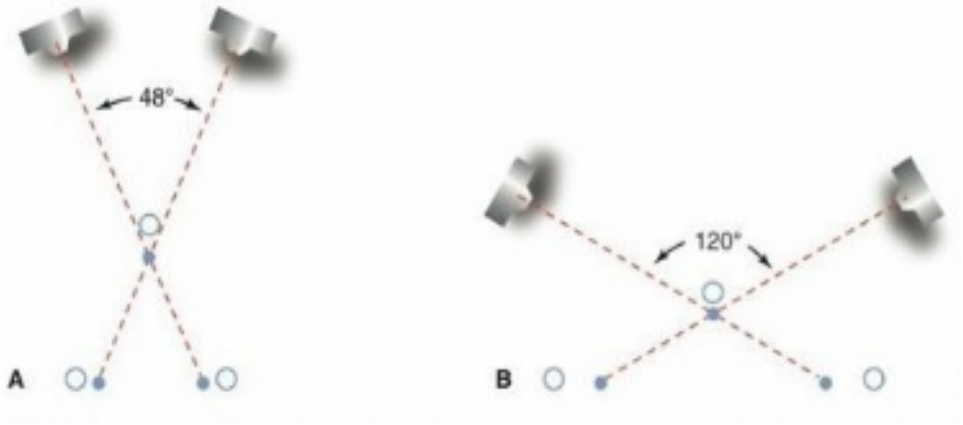

Figure 3–5. The relationship of exposure angle and slice thickness. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

There are four factors that affect the exposure angle:

- Focus to film distance (FFD)—how far the image receptor is from the focal plane.
- Amplitude—total distance the tube travels.
- Focus to object distance (FOD)—how far the object being imaged is from the focal plane.
- Distance of the object from the image receptor.

If the amplitude is constant, a longer FFD produces a thicker section than does a shorter FFD. Therefore, if the movement of the tube is the same, and the distance from the fulcrum (focal plan/focus point) to the image receptor is increased, the angle of the projection changes. Likewise, if the FFD is constant, a short amplitude produces a thicker section than does a long amplitude. The exposure angle is smaller because of the short amplitude.

Digital tomosynthesis

Digital radiographic tomography is called digital tomosynthesis because the computed radiographic images are manipulated by a post-acquisition algorithm to simulate tomographic exposures instead of actually producing additional images. Digital tomosynthesis combines digital image capture and processing with the simple tube/detector motion used in conventional radiographic tomography. Although it is similar to CT, it is a separate process. Tomosynthetic processing requires about 10 exposures to establish an adequate image base from which reprocessing can be done. Higher-resolution detectors are used, thus allowing very high in-plane resolution, even if the z-axis resolution is poor. The primary interest in tomosynthesis is in breast imaging, as an extension to mammography.

Reconstruction algorithms for tomosynthesis are significantly different from conventional CT. Flat panel image receptors (as used in direct computed radiography) are best for tomosynthetic systems because they provide rapid readout. The image dataset is acquired by making about 10 static exposures at slightly different central ray angles through the region of interest (ROI). Figure 3–6 explains the tomosynthetic process in detail. As you can see, there are three objects at different depths labeled at planes 1, 2, and 3. You will also notice that at various image receptor positions, the viewing angle changes. The plane reconstruction shows how each object becomes the focal point at each respective plane, blurring out the other shapes due to grouping distance. Post-acquisition image processing permits reconstruction of any desired plane through the exposed area. In addition, post-acquisition algorithms for modifying blur, density/brightness, contrast, and other quality factors can be used as necessary.

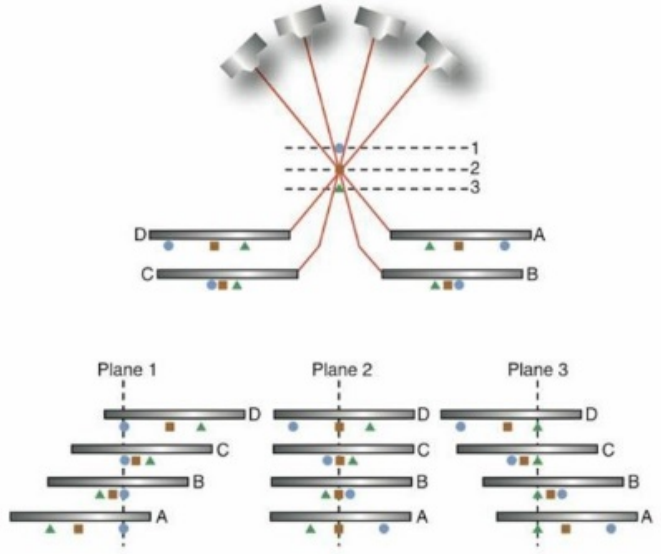

Figure 3–6. Three formulas for determining tomographic factors. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

621. Computed tomography principles

In this lesson, we'll learn about the clinical applications of CT and the principles of operation behind it. The first thing you need to know is for what procedures CT scans are used.

Clinical applications

CT scans have proven to be extremely useful and are used to image a wide variety of body structures and internal organs. In some diagnoses, physicians may choose a CT scan before any other imaging exam. There are many advantages of a CT scan over a traditional X-ray, but probably the largest benefit is the three-dimensional (3D) view a CT scan provides. Along with a 3D view, the computerized image produced by a CT scan is much sharper, more focused, and allows many tissues to be better differentiated than on standard X-rays. Common uses of CT scans include:

- Sinus studies—the CT scan can show details of sinusitis and bone fractures. Physicians may order CT of the sinuses to provide an accurate map for surgery.
- Brain studies—brain scans can detect hematomas, tumors, and strokes. The introduction of CT scanning, especially spiral CT, has helped reduce the need for more invasive procedures, such as cerebral angiography.
- Body scans—CT scans of the body are often used to observe abdominal organs (i.e., the liver, kidneys, adrenal glands, spleen, and lymph nodes) and extremities.
- Aorta scans—CT scans can focus on the thoracic or abdominal aorta to locate aneurysms and other possible aortic diseases.
- Chest scans—CT scans of the chest are useful in distinguishing tumors and in detailing accumulation of fluid in chest infections.

NOTE: In obese patients, CT scanning may be more useful than ultrasound, since large amounts of body fat can interfere with ultrasound waves.

CT scans can also be used in cancer detection to scan for abnormal masses, which might be malignant tumors. CT scans can show the size and shape of a tumor, its precise location in the body, and whether it's solid or hollow. Although a CT scan sometimes is able to tell the difference between a benign (noncancerous) or malignant (cancerous) tumor, the final diagnosis is made by a biopsy or other test. When a needle biopsy is performed for cancer diagnosis, CT scanning also can be used to guide the insertion of the biopsy needle into precisely the right location for sampling a tumor.

Contrast agents are often used in CT exams to illuminate certain details of the anatomy, which may not be easily seen. Some contrasts are natural, such as air or water; at other times, a water-based contrast agent is administered for specific diagnostic purposes. Barium sulfate is commonly used in gastroenterology procedures. The patient may drink this contrast or receive it in an enema. Oral and rectal contrast are usually given when examining the abdomen or cells, and not given when scanning the brain or chest. Iodine is the most widely used intravenous (IV) contrast agent and is given through an IV needle.

Principles of operation

Computerized tomography is a specialized modality that links the basic theory of body section radiography with a computer system to produce anatomical images. CT produces a digital tomographic image from diagnostic X-rays. By definition, CT is the process of creating a cross-sectional tomographic plane (slice) of any part of the body. By using X-ray absorption measurements collected at multiple points about the body part's periphery, a computer creates this reconstructed image through mathematical algorithms.

Primarily, the image differentiates various types of soft tissues (e.g., gray matter, white matter, blood, tumor, cerebrospinal fluid, etc.). When digital image data manipulation is added, CT provides more diagnostic information than any other ionizing radiation imaging modality. Low density tissue, usually obscured by higher density structures on a conventional radiograph, can be clearly visualized

with CT. This is why CT is excellent for evaluating the brain. A conventional radiograph of the skull is of little value in trying to visualize the brain because the high-density cranial vault obscures low-density brain tissue. Following this same line of thought, CT can see more minute differences in X-ray attenuation than can be recorded by conventional radiography. For example, conventional radiography needs a minimum difference in tissue density of 2–5 percent to separate structures. Computerized tomography can show adjacent structures that have a difference in tissue density as low as 0.5 percent. This makes CT an excellent tool for showing contrast differences of soft tissue structures within the brain and abdomen.

Because a CT image shows the entire cross section of a slice of the anatomy, the size and location of a pathologic condition can be seen accurately. Conversely, conventional radiography often needs contrast media and multiple exposures to estimate the size and location of a diseased area. Furthermore, some units have the ability to demonstrate a sagittal (fig. 3–7, A) and a coronal (fig. 3–7, C) reconstructed image of the neck and upper thorax from data obtained from various contiguous transverse (fig. 3–7, B) slices with no additional exposures. This process produces a lateral or anterior-posterior tomographic plane similar to conventional tomography. Because the CT X-ray beam is confined to a slit, much scatter radiation is eliminated from the image. The combination of the transverse sectioning procedure with slit scanning produces an image of significantly better quality than that available with other imaging methods.

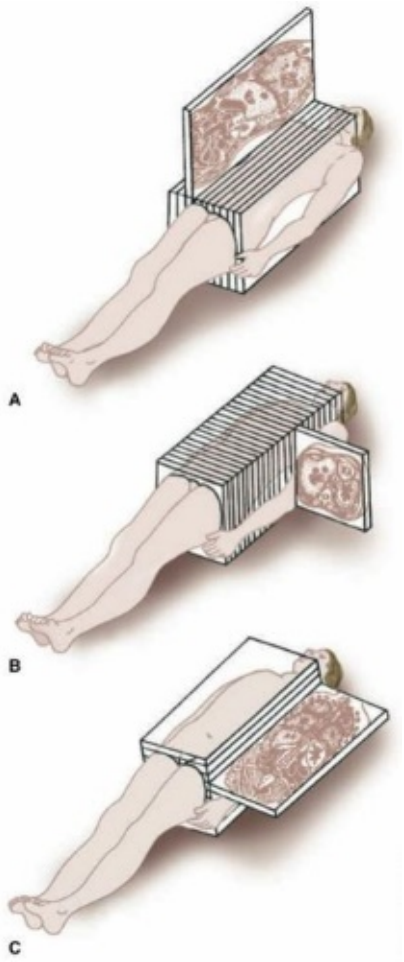

Figure 3–7. Sagittal, transverse, and coronal body slices. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

We have listed many advantages of CT, but it also has a limitation, mainly, the amount of time it takes to perform a single examination. Most exams take from 2–10 seconds, which is a long time considering radiation is being produced throughout the procedure.

Categories of scanners

Computerized tomography scanners are often best identified in terms of their geometry and detector array (number of detectors). Sometimes, they are also described by their generation. Generation is a term used in technical fields to denote an improvement or change in model. In some cases, the generation can be used to signify an improvement in capability and, therefore, an improvement in quality.

First-generation scanner

A basic configuration was used with first-generation scanners. It consisted of an X-ray tube connected mechanically to a detector array (fig. 3–8). The detector array consisted of two scintillation crystal detectors (some had only one) coupled to a photomultiplier. The X-ray beam was collimated to a pencil beam measuring about 2 mm by 16 mm.

The scanning motion consisted of translation and rotation. Translation means to move without a change in angulation; rotation means to change angulation. The X-ray tube/detector array assembly thus moves directly across the body part; it then indexes one degree and repeats the translation, rotates 1° and translates again, and so on until it transverses 180° of angulation. During this period, the detector array senses the remnant radiation and sends electronic signals to the computer.

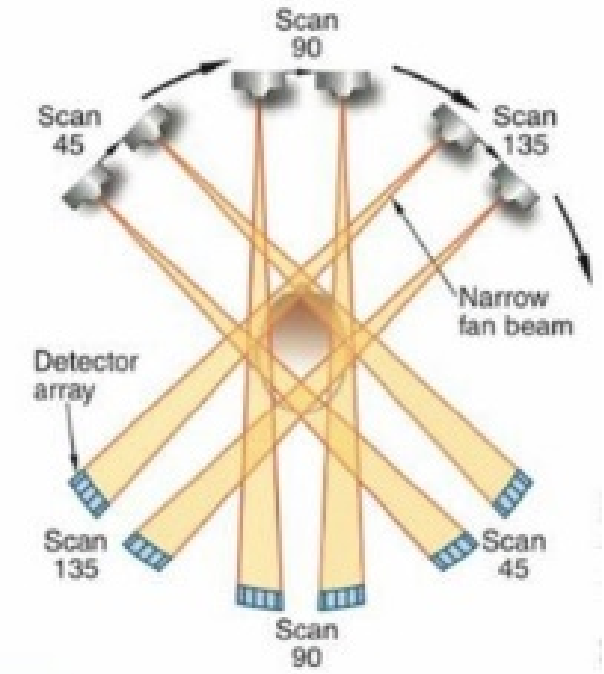

Figure 3–8. First generation CT scanner. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

These first-generation scanners offer accurate data readings for reconstructing images. This is due to its geometry—specifically 1° rotations. The principal drawback to these units was the 3–5 minutes required to complete one scan.

Second-generation scanner

A second generation of scanners was developed to reduce the scanning time. The second-generation scanners use a fan shaped beam, rather than a pencil beam, and a linear detector array of 3–60

detectors (fig. 3–9). Multiple scintillation detectors are mechanically connected to the tube head assembly for proper movement. These second generation units utilize translation rotation movement like their first generation predecessors.

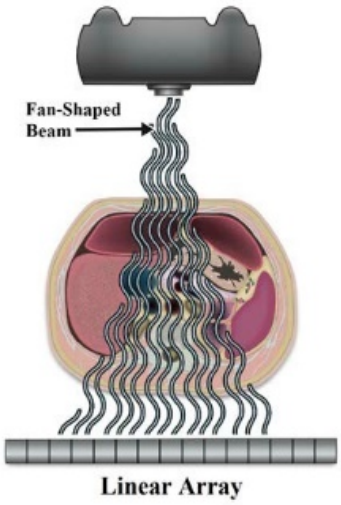

Figure 3–9. Second generation CT scanner.

The major advantage of second-generation scanners was increased speed. A single translation with the fan beam and multiple detector arrays gave the same number of data points as several translations with a first-generation scanner. Also, each translation is separated by rotation increments of 5° or more. This means that fewer exposures were needed for a 180° scan. Scanning times were reduced to 10–18 seconds per slice. Although the reduced scanning time is a significant improvement over the first scanner, it does not prevent artifacts due to peristalsis, vascular pulsations, or respiratory motion. Unfortunately, the fan beams increase the overall radiation exposure to the patient and the amount of scatter radiation, which can affect the final image in much the same way as in conventional radiography.

Third-generation scanner

A third-generation unit further reduces examination time through its 360°, rotation-only movement. This category of scanner uses a curvilinear detector array of at least 128–750 detectors, and a narrow fan beam (fig. 3–10). The number of detectors and width of the fan beam (30°–60°) are substantially larger than for second-generation scanners. The narrow fan beam provides a pulsed X-ray beam.

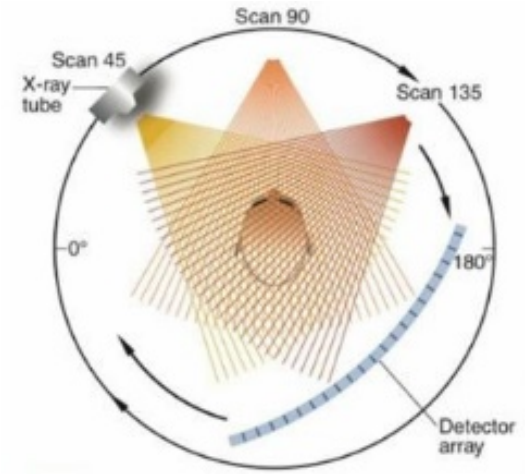

Figure 3–10. Third generation CT scanner. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

In third-generation scanners, the fan beam and detector array view the entire patient slice thickness at all times. The curvilinear detector array gives a constant source-to-detector path length, which is an advantage for good image reconstruction. This also allows for better X-ray beam collimation to reduce the effect of scatter radiation. This type of collimation is called predetector collimation and functions like a radiographic grid in conventional radiography. Predetector collimation does not eliminate the need for prepatient collimation, which restricts patient dose.

The major advantage of third-generation scanners is they reduce scanning time to one second. A major disadvantage is the occasional appearance of circular or ring pattern artifacts caused by a malfunction of a single detector or bank of detectors.

Fourth-generation scanner

In the fourth generation scanning process, the X-ray tube rotates around a stationary circular set of multiple detectors (rotate-stationary). These detectors completely encircle the patient (fig. 3–11). Unlike the third-generation units (rotate-rotate), the detectors do not move when the X-ray tube is rotated. A wide fan beam enters the patient around the edge of the slice being imaged. The scanning process consists of the X-ray tube making a 360° rotation around the patient; one rotation of the X-ray tube creates one image. A typical CT study comprises many images at different slice locations. The slice scans can be made with a rotation smaller or larger than the standard 360° movement. This parameter can be adjusted and is usually set by the radiographer performing the study. During the rotation, a continuous X-ray beam projects information to the array of detectors, which has the result of seamlessly producing the image from multiple X-ray tube locations using one detector.

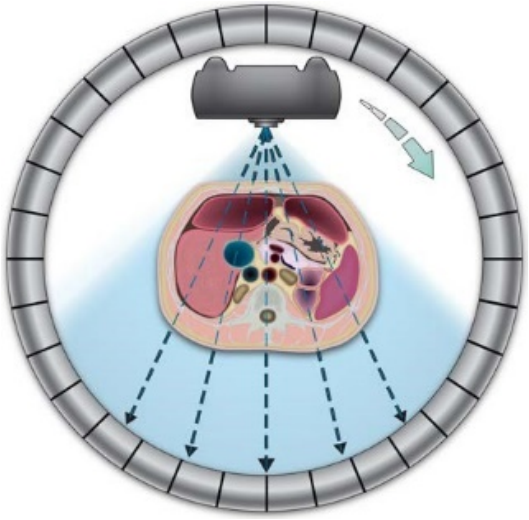

Figure 3–11. Fourth generation CT scanner.

Helical or spiral scanner

In the previous CT systems, only a single slice can be acquired at one time. If multiple slices are required to cover a large part of the body, the patient table is moved in discrete steps through the plane of the X-ray beam and detector. A single slice is acquired at each position with a time delay between to obtain the image. This process is inefficient and can result in spatial misrepresentations between slices if the patient moves. To compensate for this, a technique called helical or spiral CT was developed. This type of scanner works by acquiring data as the patient is moved continuously through the scanner. The trajectory of the X-ray beam through the patient traces out a spiral, which gives it its name. This technique greatly improves scan times. For example, with a typical fourth-generation scanner, the time for a complete chest and abdominal study could be as much as ten minutes. With a helical scanner, the time is reduced to around one minute.

Single-slice versus helical computed tomography

Single-slice (or conventional) CT obtains image data one slice at a time. This method is sometimes referred to in the CT system software as axial mode. Basically, with the CT couch (table) not moving, a slice is acquired, then the table moves into the next position where another slice is acquired, and the table moves again, and so on until all individual axial plane slices have been acquired.

Helical (or spiral) CT images are acquired while the table moves the patient at a constant speed through the exposure area (the gantry), while the X-ray tube continuously rotates around the patient in a "corkscrew" manner. Helical CT scanning allows large volumes of chest or abdominal tissue to be scanned on a single breath-hold, thereby, eliminating respiratory motion and improving detection of small lesions. Volume acquisition, using a helical scan mode, enables high-quality post-processing reconstruction of multiple overlapping slices. This improves visualization of small pathology and allows for high-detail 3D imaging.

Multidetector helical computed tomography imaging systems (multi-slice)

Multidetector helical CT (MDCT) is a major technical advancement in CT imaging. It uses the principles of helical scanning, while incorporating multiple rows of detector rings. This technique allows you to acquire multiple slices per tube rotation, increasing the area of the patient that the X-ray beam can cover in a given time. Available systems have moved quickly from 2-slice to 64-slice, which covers 40 mm of patient length for each 1-second or less of tube rotation. Though 256-slice scanners have been developed, the current workhorse MDCT scanner in most departments is the 16-slice scanner, with 64-slice scanners becoming increasingly popular for cardiac applications like coronary angiography.

The key advantage of MDCT is speed. It is 5–8 times faster than single-slice helical CT units. For body scanning, 1 mm slices can be obtained, creating isotropic voxels (voxels that are a perfect cube; equal in length, width, and height [e.g., 1 x 1 x 1 mm]) that allow image reconstruction in any anatomic plane without loss of resolution. Broad area coverage allows for high-detail CT angiography, "virtual" CT colonoscopy, and bronchoscopy. However, nothing is free, and a significant disadvantage of MDCT is radiation dose, which can be 3–5 times higher with MDCT than with single-slice CT. Thin slices (some as small as 0.4 mm) and the multiple-detector acquisition add great diagnostic capability, unfortunately, at the cost of increased radiation dose to the patient.

MDCT has introduced pitch (or spiral pitch) to the field of CT. Pitch is simply the ratio of the distance the table moves (feed) during one 360° tube rotation to the total beam collimation. Pitch also allows the patient table to be moved faster or slower while the tube continues to rotate around the patient for continuous exposure. Being able to use pitch is a major advantage of MDCT. Pitch allows for a larger volume of tissue to be imaged in a single scan or breath-hold (chest or abdomen).

Pitch is stated as a ratio, such as 0.5:1 (0.5 to 1), 1:1 (1 to 1), 1.6:1 (1.6 to 1), and so on. A pitch of less than 1:1 (e.g., 0.5:1) slows the table movement, which allows images to overlap but includes a higher radiation dose to the patient. When a pitch of greater than 1:1 (e.g., 1.6:1) is used, the table movement speeds up, resulting in greater coverage of the area being imaged and a reduction in radiation dose to the patient. Most times, in multidetector helical CT scanning, the pitch is 1:1.

Computed tomography fluoroscopy

CT fluoroscopy is an advancement in CT technology that allows for real-time CT imaging. This technique dramatically improves the ability to perform percutaneous interventions quickly and at a generally lower radiation dose than with conventional CT. The operator can step on a floor pedal while moving the CT table or observing patient motion. Rapid image reconstruction provides real-time images of the anatomy, lesions, and needle or catheter placement. CT fluoroscopy is now routinely used to guide biopsy, drainage, and interventional procedures anywhere in the body. It is particularly useful in guiding needle placements where there is physiologic motion, such as in the chest and abdomen.

Dual-source computed tomography imaging systems

Dual-source CT, or dual-energy, CT (sometimes referred to as sixth-generation scanners) uses two X-ray sources (tubes) and two X-ray detectors that expose tissues simultaneously to determine how tissue behaves at different radiation energy levels. This technique allows the addition of information about tissue composition. Differences in fat, soft tissue, and contrast agents, at different energy levels expand lesion identification and characterization. Image data can be captured in half the time required for MDCT. This vastly improves the ability to image the heart without using potentially dangerous beta-blockers (a drug used to slow the heart rate).

Though it seems a patient's radiation dose would dramatically increase with the addition of a second X-ray tube (source), the radiation dose is actually lower due to the reduction in total scan time. It may be reduced even more if certain pre-contrast scans are eliminated in conjunction with multi-phased studies.

Image production

CT produces an image by the attenuation of the X-ray beam; the patient absorbs the radiation in varying amounts depending on its interaction with the various tissues. The remnant radiation is collected by the detectors and transmitted to the computer for processing. The measurements acquired from the scan are recorded in digital form by the computer. From this information, the computer reconstructs the image. Computer software processes the data and runs the image reconstruction process. An algorithm sorts and manipulates the digital information to reconstruct the image.

The CT image is actually a composition of small blocks, or cells, arranged in rows and columns called a matrix. As discussed previously, the computer assigns a number that represents the density of the structure contained in each cell of the image. If each number represents a shade of gray and projects the matrix onto a monitor, we end up with a viewable image of the body part. The most common matrix sizes used in CT scanning are 256 x 256, 512 x 512, and 1024 x 1024. As the image matrix gets larger, the individual cells become smaller on the screen, and the image detail (resolution) increases.

Pixels

Each of the cells of a matrix is called a pixel, which stands for picture element. A pixel on a CT image is a two-dimensional representation of the average density of a volume of tissue. A display monitor provides an image by painting dots or pixels, each with its own gray shade ranging from black to white. These pixels are the smallest objects that can be imaged by the monitor. If a monitor has a pixel dimension of 512 x 512 pixels, and the field size of the CT scanner is 48 cm, the smallest imaging area of a patient can be calculated using the following formula:

$$\text{Imaging area} = \frac{\text{pixel dimension}}{\text{CT scanner field size}}.$$

Using this formula and the information from our example, let's determine the smallest imaging area of a patient:

$$\text{Imaging area} = \frac{48\text{ cm}}{512\text{ pixels}} = 0.09375\text{cm}\ (0.9375\text{mm}).$$

Voxel/attenuation coefficient

The area of a patient that corresponds to a pixel on the monitor is referred to as a voxel or volume element. A voxel is the area of a pixel multiplied by the thickness of the slice. You can see the relationship of a pixel to a voxel represented in figure 3–12.

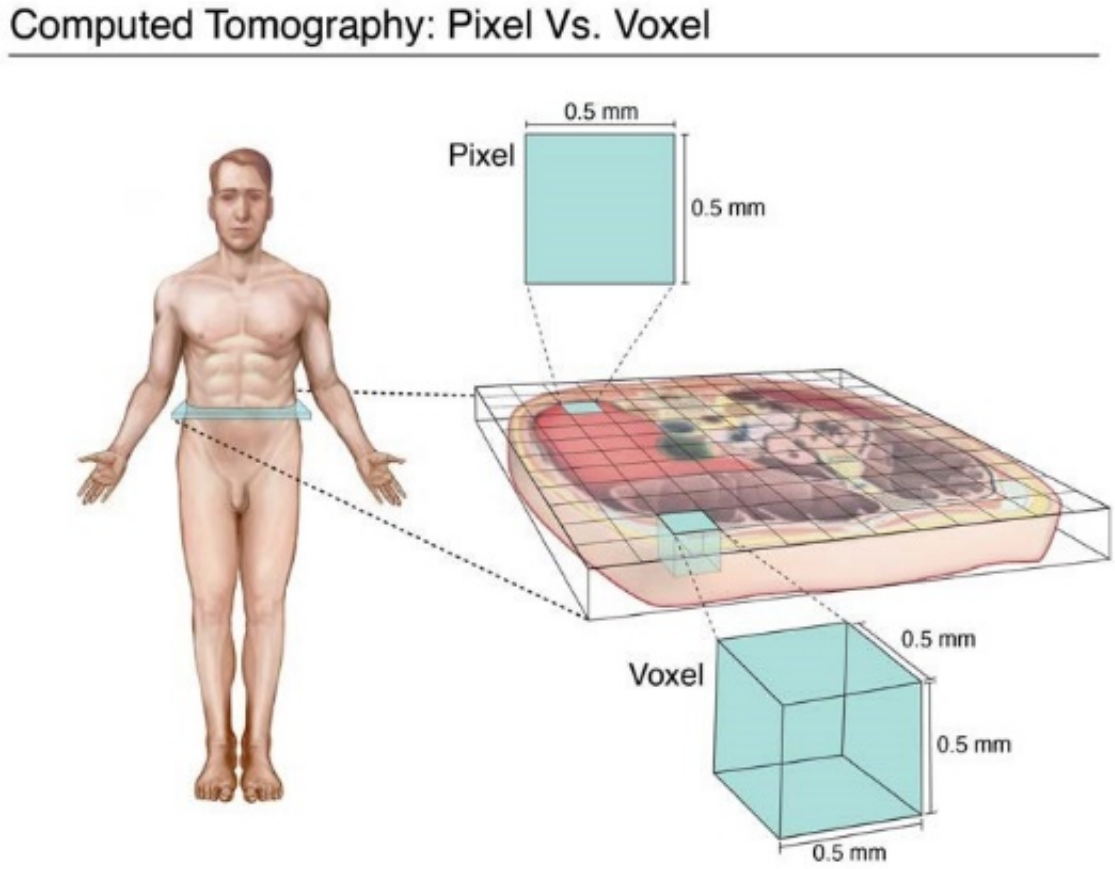

Figure 3–12. Pixel and voxel CT slice illustration.

As X-rays pass through the voxel, a certain amount of the X-ray energy is absorbed, depending on the material of the voxel. This amount of absorption is called attenuation coefficient or μ. The formula for μ is:

$$\mu X = 1n\left(\frac{Io}{It}\right).$$

In the formula, X equals the thickness of the field of the scan, 1n equals the natural log, Io equals the incident X-ray radiation from the tube, and It equals the transmitted X-ray radiation from the voxel.

Ray sums

Each projection (data acquisition cycle) consists of several ray sums. Each ray sum is considered to be the information received by one detector during one data acquisition cycle (fig. 3–13). A ray sum is the sum of all μ of the voxels between the X-ray tube and receiving detector. That is, μX1 + μX2 + μX3 + μX4 + μXn, and so on. The job of the reconstruction circuits is to separate the different values of each μ within each ray sum. This is accomplished during the “back projection” process.

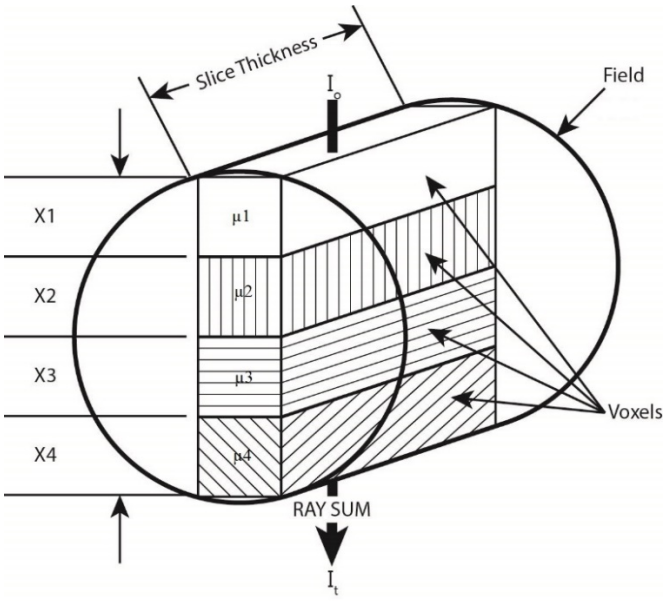

Figure 3–13. Ray sums, each lined area represents a different shade of gray.

Projections into the memory array

The primary mathematical method used in the creation of computerized medical images is the Fourier transformation. Transformations are simply conversions of data to more useful forms, as when radiation doses are changed from rems to mSv. Fourier's mathematics accomplishes transformations of extremely complex functions into separate but simpler functions. In digital imaging, the Fourier transformation is used on data representing image intensities at specific locations. For example, Figure 3–14 demonstrates different attenuation coefficients at specific image receptor locations. The image receptor data correlates to specific spatial locations. The information received by the image receptors can be processed through a variety of mathematical formulas (often referred to as algorithms or kernels). The Fourier algorithm is a fundamental formula used in image reconstruction. It is based on algebraically adding several sets of incoming data from the image receptor.

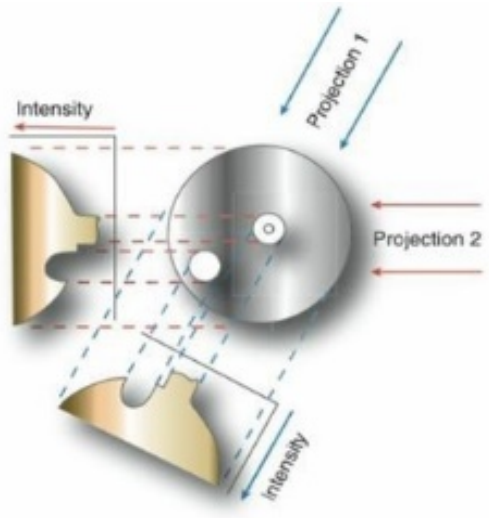

Figure 3–14. Image receptor location to attenuation coefficient. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

CT numbers

Each pixel is displayed on the video monitor as a level of brightness and on the photographic image as a level of optical density. These levels correspond to a range of CT numbers from –1,000 – +1,000 for each pixel. A CT number of –1,000 corresponds to air; a CT number of +1,000 corresponds to dense bone. A CT number of zero indicates water. The precise CT number of any given pixel is related to the X-ray μ of the tissue contained in the voxel. The following table gives the CT number for various components found within the human body. You need to remember these numbers are calculated using the linear attenuation coefficient.

Tissue	CT Number
Air	–1,000
Fat	–100
Water	0
Cerebrospinal fluid	15
White matter	46
Gray matter	43
Blood	40
Dense bone	+1,000

Windowing

The transmitted X-rays represent some amount of attenuation in the patient. These are compared with the intensity of the radiation from the tube, which is measured by a special reference detector, to give relative transmission values after digitization. The attenuation values of various tissues are related to the attenuation value of water, and may be arranged on a scale. These scales are called EMI, or Hounsfield numbers, and represent the various CT digital numbers used to reconstruct the image. When the image is produced, the scale (CT digital) numbers represent a certain brightness level. The brightness level (gray scale) can be manipulated to demonstrate different structures in the image. This manipulation or variation in the relation between the scale numbers and level of brightness is often called windowing or setting a window.

The window is controlled by the operator of the CT unit and is usually set as a window-width and window-level. The window-width represents the range of scale numbers used for the gray scale. Adjusting the window width is equal to adjusting the contrast of the image. The window-level represents the midpoint of the gray scale and can be considered a density adjustment. When viewing an image, the radiographer can usually adjust these values to enhance the anatomical structures.

622. Computed tomography system components

CT systems can be broken down into three main groups—imaging, computer, and control. A fourth group takes into consideration image storage.

Imaging group

The imaging group contains all the elements needed to produce an image, including the X-ray generator, X-ray tube, detector system, and Data Acquisition System (DAS). The equipment varies according to which generation scanner you have.

Gantry

The gantry itself is the largest component of CT scanner systems. It is a housing that contains the X-ray tube, the detector assembly, slip rings, collimators, DAS, and also contains the mechanics that provide motion used in the CT unit (fig. 3–15). The gantry housing conceals the motion of the tube and detectors. The gantry has a hole in its center, named the aperture, into which the patient transverses in and out of for the CT scan. The gantry may be tilted, forward or backwards up to

30 degrees, to obtain a more accurate cross-sectional image of an angled body part; the gantry tilt feature is most often used when imaging the head and spine. With the images collected from a series of axial scans, it is possible for the computer to combine segments of the images to create a new image in an orthogonal plane (a plane composed of right angles).

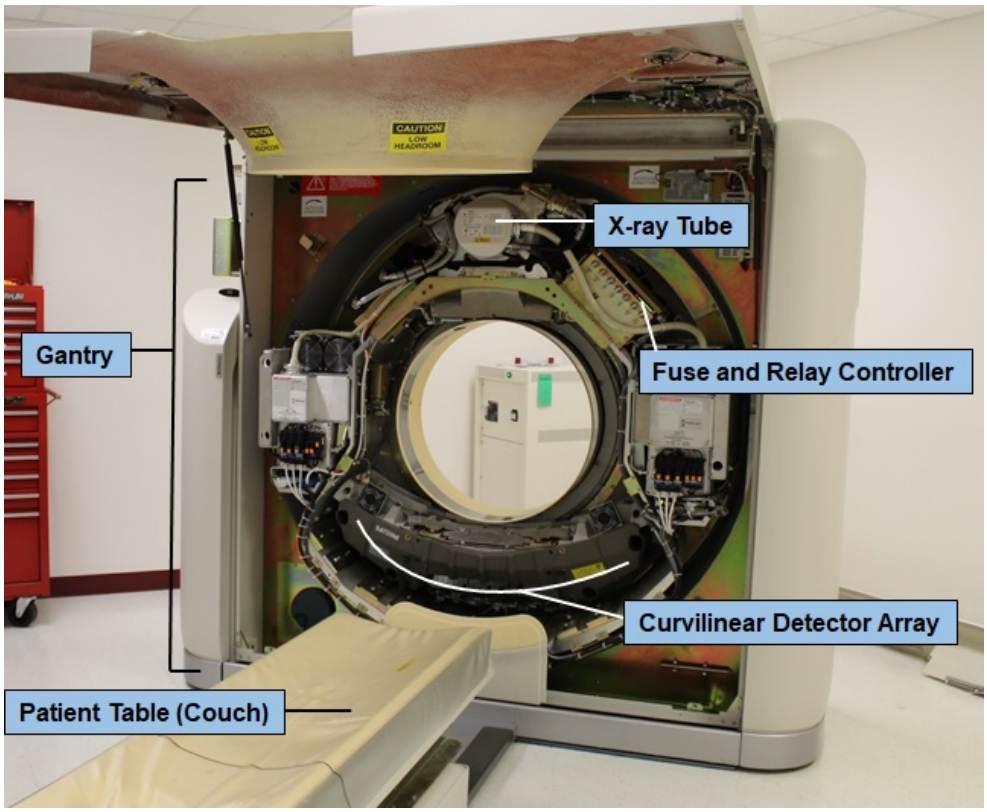

Figure 3–15. Third generation CT components.

Table

Each CT gantry comes with a patient table or couch styled according to the individual manufacturer's specifications. The purpose of the couch is to support and move the patient through the central opening of the gantry. Movement of the patient couch can be controlled by the computer or manually in a horizontal plane. Newer couches are fabricated from carbon fiber material low in Z (impedance), which are very thin, yet adequately strong. It should have a smooth and accurate motor drive so precise patient positioning is possible. When positioning is not exact, the same tissue can be scanned twice or missed altogether. It should be capable of autoindexing, so the operator does not need to enter the room between each scan. Such a feature reduces the examination time required for each patient.

X-ray tube

The early scanners used a stationary anode with a 2 mm x 16 mm focal spot operating at 120 kVp and 30 mA. This reduced image resolution significantly. However, since first-generation images were usually displayed on an 80 x 80 matrix, the X-ray tube was not the problem in this system. As matrix size increased to 512 x 512, rotating anode tubes with focal spots as small as 0.6 mm x 1.2 mm came into use. Small-focal-spot scanners use a pulsed beam to reduce the heat load. Modern pulsed scanner tubes operate at 120 kVp, 1–5-msec pulses, and up to 1,000 mA. Some units permit 80 and 140 kVp to be selected, sometimes in alternate pulses, for dual energy scanning in which comparisons can be made between images at different kVp values.

In addition, 0.5–5.0 million heat unit anodes made of layered alloys, and cylindrical anodes, as well as liquid-cooled and air-cooled tube housing designs have been developed. These tubes should be relatively heavy duty to tolerate the production of many images in a short period of time, as a CT tube may produce 30 exposures per examination. Because most CT units are scheduled for 10–20 examinations per day, a tube may accumulate 10,000 exposures in a single month. X-ray tube failure is a principal cause of CT scanner malfunction, and the principal limitation on sequential scanning frequency. It is not unusual for a CT tube to fail after several months.

Detectors

Another component housed in the gantry is the radiation detector system. CT detectors should have high capture efficiency, high absorption efficiency, and high conversion efficiency. These three parameters are called the detector dose efficiency. There are two categories of detectors: solid-state scintillation and xenon-gas ionization (fig. 3–16). Each has advantages and disadvantages.

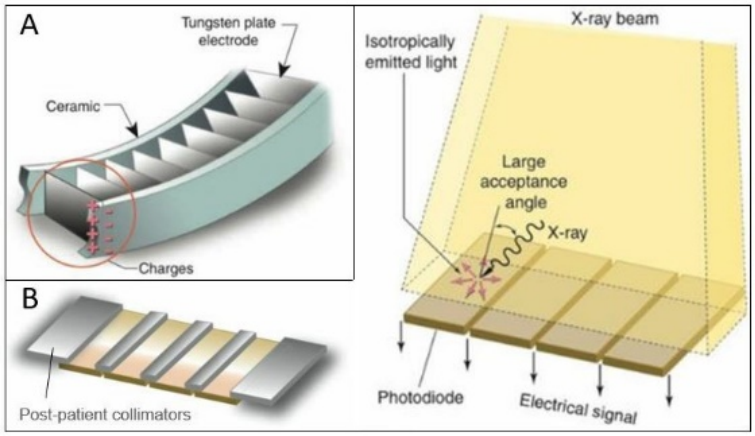

Figure 3–16. Two types of CT detectors. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Solid-state scintillation

A scintillation detector consists of a crystal connected to a solid-state detector assembly (fig. 3–16B). The crystal, usually sodium iodide, emits a flash of light when irradiated. Each flash of light is directly proportional to the amount of radiation reaching the detector. The solid-state component changes these light flashes into electrical impulses and transmits them to the computer. They have the advantage of offering a high current amplification. The spacing of these detectors varies from one design to another, but generally, 1–8 detectors per centimeter, or 1–5 detectors per degree are available. The concentration of scintillation detectors is an important characteristic of a CT scanner that affects the spatial resolution of the system.

The primary advantage of a solid-state detector assembly is that it is extremely efficient, converting 99 percent of the absorbed photons into light (output signal). This results in less image noise and fewer image artifacts. Unfortunately, it is not possible to pack the detectors so the space between them is small enough. The detector interspace may occupy 50 percent of the total area intercepting the beam. Consequently, the overall detection efficiency is only about 45 percent. Approximately 55 percent of the remnant X-rays exiting the patient contribute to patient dose without contributing to the image.

Xenon-gas ionization

Gas-filled detectors are also used in CT scanners. This type of detector is currently used in more CT units than any other detector. They are constructed of a large metallic chamber with baffles spaced at approximately 1-mm intervals (fig. 3–16A). The baffles are like grid strips and divide the large chamber into many small chambers. Each small chamber functions as an individual detector. The

entire detector array is hermetically sealed and filled under pressure with a high Z inert gas, such as xenon (Xe) or an Xe-krypton mixture. The remnant radiation impinges on the detector cell and ionizes the Xe. The positively charged ions move toward the negatively charged side of the detector; the negatively charged ions move to the positively charged side of the detector. This action creates a weak electrical signal directed through an amplification system. The output of the detector is proportional to the number of X-ray photons involved in the ionization. One problem with the gas ionization detector is its low absorption efficiency; many X-ray photons can pass through the detector cell without being ionized. Gas detectors range in efficiency from 60 to 90 percent, but exhibit no lag. Although gas-ion detector efficiency may be only 60 percent, these detectors can be packed extremely close in the detector array, permitting a detection efficiency of slightly less than 50 percent, similar to that of the scintillation crystal detector array. All other characteristics being equal, therefore, patient dose is about the same for both types of detector arrays. The choice of detector depends on the manufacturer and equipment design specifications, although gas-ionization detectors are much less expensive, making them quite popular to manufacturers. The information produced by the detector system is transferred to the computer group for processing.

Data Acquisition System

The DAS converts the electrical impulses from the detectors into digital information (numbers). These numbers represent the strength of the electrical impulses and are actual measurements of the structure's density through which the beam passed. Once the DAS converts the electrical signals into digital information, the data is sent to the CT computer system.

Computer group

The CT image system uses a unique computer, capable of performing in the neighborhood of 250,000 mathematical equations simultaneously to reconstruct cross-sectional images of the body part being imaged. This is based off remnant radiation measurements received by the detector array in the gantry. In simple terms, the basic function of a CT computer is to receive and analyze information from the DAS and convert it into video form so that an image can be displayed on a monitor. A key component in all CT computer systems is the array processor.

Reconstruction often requires up to 30 seconds. However, when an array processor is used, reconstruction time can be decreased to less than 1 second, which is desirable for dynamic studies. The array processor consists of dedicated circuitry capable of performing high-speed mathematical calculations. As the computer receives the digital information from the DAS, the computer processes the data to reconstruct cross-sectional anatomy images. Once you select the CT scanning parameters, the computer tells the DAS how to interpret your commands (scanning parameters); tells the patient couch to move; applies power to the X-ray tube to generate the production of X-rays; controls the detector array; processes and transfers the electronic data signals; and in general, oversees the performance of the whole CT system while providing feedback to the user.

The computer group can also contain peripheral system components, including the interface connection between the imaging group and the computer group, a laser printer, the storage system, or a PACS.

Image reconstruction

Image reconstruction is the actual operation the computer performs when mathematical equations turn the raw data (matrix of whole numbers) into a cross-sectional image. The array processor performs the mathematical calculations; therefore, it frees up the host computer for other operations. Most CT computers and array processors work so fast that scans are acquired in less than a second, and then the image reconstruction phase only takes a few more additional seconds to complete.

Multiplanar reformations are the most common post-processing function you will likely perform in CT. Using the system software and multiplanar reformation application, axial plane images can be reformatted into coronal, sagittal, or oblique planes. Another function of this type of application

includes sending reformatted images digitally to a laser printer, or more commonly, saving an additional series of images within the study itself and sending them to the PACS.

The computer room

The computer system being referred to here is not your normal two-foot tall personal computer tower. The CT computer system is typically one or two large floor-mounted cabinets (towers). The computer system, along with the high-voltage generator and system transformer(s) are housed in an enclosed, climate-controlled room, due to the high amount of heat generated by these machines. The room, commonly referred to as the computer room, is kept colder than patient care or technologist areas. Heat is the enemy of a central processor unit (CPU) within computers and when CPUs get hot, they slow down. The colder air temperature serves to cool the CPUs because computers run more efficiently in cool versus hot environments. The recommended temperature for a dedicated computer room is around 65–68° Fahrenheit.

Control group

The control group contains the display control console and keyboard. Many scanners have two consoles—one for the technician to operate the unit, and one for the physician to view the scanned image and change its contrast, size, and general visual appearance. The operator's console has meters and controls to select proper radiographic technique factors, to the gantry and patient couch, and to make computer commands for image reconstruction and transfer. The physician's viewing console retrieves data from the computer and allows the display of an image for viewing, manipulation, and diagnosis.

Operator's console

Using the console, the operator enters the patient information, selects scanning parameters (including kVp, mA, time, slice thickness, scanning mode, reconstruction algorithm, etc.), initiates the CT scan, monitors the scanner performance, and views the images on the computer monitors (fig. 3–17). The kVp controls are usually set above 100 kVp. The mA station used depends upon the type of beam employed. That is, a low mA station (20–50 mA) should be used if the X-ray beam is continuous. Several hundred mA may be used if it is a pulsed beam. A typical technique for a continuous beam may be 120 kVp at 50 mA. Scan time (length of time required per scan slice) is often selectable and falls within the range of 1–10 seconds with the faster units.

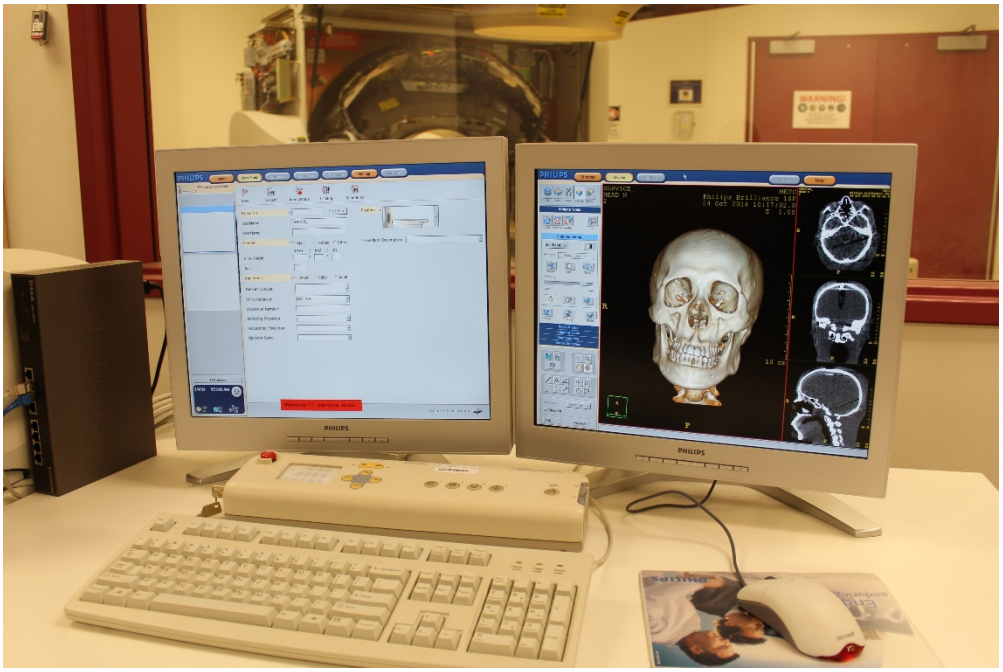

Figure 3–17. CT operator's console.

The operator can also adjust the thickness of the slice. Most units offer a slice thickness of 1.5 mm, 5 mm, or 10 mm (1 cm). Selecting a slice thickness from the console produces an automatic collimator adjustment at the X-ray tube. The thinner cuts require an increase in radiographic technique because collimation reduces exposure to the patient, decreasing detectable remnant radiation. The operator's console has one or two monitors that allow the computer to reconstruct electronic data into a video image. One monitor allows an operator to show patient data on the scan (hospital identification, name, Social Security number, age, sex, etc.) and to identify each scan (scan number, technique, couch position, etc.). The second video screen might be used to view an image before it is transferred to a permanent image format.

Physician's viewing console

This console is essential in a department that has a high workload and fully utilizes the CT system. It allows a physician to review and manipulate any previous image for maximum information without interfering with scanner operations. For maximum effectiveness, an independent computer should support the physician's viewing console. If it requires the main computer for image manipulation, viewing can be slowed considerably when attempted during scanning, since the scan mode has precedence. This console allows the physician to call any previous image and manipulate that image for maximum information. The manipulative controls provide for contrast and brightness adjustments, magnification techniques, region-of-interest viewing, and use of on-line computer software packages. This software may include computer programs to generate CT number histograms along any preselected axis; computation of mean and standard deviation of CT values within a region of interest; subtraction techniques; and planar and volumetric analysis.

Image storage

Image data is stored temporarily on the host computer's hard drive to facilitate rapid image recall for viewing and manipulating. However, the hard drive can only hold so many images. For this reason, electronic data is transferred periodically to a more permanent digital storage medium, such as a compact disc (CD), magnetic tape, or optical disk. A CD can be used to store data from a single patient; whereas, magnetic tapes and optical disks can store examination data from many patients. Optical disks tend to be the storage medium of choice in most departments, since their capacity to store images/data is dramatically more than that of magnetic tapes. No matter the image storage medium, it is labeled and stored relatively close to the CT scanner, so historical images can be retrieved for additional manipulation.

623. Cone beam computed tomography

When you think of CT scanners, you usually think of the large machines taking up an entire room and scanning patients from head to toe as they lie on a table. You may also think 4-slice, 16-slice, or 64-slice. Now, however, CT is much more than that. CBCT is a variation on traditional CT and is mainly used as a dental CT modality for dental and maxillofacial imaging. The datasets received from these systems are used to reconstruct 3D images predominantly in the following specialties:

- Dental—visualization of teeth and diagnosis of dental cavities.
- Oral—maxillofacial region (mouth, jaw, neck), dental implant planning, endodontic (root canal) diagnosis, and cleft palate assessment.
- Ears, nose, and throat assessment.

This lesson breaks down the differences in technology and applications between the traditional CT scanners and conventional dental X-rays systems as compared to the more specialized CBCT.

CBCT principles

The principle behind this technique, as its name implies, is a tightly collimated narrow cone-shaped X-ray beam. It uses a rotating gantry, with the X-ray source and flat panel detector rotating around a point (or field) of interest of the patient (fig. 3–18). The conical shape of the beam distinguishes this technique from helical CT, which used a fan-shaped beam. As a result of the acquisition of two-

dimensional (2D) projections throughout this rotation, only one rotation or less is needed to acquire a full 3D dataset. The acquisition time of CBCT devices ranges from between 6 and 20 s. The images received by the detector are then compiled by the computer into volumetric data (primary reconstruction). These images appear as 2D multi-planar reformatted slices, and become 3D images by using surface reconstruction or volume rendering.

By rotating the beam around a fixed point in the object of interest and acquiring projections from many different angles, a typically cylindrical 3D volume can be reconstructed. Typically, the unit collects a few hundred projections. Although units generally use a 360° rotation, some devices have implemented a 180° or slightly greater rotation arc, which suffices for image reconstruction and leads to significant radiation reduction. The 2D attenuation profiles obtained from all angles are then reconstructed into a 3D matrix, containing voxels which all have a certain grey value, representing the average density within each specific voxel. The grey value for each voxel is determined by the reconstruction algorithm, by combining the information from all obtained projections.

Figure 3–18. Typical CBCT unit.

CBCTs can vary by the types of exposures that they provide; pulsed and continuous exposure. Some X-ray tubes use a pulsed exposure to ensure that there is no exposure being made between projections, resulting in a large discrepancy between scan time (the time between the first and last projection) and exposure time (the cumulative time during which the unit is making an exposure). For example, the total scan time may be 20 s, but each pulse may be just 10 ms (giving a total exposure time of 2 s for a scan with 200 projections). Other X-ray tubes allow only continuous exposure, for which the total scan time and exposure time are equivalent. The dose is proportional to the product of the exposure time and tube current (mAs). Both pulse and continuous exposure approaches are susceptible to effects of detector lag, but pulsed X-ray systems may exhibit improved spatial resolution due to reduced motion effect, that is, motion of the gantry during each exposure/read-out frame.

For most CBCT systems, the kVp is fixed, and the tube current (mA) and exposure time (s) can be varied depending on the desired image quality and patient size. In current dental CBCT practice, this is typically performed either manually or through the selection of pre-set exposure protocols. Using AEC, which is commonly applied in medical CT, the exposure is varied before or during the scan automatically by a feedback circuit depending on patient size and attenuation, assuring no under- or overexposure takes place. One simple implementation in dental CBCT determines the mAs based on a 2D scout image. A lower detector signal for the scout image results in a higher mAs for the actual CBCT scan, and vice versa.

Most CBCT systems have multiple pre-defined field of view (FOV) sizes, so a collimator will have several pre-defined openings according to the FOV sizes. However, a few CBCT machines do have

freely adjustable collimation along the *z* axis, allowing for FOVs of any height. While early model CBCTs used image intensifiers for image acquisition, current versions utilize various types of FPDs. Most CBCT systems use indirect FPDs where a layer of scintillator material, either gadolinium oxysulfide or cesium iodide, converts the X-ray photons to light photons. The unit then converts the light photons into electrical signals.

Image reconstruction

In general, image reconstruction can be grouped into two categories: filtered back projection (FBP) such as the Feldkamp algorithm and algebraic reconstruction techniques (ART) such as iterative reconstruction. The most commonly used algorithm is the modified Feldkamp algorithm. FBP is basically the inverse or back projection process for weighted and filtered projections, in which the value for each pixel in the projection image is assigned to every voxel along the path of the X-ray. When performed for every projection, an image of the scanned object is reconstructed. The filter consists of two parts: (1) a ramp filter to correct for blur that is intrinsic to the projection/back projection process and (2) a smoothing filter to reduce high-frequency noise that is amplified by the ramp filter.

Recently, iterative reconstruction methods (ARTs) have gained attention as an alternative reconstruction method. Using iterative reconstruction, the acquisition process is modeled and consecutive cycles of reconstruction and reprojection are performed. After an initial reconstruction, projections of the reconstructed volume are simulated and compared with the actual raw data (i.e. the acquired projections). The reconstruction is then repeated and altered based on the difference between actual and simulated projections. This process is repeated for a number of iterations. One benefit of this technique is that every part of the acquisition process can be modelled, aiding in artefact reduction. However, iterative reconstruction requires a great amount of computing power; fortunately, this practice is generally expected to increase as workstations continue to increase their processing speeds.

CBCT versus MDCT

While both standard MDCT and CBCT scanners fall into the computed tomography category, there are some big technical differences between them. Traditional CT uses a high-output, rotating anode X-ray tube. Cone beam tomography utilizes a low-power, medical fluoroscopy tube that generally provides continuous imaging throughout the scan. Traditional CT records data with a fan-shaped X-ray beam onto image detectors arranged in an arc around the patient, producing a single slice image per scan. Each slice must overlap slightly in order to properly reconstruct the images. Unlike traditional CT scanners, in CBCT, an X-ray tube and detector panel rotate around the patient capturing data with a cone-shaped X-ray beam instead of the slices. The cone-shaped X-ray beam transmits onto a solid-state area sensor for image capture, producing the complete volume image in a single rotation. The single-turn motion image capture used in CBCT is quicker than the traditional spiral motion, and can be accomplished at a lower radiation dose as a result of no overlapping of slices. This type of imaging exposes a patient to less radiation than traditional CT scanners.

Technical limitations hinder its use over standard CT for general diagnostic use. The two most significant factors that affect successful integration are image quality and time (for set up, image acquisition, and image reconstruction). Compared to MDCT, the wider collimation in CBCT leads to increased scatter radiation and degradation of image quality, as demonstrated by more artifacts, decreased contrast-to-noise ratio, and the lack of appropriate bone density determination. The temporal resolution of cesium iodide detectors in CBCT slows data acquisition time to approximately 6 to 20 seconds, which increases motion artifacts. The time required for image reconstruction also takes longer for CBCT (1 minute) compared to MDCT (real time) due to the computationally demanding cone beam reconstruction algorithms. CBCT however has an advantage over standard CT in some applications, because of its lower cost, smaller size, and smaller radiation dose. Total radiation doses from 3D dental CBCT exams are 96% lower than conventional CT exams, but deliver more radiation than the standard dental 2D panoramic X-ray.

CBCT versus conventional dental X-ray

Similar to dental 2D X-ray techniques, CBCT is used for dental applications such as implant treatment planning, endodontics, maxillofacial surgery, and orthodontics. The basic principle of the X-ray tube is the same for each radiographic modality using X-rays. Differences between tubes used for 2D radiography and CT or CBCT scanning are mainly found in the size of the exit window (collimation), the range of exposure factors, and the amount of beam filtration. The images from CBCT provide more information than conventional dental X-rays, allowing for more precise treatment planning. A single scan provides a wide variety of views and angles that the technician can manipulate to provide a more complete evaluation. A major advantage of CBCT is its ability to image both bone as well as soft tissues at the same time. CBCT has the advantage over panoramic studies, due to its high spatial resolution, shorter scan times, and rapid image acquisition. It is however, much more expensive than most conventional dental X-ray methods.

Some systems allow CT scan reconstructions that are comparable to cephalometric projections. The advances in 3D imaging improve the ability to identify anatomical cephalometric landmarks that are not easily detectable in current dental images, thereby increasing the accuracy and reliability of orthodontic diagnosis and treatment planning. Likewise, the lower mandibular border is easier to identify. However, 3D images are not as reliable for identifying the long axes of the upper and lower incisors.

Self-Test Questions

After you complete these questions, you may check your answers at the end of the unit.

620. Tomography

1. Match each definition in column A with the appropriate term in column B. Items in column B may be used once or not at all.

Column A	*Column B*
____ (1) The point about which the lever pivots.	a. Lever.
____ (2) The angle, in degrees, of the tube travel during exposure.	b. Fulcrum.
____ (3) The connecting rod that couples the tube and film carrier so the tube and film movement are proportional.	c. Focal plane.
____ (4) The distance the tube travels during exposure expressed in inches.	d. Blurring.
____ (5) Tomography with a small exposure angle.	e. Amplitude.
____ (6) The layer of the body that appears sharpest on a tomogram.	f. Rate.
____ (7) The unsharp body area outside the focal plane.	g. Exposure angle.
____ (8) The speed of tube travel, usually expressed in inches per second.	h. Zonography.

2. What is the simplest and most common tube movement?

3. What determines section thickness during the tomogram?

4. What are the four factors that affect exposure angle?

5. Explain the process of digital tomosynthesis.

621. Computed tomography principles

1. What makes a CT image so useful?

2. List at least four common uses of CT scans.

3. What is the purpose of using CT scans for cancer detection?

4. What is the most widely used IV contrast agent used during CT scans?

5. Briefly define CT.

6. What is the minimum difference in tissue density that separates structures in conventional radiography as opposed to CT?

7. Match the generation of scanner in column B with the scanning motion in column A. Answers in column B can be once or not at all.

Column A	*Column B*
____ (1) Translate-rotate with pencil beam at 1° increments with an angle of 180°.	a. First generation.
____ (2) Rotate-stationary with wide continuous beam and 360° rotation.	b. Ultrafast.
____ (3) Translate-rotate at 5° increments with pencil beams for a 180° rotation.	c. Second generation.
____ (4) Rotate-rotate with narrow pulsed fan beam.	d. Third generation.
	e. Fourth generation.

8. Why was the spiral technique developed for CT scanning?

9. What is the key advantage of MDCT over single-slice helical CT units?

10. What is an isotropic voxel?

11. Define pitch.

12. What type of CT technology allows for real-time CT imaging?

13. Explain how dual-source CT works.

14. What is a matrix?

15. What happens as the CT image matrix becomes larger?

16. What does a pixel on a CT image represent?

17. What is the smallest imaging area of a patient when given a 1024 x 1024 monitor and a field size of 48cm on a CT scanner?

18. What is a ray sum?

19. Give the CT number that corresponds to the following tissues: water, air, and dense bone.

20. What is the term window-width and how does it affect the image?

622. Computed tomography system components

1. What components are housed in the gantry?

2. What is the purpose of the patient table or couch?

3. What can happen if positioning is not exact with the patient table?

4. What is the principal cause of CT scanner malfunctions and the principal limitation on sequential scanning frequency?

5. What are the two types of detectors used in CT systems today?

6. How many detectors can be placed per centimeter and per degree?

7. What is the *overall* detection efficiency for scintillation detectors and what is it due to?

8. What is the *overall* detection efficiency for gas ionization detectors and what is it due to?

9. What is the basic function of the CT computer?

10. What is the recommended temperature range for the computer room?

11. Which console controls scanning parameters?

12. Which has precedence with the CT system—viewing or scanning an image?

623. Cone beam computed tomography

1. How is the CBCT beam different from a helical CT beam?

2. How many rotations are required to construct a full 3D image?

3. List the two types of exposures that CBCT systems provide.

4. How does the unit compensate for a low detector signal when using a 2D scout image to determine mA?

5. List the two general categories of image reconstruction.

6. What are significant factors that hinder the use of CBCT over standard CT?

7. How much lower is the total radiation dose of 3D dental CBCT compared to conventional CT?

8. What are the differences between X-ray tubes used for 2D radiography and CT or CBCT scanning?

3–2. Nuclear Medicine

The Nuclear medicine department is usually an offshoot of the X-ray department, and presents a twist to your current X-ray thinking. This section is broken down into three lessons. First, we'll cover the clinical applications of nuclear medicine; second, we'll familiarize ourselves with the basic principles behind it; and lastly, we'll discuss the various types of equipment.

624. Clinical applications

Nuclear medicine studies require the oral or intravenous introduction of very low-level radioactive chemicals (called radionuclides, radiopharmaceuticals, or radiotracers) into the body. These chemicals are specially formulated to temporarily collect in a specific part of the body to be studied. The organs in the body take up the radionuclides, and then emit faint γ ray signals, which are measured by a γ camera. The γ camera has a large crystal detector (scintillation crystal). These crystals detect the emitted radiation signal and convert that signal into faint light. Like other digital modalities, this light converts to an electric signal, which is then digitized and reconstructed into an image by a computer. The nuclear medicine image can be in gray scale, such as in a bone scan, or it can be color coded to clearly show functional activity as in a cardiac study.

As you can see, the main difference between nuclear medicine and a conventional X-ray is that during an X-ray, the radiation comes out of the X-ray tube, and then passes through the patient's body before detection and recording. Nuclear medicine uses the opposite approach: a radioactive material is introduced into the patient, and is then detected by the system. The radiation, which is emitted by the body during nuclear medicine imaging, is γ. These γ rays are similar to X-rays, but have a shorter wavelength. If you remember the difference between X-rays and γ radiation from unit 1, they are nearly identical, except for their origin and method of production. X-rays are man-made and originate in an X-ray tube, while γ radiation emits from radioactive nuclei.

The radionuclide substances used in nuclear medicine imaging are usually synthesized radioactive substances (e.g., technetium) or radioactive forms of elements naturally found in the body (e.g., iodine). The levels of radiation involved in nuclear medicine studies are usually considerably lower than a patient would receive in a conventional X-ray study or a CT scan.

Nuclear medicine is mainly used to allow visualization of organs or areas within organs that cannot be seen on conventional X-ray images. Injuries or abnormalities within an organ, especially tumors, may stand out on nuclear medicine images. Usually, these abnormalities are seen as areas of reduced radioactivity (called a "cold spot"); however, in some instances (e.g., bone scanning), areas of increased activity (called a "hot spot") represent disease or injury. Let's look at some particular applications of nuclear medicine.

Bone scan

Performing a bone scan with nuclear medicine can be an important step in diagnosing and assessing treatment for various kinds of cancer, including breast cancer, because it can be revealed if the cancer has spread beyond its primary site and developed secondary cancer growths in the bone. Also, with an X-ray, a doctor can only see if a bone is broken or unbroken; a bone scan allows physicians to see metabolic changes caused by fine fractures, small tumors, or degenerative diseases, such as arthritis.

Heart disease

Nuclear medicine is also an important component in the diagnosis and treatment of heart disease. For example, cardiac angiography yields excellent images of the beating heart and blood vessels (coronary arteries) that supply the heart muscle with blood. However, a stress thallium nuclear medicine study provides additional information by showing the function of the myocardium. In a stress thallium study, two data acquisitions are performed. First, the patient is strenuously exercised on a treadmill or stationary bicycle to elevate cardiovascular activity and "stress" the heart. This is followed immediately by a nuclear medicine examination. The patient is then given a period of time to rest. When the patient's heart activity has again become normal, a second nuclear medicine study is completed. The physician can then compare the images and function of the heart at rest versus the heart under stress. We refer to this as a nuclear stress test. Areas of the heart that may have been previously damaged by myocardial infarction (heart attack) or may have insufficient blood supply due to a blockage of a coronary artery will not show the proper function in the stress image. Another common cardiac application of nuclear medicine is an electrocardiographic multiple gated acquisition (MUGA) scan, which allows for the study of the heart's muscular wall motion and chambers.

Other procedures

Some other areas of the body that can be imaged using nuclear medicine to clarify their image or show their function include:

- The abdomen—for gastrointestinal (GI) bleeding and other such disturbances.
- The brain—for tumors, aneurysms, or to evaluate stroke.
- The breast—to image breast cancers.
- The heart—for coronary artery disease, valve disease, or heart attack.
- The kidneys—for renal function or tumors.
- The liver/spleen—for cirrhosis or cancer.
- The lungs—for pulmonary embolism (blood clot) or to test for smoke inhalation injury in burn patients.
- The skeletal system—for cancer or to test for hidden bone trauma in sports injuries.
- The stomach—for stomach function and to confirm ulcers or cancer.

It is important to note that nuclear medicine images are produced over time; each picture may take anywhere from one to thirty minutes, depending on the study being performed. To enhance the image quality, the camera must be as close to the patient as possible—the closer the camera, the better the picture. Because nuclear medicine images are acquired over time, the patient must remain as still as possible during the tests. Motion causes blurring of the pictures and often means the picture must be restarted.

625. Principles

Now that you know what nuclear medicine is used for, you now need to become more familiar with the principles behind it, in case you're called to work on equipment within the department. In this lesson, we will cover some terms you will likely hear. We will also discuss radiopharmaceuticals used in some of the procedures.

Terms

Before we go any further with our study of nuclear medicine, let's go over some common terms you will hear in nuclear medicine. Planar describes a 2D view of the process or function of the organ being imaged (7). Single photon emission computed tomography (SPECT) provides a 3D computer-reconstructed image of multiple views and functions of the organ being imaged. Positron emission tomography (PET) produces a high-energy, 3D computer-reconstructed image measuring and determining the function or physiology in a specific organ, tumor, or other metabolically active site. As you should remember, tomography is a method of separating interference from the area of interest by imaging a cut section of the object. Radiopharmaceutical **(**may also be referred to as tracer or radionuclide) is the basic radioactively tagged compound necessary to produce a nuclear medicine image. A γ camera is the basic instrument used to produce a nuclear medicine image. *In vitro* procedures are done in test tubes. Radioimmunoassay (RIA) is a special type of *in vitro* procedure that combines the use of radiochemical and antibodies to measure the levels of hormones, vitamins, and drugs in a patient's blood. *In vivo* procedures are when trace amounts of radiopharmaceuticals are given directly to a patient. The majority of nuclear medicine procedures are *in vivo*.

Radiopharmaceuticals

One of the most important parts of nuclear medicine is the radiopharmaceutical used in the processes. The chemical structure of a radionuclide determines how it is distributed within the body and where it accumulates. Of course, the particular organ that needs to be imaged determines which radionuclide is selected. Once the radionuclide is introduced into the body and it accumulates in the suspect organ, it emanates radiation, usually in the form of γ rays, due to the radioactive decay of the radionuclide. This radiation is then detected using a γ camera.

Radiopharmaceutical decay

An unstable nuclide, termed a radionuclide, will try to become stable by undergoing radioactive decay. There are several ways in which a radionuclide can decay: including α particle decay, β particle decay, γ ray emission, and electron capture. Radionuclides that emit γ rays are the most useful because the rays can pass through tissue and reach the γ detector outside the body. Compounds that produce α particles are not useful because they can only penetrate a few millimeters of tissue and, therefore, cannot be detected. Like the previous compound, those that emanate β particles are also not useful for detection. However, compounds that emanate α and β particles can be used as radiotherapeutic agents if they can target tumors or other destructive areas. The final type of radioactive decay is electron capture. When this reaction takes place, an orbital electron from the K or L shell is captured by the nucleus, which produces a gap in the orbital shell. Electrons from outer shells fill the gap in a cascade process and this produces X-rays.

Radionuclide half-life

The half-life of a material is very important because it tells the physician when the material has localized, when it will be emitting the strongest, and when it will leave the body. This information is required for a good imaging product.

Radionuclides decay in an exponential fashion, and the term half-life is often used to characterize decay. Half-life usually refers to the physical half-life, which is the amount of time necessary for a radionuclide to be reduced to half of its existing activity. Activity refers to the amount of radioactivity present in number of disintegrations per second. In addition to the physical half-life or physical decay of a radionuclide, two other half-life terms are used—biologic and effective half-life.

Biologic half-life

The biologic half-life refers to the time it takes an organism to eliminate half an administered compound or chemical on a strictly biologic basis. If a stable chemical compound is given to an individual and half is eliminated by the body, perhaps in the urine within 3 hours, the biologic half-life is 3 hours.

Effective half-life

The effective half-life incorporates the physical and biologic half-lives. When speaking of the effective half-life of a particular radiopharmaceutical in humans, you need to know the physical half-life of the radioisotope used as a tag or label, as well as the biologic half-life of the tagged compound. If these are known, then the equation in figure 3–19 can be used to calculate the effective half-life. If the biologic half-life is 3 hours and the physical half-life is 6 hours, then the effective half-life is 2 hours. Note the effective half-life is always shorter than either the physical or biologic half-life.

$$T_e = \frac{T_p \times T_b}{T_p + T_b}$$

T_e = Effective half-life
T_p = Physical half-life
T_b = Biologic half-life

Figure 3–19. Effective half-life.

Many things must be taken into consideration when selecting a radionuclide for use within the nuclear medicine laboratory. Concerning the decay characteristics of a radionuclide, it should have a physical half-life greater than the time required to prepare the material for injection, and an effective half-life longer than the examination time.

Radionuclide production

There are four basic methods used to produce radiopharmaceuticals: neutron capture, nuclear fission, using radionuclide generators, and charged-particle bombardment. The first two methods use a small nuclear reactor specially designed for radionuclide production. The third method, use of radionuclide generators, is the most convenient because the generator is usually on-site. The generator continuously produces the radionuclide and is usually "milked" on a daily basis.

Let's focus on charged-particle bombardment. This method uses a piece of equipment called a cyclotron. A cyclotron works by ionizing hydrogen or deuterium gas using a radio-frequency (RF) field in the center of the cyclotron. The resulting negatively charged ions are then accelerated in a spiral trajectory toward the outside of the round cyclotron. Next, there are superconducting magnets at the top and bottom of the cyclotron, which are used to contain the ion beam. At the outer edge of their trajectory, the ions exit the cyclotron and are magnetically steered through a thin copper foil strip. This foil strip removes the electrons from the ions, thus leaving protons or deuterium nuclei to collide with a target, which is stationed within a thick block of borated polyethylene. The block absorbs the scattered neutrons produced during bombardment. The proton or deuteron nuclei collide with a particular target to produce the desired radioactive isotopes. As you can imagine just from the description, radioisotopes produced by this method are quite costly.

Types of radiopharmaceuticals

Currently, more than 2,700 radioisotopes have been artificially produced in reactors, cyclotrons, linear accelerators, and generators. Radionuclides used in nuclear medicine are mostly artificial, produced in cyclotrons, reactors, or generators. As you can see, there are many types of radionuclides used in nuclear medicine; therefore, this is a very broad and technical subject. Fortunately, you do not need to be an expert on the subject. We will, however, briefly touch on some of the more common radionuclides used in nuclear medicine today.

Technetium 99m

The Technetium 99m (99m-Tc) fulfills many of the criteria of an ideal radionuclide, and is used in over 90 percent of nuclear imaging procedures in the United States. It has no particulate emission, a 6-hour half-life, and a predominant 140 keV photon with only a small amount of internal conversion.

Iodine 123 and 131

Iodine 123 (123-I) and Iodine 131 (131-I) are clinically useful for imaging and may be administered as iodide. Iodine 123-I has a 13.3-hour half-life and decays by electron capture to tellurium 123. Iodine 123-I is produced in a cyclotron by bombardment; therefore, it is expensive and nationwide distribution is difficult.

Iodine 131-I is a much less satisfactory isotope from an imaging viewpoint because of the high radiation dose to the thyroid and its relatively high photon energy. However, it is widely available, relatively inexpensive, and has a long shelf life. It has a half-life of 8 days, and decays by a combination of beta and gamma emission. When iodine is administered orally, it is readily absorbed from the GI tract and distributed in the extracellular fluid. It concentrates in the salivary glands, thyroid, and gastric mucosa. Iodine is an extremely useful radionuclide because it is chemically reactive and produces a variety of radiopharmaceuticals.

Xenon

Xe is a relatively insoluble and inert gas, and is most commonly used for pulmonary ventilation studies. Xe is highly soluble in fat and oil, and there is some absorption of Xe into plastic syringes. Xe 133 has a physical half-life of 5.3 days. With normal pulmonary function, the biologic half-life is approximately 30 seconds.

Gallium

Two isotopes of gallium have been useful as imaging agents: gallium 67 and gallium 68. Gallium 67 has a half-life of 78 hours. It can be produced by a variety of reactions in a cyclotron. Gallium 68 can be generator produced in a germanium 68/gallium 68 generator. Gallium 68 has a 68-minute half-life and is a positron emitter.

Gallium is usually administered as a citrate and has a 12-hour half-life in the blood pool, where it is largely bound to plasma proteins, especially transferrin, lactoferrin, and ferritin. It ultimately leaves the blood pool to localize primarily in the liver, spleen, bone marrow, and skeleton. Tumor localization is not well understood, but lysosome binding appears to be involved, as well as the previous-mentioned protein products with the tumor. The biologic half-life of gallium is approximately 2 to 3 weeks.

Indium

It is a metal that can be used as an iron analog and is quite similar to gallium. Isotopes of interest are Indium 111 and 113m. Indium 111 has a physical half-life of 67 hours and is cyclotron produced. It has a biological half-life of 1.7 hours.

Thallium

Thallium 201 has a physical half-life of 73.1 hours and decays by electron capture to mercury 201. Since thallium 201 is cyclotron produced, it is quite expensive. It is normally administered as a chloride and rapidly clears the blood with a half-life of 30 seconds—3 minutes. Because it is roughly a potassium analog, it is rapidly distributed throughout the body, particularly in muscle.

Fluorine

Fluorine-18 (F-18) is one of the most commonly used isotopes in PET imaging. F-18 is a positron emitter and has a half-life of 110 minutes. It is manufactured in a cyclotron, which owing to the short half-life of F-18 must be geographically close to the imaging center. The most common use for F-18 is synthesis of F-18 fluorodeoxyglucose (FDG), used to demonstrate glucose metabolism in the heart, brain, and oncologic processes.

626. Nuclear medicine equipment

Now, let's talk about the equipment used to record and produce the image radiating from the inside out. This lesson will cover the gamma camera and various specialty and hybrid equipment, as well as some forms of measurement equipment.

Gamma scintillation camera

The most widely utilized imaging device in the nuclear medicine laboratory is the γ scintillation camera. A γ camera converts photons emitted by the radionuclide inside the patient into a light pulse and a subsequent voltage signal. This signal forms an image of the distribution of the radionuclide.

The basic components of the γ camera system (fig. 3–20) are the collimator, scintillation crystal, array of photomultiplier tubes, pulse height analyzer (PHA), imaging monitor, and control console.

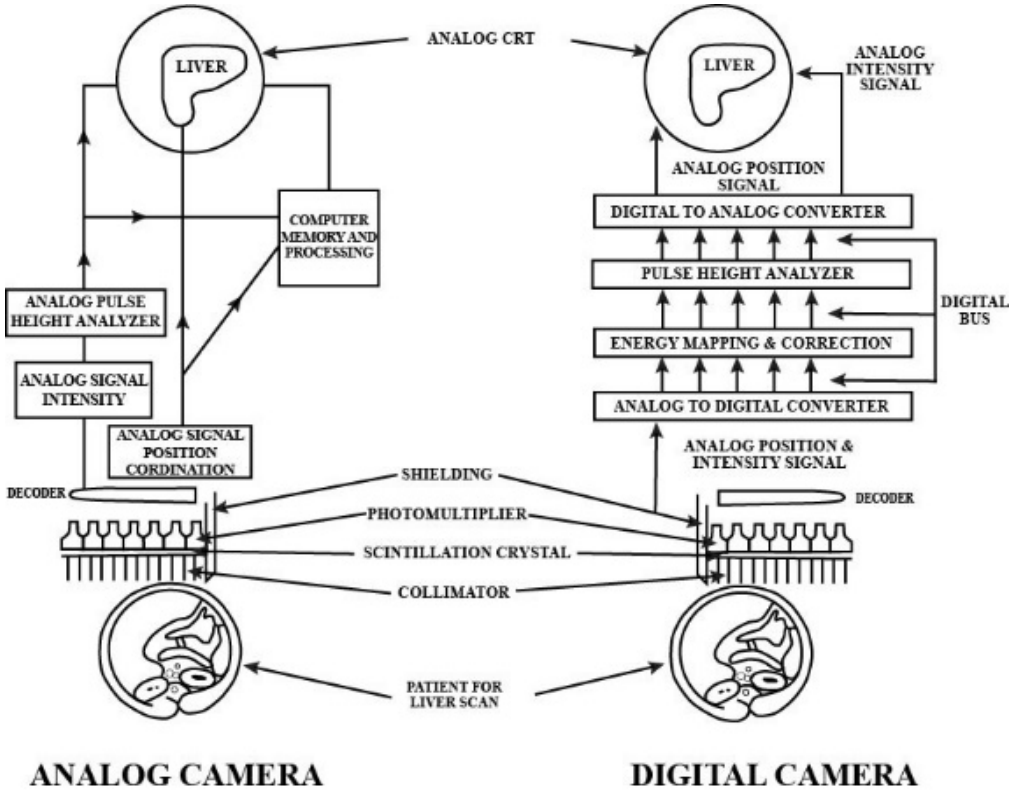

Figure 3–20. Gamma camera system.

Collimator

The collimator is made of perforated lead and is positioned between the patient and scintillation crystal, to reduce the number of gamma photons that are incident on the crystal. It is designed to reduce scatter and; thus, allows the γ camera to localize the radionuclide in the patient. Collimators are able to perform this function by absorbing and stopping most radiation. However, the one exception involves radiation that arrives almost perpendicular to the detector face; note that radiation striking the collimator at oblique angles is not included in the final image. Several different collimator designs are available for various special applications: converging, diverging, fan beam, slant hole, and pin hole.

We will just discuss the parallel-hole collimator, as it is the most widely used in nuclear medicine laboratories. It consists of parallel-holes with a long axis perpendicular to the plane of the scintillation crystal. It is a lead plate, approximately 2–3 cm thick, that covers the entire surface of the crystal. The lead is pierced by thousands of parallel-oriented holes, each less than 1 mm in diameter. Each hole must be at least 10 times as long as it is wide. The lead walls between the holes are referred to as septa. The septa absorb γ rays that do not emanate from the direction of interest; therefore, a collimator for use with high-energy γ rays has much thicker septa than a collimator for low-energy rays. The septa are generally designed so septal penetration by unwanted γ rays does not exceed 10–25 percent.

In general, the longer the septa, the better the resolution and the lower the count rate (sensitivity) for a given amount of radionuclide. The count rate is inversely proportional to the square of the collimator hole length. If the length of the septa is decreased, the count rate increases and resolution decreases. With a parallel-hole collimator, neither the size of the image nor the count rate changes significantly with the distance of the object of interest from the collimator. This is because, as the object is moved a small distance away from the crystal, the inverse square law reduces the number of counts. However, this is compensated for by increasing the viewing area of the collimator. On the other hand, the resolution is the best when the object of interest is as close to the collimator face as possible; scans with multi-hole collimators are usually obtained with the collimator in contact with the patient.

Collimator transmission is typically in the order of 0.1 percent of incident photons. The collimator defines a line of sight. The tubular holes through the parallel-hole collimator only allow through those photons travelling close to the axis of the hole, perpendicular to the crystal face. Photons traveling at oblique angles are absorbed by the lead septa forming the walls of the collimator holes. Collimators can be categorized according to sensitivity or spatial resolution: high sensitivity, all-purpose, and high resolution. An appropriate collimator must be selected according to the energies of the γ photons of the radionuclide in use. Low-energy collimators are suitable for use with 0- to 200-keV units, medium-energy collimators for 200- to 400-keV units, and high-energy collimators for 400- to 600-keV units.

Scintillation crystal

The scintillation crystal is a single crystal sheet. For most nuclear medicine applications, the crystal is thallium-activated sodium iodide NaI(Tl). Choice of scintillator material involves a trade-off between the sensitivity for detection and the decay time of the scintillation. NaI(Tl) is chosen for the efficiency with which it converts an ionization event into visible light, its long decay time, and high light output. Radiation emerging from the patient and passing through the collimator interact with the NaI(Tl) crystal. Interaction of the γ ray with the crystal results in the ejection of an orbital electron (photoelectric absorption), producing a pulse of fluorescent light (scintillation event) proportional in intensity to the energy of the γ ray. Photomultiplier tubes (PMT) along the posterior crystal face detect this light and amplify it. About 30 percent of the light from each event reaches the PMTs. The crystal has an Al housing that protects it from moisture, extraneous light, and minor physical damage.

The crystal may be 10–21.5 inches in diameter and 0.25–0.50 inches thick. A large-diameter crystal has a larger field of view and is more expensive, but has the same inherent resolution as a smaller-diameter crystal. The thicker the crystal becomes, the worse the spatial resolution, but the more efficient the detection of γ rays. In general, with a 0.5 inch thick crystal, the efficiency for detection of γ rays from Xe 133 (81 keV) and 99m-Tc (140 keV) is almost 100 percent; that is, very few of the photons pass through the crystal without causing a light pulse. As the γ energy isotope is increased, the efficiency of the crystal is markedly reduced. For example, with 131-I (364 keV), efficiency is reduced to approximately 20–30 percent. Most crystals in new γ cameras are 0.25 or 0.375 inches thick. With a thinner crystal, the overall sensitivity (count rate) decreases by about 10 percent because more photons pass through, but there is approximately a 30 percent increase in spatial resolution because the PMTs are closer to the event and, thus, can localize it more accurately because there is an increase in light collection.

Photomultiplier tube

A PMT converts a light pulse into an electrical signal of measurable magnitude. An array of these tubes are situated behind the sodium iodide crystal and may be placed directly on the crystal, connected to the crystal by light pipes, or optically coupled to the crystal with a silicone-like material. One or more PMTs record a scintillation event occurring in the crystal. Localization of the event in the final image depends on the amount of light sensed by each PMT; thus, the pattern of the PMT voltage output. Weighing the output of each tube then forms the summation signal for each

scintillation event. This signal has three components: spatial coordinates on X and Y axes, as well as a signal (Z) related to intensity. The X and Y coordinates may go directly to instrumentation for display on the imaging monitor or may be recorded in the computer. The PHA processes the signal intensity.

The light interaction caused by a γ ray generally occurs near the collimator face of the crystal. Thus, while a thicker crystal is theoretically more efficient, the PMT is farther away from the scintillation point with a thick crystal and is unable to determine the coordinates as accurately. Therefore, spatial resolution is degraded. The number of PMTs is also very important for the accurate localization of scintillation events and, thus, spatial resolution: the greater the number of PMTs, the greater the resolution. Early γ cameras used 19 round PMTs; newer cameras use 60 – 93 hexagonal or round PMTs.

Pulse height analyzer

The basic principle of the PHA is to discard signals from background and scattered radiation, or radiation from interfering isotopes, so only photons known to come from the photopeak of the isotope being imaged are recorded. After events occur in the crystal, the PHA handles them in one of two ways—they are either displayed and stored in the computer, or rejected. The PHA is able to make this distinction because the energy deposited by the scintillation event in the crystal bears a linear relationship to the voltage signal emerging from the PMTs.

Signal intensity information is matched in the PHA against an appropriate window, which is really a voltage discriminator. To allow energy related to the desired isotope photopeak to be recorded, the window has upper and lower voltage limits that define the window width. Thus, a 20 percent symmetric window for a 140 keV photopeak means the electronics will accept +140/–14 keV γ rays. Any signals higher or lower than this, particularly from scattered radiation, will be rejected. Some cameras have three PHAs, which allow several photopeaks to be utilized at once. This is particularly useful for radionuclides such as gallium 67.

On newer cameras, the signal processing circuitry, such as preamplifiers and PHAs, are located on the base of each PMT so there is little signal distortion between the camera head and console.

Console controls

The position of the PHA window can be adjusted visually using a multichannel analyzer or Z-pulse display. The multichannel analyzer display is generally more accurate because it indicates the spectrum of the γ ray energy, and shows the photopeak and scatter. New γ cameras allow for fine adjustment, known as autopeaking of the isotope. This essentially divides the photopeak window into halves and calculates the number of counts in each half. If the machine is correctly peaked, each half of the window has the same number of counts from the upper and lower portions of the photopeak. Occasionally, an asymmetrical window is used to eliminate some of the Compton scatter. Figure 3–21 shows where on the spectrum the window would be adjusted for symmetrical and asymmetrical operation.

Image exposure time is selected by console control and is usually a preset count, preset time, or preset information density for the image accumulation. Image density refers to the number of counts per square centimeter of the γ camera crystal face. Other controls are present for orientation and image reversal on the X and Y axes.

In addition, the monitor image may be manipulated by intensity control (simply adjusts the brightness of the image) or persistence control (regulates the length of time the light dots composing the image remain on the screen). Most γ camera systems have two monitors—one for operator viewing and another for photographic purposes.

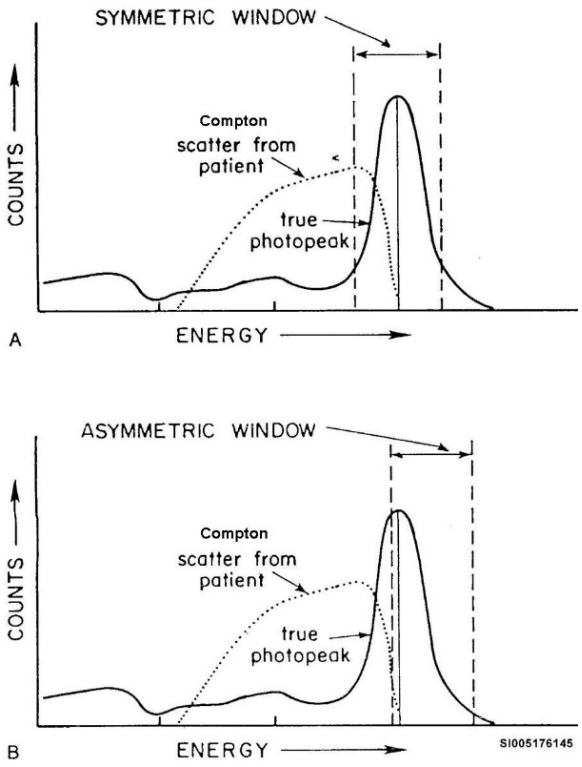

Figure 3–21. Window symmetrical and asymmetrical.

Resolution

This term has been mentioned throughout this whole section. Let's get a fix on exactly what we mean by the various resolutions when discussing γ cameras.

Resolution usually refers to energy or spatial resolution. Energy resolution is the ability to discriminate between light pulses caused by γ rays of differing energies. If the energy resolution is good, the photopeaks are very tall and narrow; if energy resolution is poor, the photopeaks appear as broad bumps in the energy spectrum. Spatial resolution refers to the ability to display discrete, but contiguous, sources of radioactivity. The spatial resolution of γ cameras is usually given in terms of inherent or overall resolution. Inherent resolution is the ability of the crystal PMT detector and accompanying electronics to record the exact location of the light pulse on the sodium iodide crystal. The inherent resolution of the system or its ability to localize an event is directly related to the energy of the isotope being imaged and Compton scatter. When radioisotopes with low-energy γ rays are used, the camera has less inherent spatial resolution. When γ rays interact with the crystal, there is photoelectric absorption that results in a light pulse at the point of interaction. However, with higher-energy γ rays, the initial event may be a Compton interaction, or scatter (i.e., a collision between the γ ray and a loosely bound orbital electron). This results in scattered photons with light coming from several points, even though only a single γ ray interacted with the crystal initially.

Overall spatial resolution is the resolution capacity of the entire camera system, including such factors as the collimator resolution, septal penetration, and scattered radiation—each having the possibility of being the weakest link through the various energy levels it is subjected to.

Specialized γ cameras

The science of nuclear medicine is growing rapidly with the discovery of its usefulness and acceptance in medicine. With this growth has come the development of new technologies and equipment. The basic piece of equipment used in nuclear medicine is still the γ camera, but there are now specialized versions you will likely see.

Whole body and portable system

The material on γ cameras would not be complete without the mention of two systems—whole body and portable. To accomplish whole body imaging, the patient is placed on a moving table or the γ camera detector head is moved over the patient. This method has become popular for bone or gallium scanning. Portable γ cameras are used primarily in cardiac stress laboratories and ICUs. These instruments are designed mainly for use with thallium 201 and 99m-Tc, and may have limited application for general imaging.

Single photon emission computed tomography

The technique of SPECT applies tomographic principles to produce a series of 2D scans from adjacent slices of tissue. The difference between SPECT and planar nuclear medicine is comparable to the difference between regular X-ray imaging and CT. SPECT techniques use many of the same radiopharmaceuticals and equipment as planar nuclear medicine, and most SPECT machines can be used for either. SPECT is used for brain imaging, myocardial perfusion studies, and oncological investigations, among other uses.

SPECT studies use a multidetector or rotating head γ camera system. With a multidetector γ camera, a large number of scintillation crystals and associated electronics are placed around the patient. This system has the advantage of high sensitivity, high spatial resolution, and rapid image production. On the down side, these systems are expensive and complex, and can only be used for SPECT studies; therefore, they are not widely used.

The rotating head system is more common because it can be used for traditional nuclear imaging studies, as well as for SPECT. Data is collected from multiple views obtained as the camera rotates around the patient's head. As the camera rotates, it collects a series of signal projections. These signal projections are corrected for scatter and attenuation, and then filtered and projected as an image. One way to improve the image is to increase the number of cameras in the system. It is now common to see two- and three-camera systems in a nuclear medicine facility.

Positron emission tomography

PET is a relatively new procedure used to measure body function. The image is similar to a CT scan, but unlike CT that only shows anatomy, PET shows how the body part is functioning. PET can be used to study the effectiveness of cancer treatments, diagnose certain diseases (i.e., Alzheimer's), and map brain function. To produce a PET image, patients are injected with a glucose that contains a radioactive tracer. Cells that are the site of higher metabolic activity, which is usually the case with cancer and infection, take up more glucose. Radioactivity emitted by the glucose is recorded on the PET camera and reconstructed by a computer to form an image. The image highlights tissue metabolism, which shows up as bright color on the computer-generated image.

During a PET procedure, positron-emitting radionuclides are introduced into the patient. This special class of radionuclides has an atomic nucleus that lacks a balancing number of negatively charged electrons, causing positron particles to be continually shed. When this phenomenon occurs within body tissues, spare electrons rapidly capture the escaping positron. During this process, called annihilation, two γ rays of identical energy are emitted at 180° to each other.

A PET camera performs a similar function to a γ camera in localizing the distribution of γ rays introduced into the body. However, because a PET procedure emits two γ rays, the PET camera has opposing detector systems to capture both rays. The detector system is aligned within a circular array and coupled electronically to register only γ ray events that coincide with those on the opposite side of the ring—this is called coincidence counting. With the patient encircled by a ring of detectors, a true event is defined as two interactions occurring simultaneously in two detectors on opposite sides of the ring. The line joining these two detectors represents a line of response (LOR) through the patient, somewhere along which the annihilation took place. Many pairs of events are detected at all angles around the patient. The source of radioactivity can then be pinpointed to a straight line between the opposing detectors. With a circular array of detectors, the X and Y coordinates of the

radioactive emissions can be mapped with precision. Also, in contrast to γ cameras that work in 2D or 3D mode, PET cameras operate only in 3D mode.

In planar or SPECT gamma imaging, the line of sight for the projection images is established using collimators. In PET, since the line of sight is established by coincidence detection, collimators are unnecessary. However, ring-shaped septa of lead or tungsten are sometimes positioned between the rings of detector blocks to decrease gamma ray scatter and improve resolution. Some advantages of PET over SPECT imaging are its higher sensitivity and resolution. Because a PET scanner does not require a collimator, it is 100 times more sensitive than a SPECT scanner, and has a higher count rate for similar quantities of radioactivity.

Hybrid SPECT/CT and PET/CT

PET and SPECT imaging have an advantage over CT and MRI, as they are both very high in sensitivity, with the ability to detect picomolar amounts of radiopharmaceutical. However, spatial resolution of PET and SPECT is low and interpretation of PET and SPECT images can be difficult because they contain few anatomical landmarks. Co-registration, or fusion, of PET or SPECT with CT or MRI acquired images in separate imaging sessions has been used as a method of providing anatomical context to functional/metabolic images. To do this in the past, software fusion was used with a limited amount of success. Unless patient positioning was identical on the two scans, the images would not match up exactly when fused, and the anatomical location of functional abnormalities was incorrectly determined.

Technology now allows us the opportunity to use a hybrid PET/CT scanner in which the patient stays on the same imaging table and undergoes PET and CT in the same session, within minutes of each other. This results in considerably more successful image fusion and can improve patient outcomes, particularly in the field of oncology. Most hybrid scanners make use of a third-generation CT scanner in which the X-ray tube rotates around the gantry diametrically opposed to an arc-shaped, rotating detector array. After a scout or topogram, CT image to set the scan area, a noncontrast CT scan is performed. This is followed by a PET or SPECT image of the same body part. The CT image is used to generate a matrix of attenuation coefficients, which is then used for attenuation correction of the PET/SPECT data. The attenuation-corrected PET/SPECT and CT images are fused. Following acquisition of the PET or SPECT, an optional CT with contrast can be performed.

Measurement equipment

We are going to discuss three types of equipment used for measuring radioactivity and dose size—the Geiger counter, sodium iodide well counter, and dose calibrator. Each are used extensively in nuclear medicine laboratories today.

Geiger counter

A Geiger-Mueller counter, commonly called a Geiger counter or GM meter, is a portable ionization chamber. It consists of a gas-filled cylinder attached to a readout meter and a battery pack (fig. 3–22). A wire protrudes into the center of the cylinder. Electrical current from the battery establishes a potential difference, such that the wire becomes a positively charged anode and the wall of the chamber becomes a negatively charged cathode. The end of the cylinder is closed with a mica window and mesh screen. Photons pass through the window into the chamber and ionize the argon gas inside. Positively charged ions are collected at the cathode and negatively charged

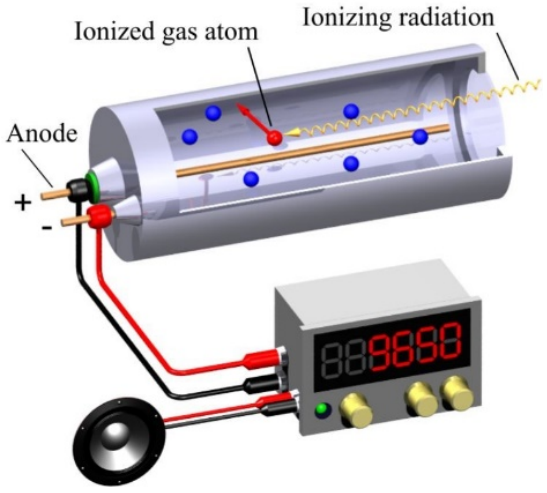

Figure 3–22. Geiger counter (licensed by CC BY-SA 3.0).

ions are collected at the anode. The meter readout is proportional to the number of photons entering the chamber. In the nuclear medicine clinic, Geiger counters are used to monitor areas for radiation and detect contamination of items and people. The meter must be calibrated regularly to ensure proper function.

Sodium iodide well counter

Well counters are very common in nuclear medicine laboratories for performing *in vitro* studies, as well as for quality control and assurance procedures. Many sodium iodide well counters are designed for counting radioactive samples in standard test tubes. Generally, there is a solid cylindrical well cut into the crystal into which the test tube is placed. A PMT is optically coupled to the crystal base. Radiation from the sample interacts with the crystal and is detected by the PMT, which feeds into a scalar. The scalar readout directly reflects the amount of radioactivity in the sample, and is usually recorded in counts for the time period over which the sample was measured. Figure 3–23 shows a typical well counter.

Reflected light, scattering inside the well surface, and the thickness of the crystal limit the energy resolution of the standard well counter. Since the crystal essentially surrounds the sample, the geometrical efficiency for detection of γ rays is high. Geometric efficiency is defined as the fraction of emitted radioactivity incident upon the detection portion of the counter, in this case, the crystal. Because the crystal is relatively thick, most low-energy photons undergo interaction and very few pass undetected. Due to these factors, the overall crystal detection efficiency is better than 95 percent in energy ranges below 200 keV.

It is important to keep the sample volume small because the top of the well counter is open. If varying sample volumes are placed in the well counter, different amounts of radiation will escape near the top of the crystal, resulting in unequal geometric efficiency. Absorption of γ rays within the wall of the test tube is a factor when low-energy sources, such as iodine 125, are counted; therefore, the sample tubes should also be identical.

Due to the fact that sodium iodide well counters have such high detection efficiency they create a serious problem with electronic dead time. If high levels of activity are employed, much of the radiation is not detected. Well counters can typically count only a fraction of a μCi, at rates of approximately 5,000 counts per second. Attempts to measure amounts of activity greater than this in a well counter can lead to serious errors.

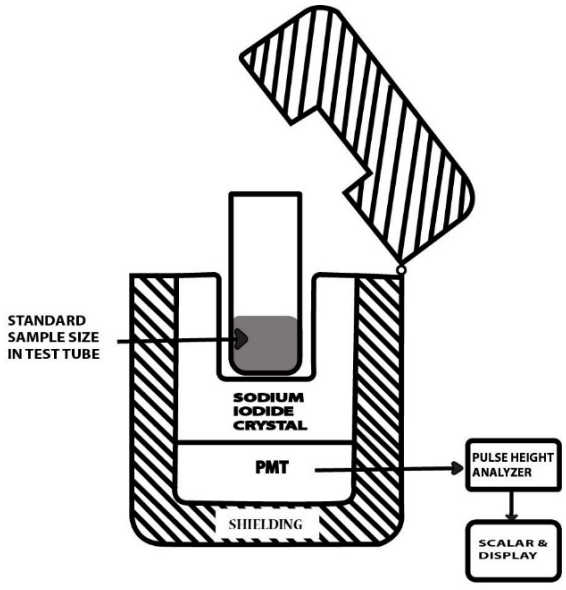

Figure 3–23. Standard well counter.

Dose calibrator

The dose calibrator is an essential piece of equipment because it is extremely important to calibrate a dose of isotope prior to injection. For larger doses, a standard well counter is not useful, since the upper limit of sample activity that can be measured accurately is in the µCi range. A dose calibrator is a well-type ionization chamber capable of measuring quantities in the millicurie range. The cylindrical chamber does not contain a sodium iodide crystal, but holds a defined volume of inert gas. Within the chamber is a collecting electrode. As radiation emanates from the radiopharmaceutical in the syringe, it enters the chamber and interacts with the gas, causing ionization. An electrical differential applied between the chamber and the collecting electrode causes the ions to be captured and measured. Figure 3–24 is a simple dose calibrator.

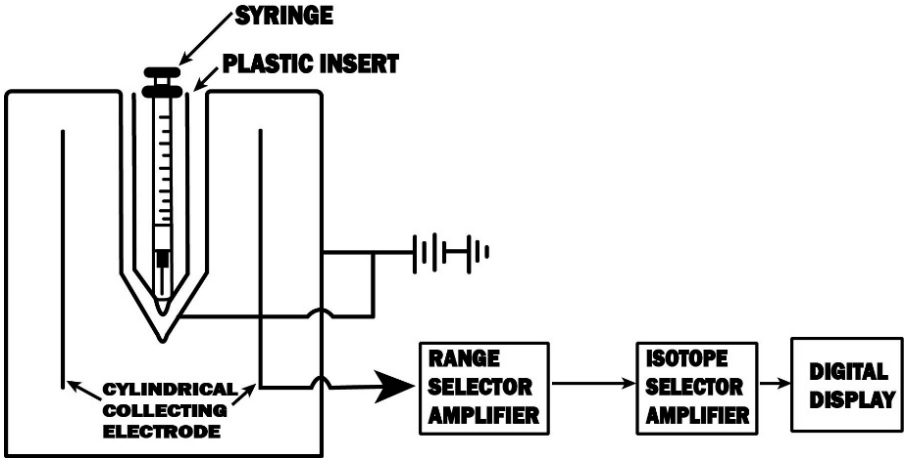

Figure 3–24. Dose calibrator.

Most dose calibrators have a digital readout that indicates the amount of activity in µCi or millicuries, once the radionuclide being measured has been specified. Since all radionuclides do not generate the same number of photons per radioactive decay, the dose calibrator must be recalibrated for each radionuclide to be measured.

Self-Test Questions

After you complete these questions, you may check your answers at the end of the unit.

624. Clinical applications

1. What measures the radiation emitted from the body of a patient during a nuclear medicine study?

2. What is the main difference between nuclear medicine and X-ray imaging?

3. What is the purpose of a stress thallium study?

4. List at least five areas of the body, along with the purpose, that can be studied with nuclear medicine.

625. Principles

1. Match the definitions in column A with the corresponding terms in column B. Items in column B may be used once or not at all.

Column A	*Column B*
____ (1) The basic instrument used to produce a nuclear medicine image.	a. PET.
____ (2) A 3D computer reconstructed image of multiple views and functions of an organ.	b. Planar.
____ (3) A procedure where trace amounts of radiopharmaceuticals are given directly to a patient.	c. SPECT.
____ (4) A method of separating interference from an area of interest by imaging a cut section of an object.	d. RIA.
____ (5) The basic radioactively tagged compound necessary to produce a nuclear medicine image.	e. In Vivo.
____ (6) Procedures done in test tubes.	f. Radiopharmaceutical.
____ (7) A 2D view of the process or function of the organ being imaged.	g. In Vitro.
____ (8) A high-energy, 3D image measuring and determining the function or physiology in a specific organ.	h. γ camera.
	i. Tomography.

2. What causes the emission of γ rays from a radionuclide?

3. Why are compounds that emit α particles not useful in nuclear medicine?

4. Why is the half-life of a radionuclide important?

5. What are the four basic methods used to produce radionuclides?

6. Name at least three common radionuclides used in nuclear medicine.

626. Nuclear medicine equipment

1. Name the basic components of a γ camera system.

2. What is the purpose of a collimator?

3. What is the most widely used collimator in nuclear medicine?

4. What are septa and what purpose do they serve?

5. What kind of scintillation crystal is used for most nuclear medicine applications and why?

6. What does a PMT do?

7. What is the basic principle behind a PHA?

8. Briefly describe energy resolution and spatial resolution.

9. What type of γ camera system is most commonly used in SPECT equipment? Why?

10. How does a PET study differ from a CT scan?

11. Explain the concept of a hybrid PET/CT and what benefit it provides.

12. What three types of equipment are used for measuring radioactivity and dose size?

13. Why is a dose calibrator important in the nuclear medicine laboratory?

3–3. Magnetic Resonance Imaging

Magnetic resonance (MR) technology is based on the existence of nuclear spin by protons within a compound or substance. They were first experimented with in 1944. With the discovery of chemical shift (a small, but specific, displacement of the resonance frequency of a particular nucleus in different chemical compounds), MRI has rapidly progressed into the most powerful nondestructive analytical method to date.

MRI is a form of diagnostic imaging that utilizes powerful magnets and radio waves to produce images, as opposed to ionizing radiation. Based on long understood principles of physics, MRI is now becoming a potentially important method for detecting functional and anatomical changes in living tissues. This is exciting because of the significant new capabilities it is bringing to diagnostic medicine—advances helping physicians diagnose and assess a wide range of disorders more quickly and accurately than ever before, without surgery, invasive biopsy, or radiation exposure.

627. Clinical applications

MRI has proven to be an extremely useful tool in diagnostic imaging. Because of MRI's ability to image on different planes and its sensitivity to tissue differentiation, it has become the procedure of choice for detecting abnormalities or lesions in most parts of the body. Let's take a look at some various anatomies an MRI is currently being used to image.

Brain

MRI on the brain is common. Some symptoms that may indicate the need for an MRI of the brain include headaches, dizziness, visual changes, hearing loss, seizures, nausea, history of cancer, autoimmune disease, and tingling or numbness in extremities. MRI of the brain can help detect tumors and other lesions, metabolic disorders, and multiple sclerosis or other debilitating conditions.

Neck

MRI can help distinguish differences between lymph nodes and blood vessels. Typical indications for MRI of the neck include enlarged lymph nodes or a palpable mass. MRI of the neck can also help detect tumors and other lesions, vascular abnormalities, and structural abnormalities.

Spine

An MRI of the spine may be ordered if a patient is complaining of back pain; numbness or tingling in the extremities; or loss of bladder or bowel control. An MRI of the spine may also be ordered if the patient has a history of cancer. MRI of the spine can help detect herniated or bulging discs, arthritic changes, tumors and other lesions, differences between a post-operative scar or recurrent disc, and structural abnormalities.

Chest

In the past, it was difficult to obtain diagnostic pictures of the chest with MRI due to heart and breathing motion. However, MRI has improved its capability to take pictures by using a set of cardiac leads to monitor heart rhythms and acquire pictures with a "snapshot," eliminating heart motion. Breath-hold imaging utilizes ultrafast techniques while a patient holds his or her breath to acquire motion-free pictures. Times when MRI of the chest may be ordered include a history of cancer and a questionable mass.

Cardiac

MRI is currently improving methods for evaluating the heart. Cardiac gating (coupling with an electrocardiogram [ECG]) reduces heart motion and allows visualization of heart structures, however, coronary arteries are still difficult to evaluate. Typical indications for a MRI of the heart include congenital and acquired heart disease.

Breast

The ability of MRI to differentiate between water, fat, and silicone makes it the procedure of choice for evaluating silicone breast implants or residual silicone from removed implants. MRI of the breast has become increasingly useful for breast cancer diagnosis and staging, as well as screening for women considered higher risk.

Abdomen

With the advent of breath-hold imaging techniques and new equipment, MRI is increasingly used to evaluate the liver, spleen, kidneys, and pancreas. Typical indications for an MRI of the abdomen include history of cancer, pain, loss of organ function, bleeding, cirrhosis of the liver, and hepatitis. MRI of the abdomen can help detect enlarged lymph nodes, metastasis disease, tumors and other lesions, aneurysms, and structural abnormalities.

Pelvis

In women, MRI of the pelvis is increasingly used to evaluate the uterus, cervix, ovaries, bladder, fetus, and placenta. In men, MRI of the pelvis is increasingly used to evaluate the prostate, bladder, penis, and scrotum. Typical indications for an MRI of the pelvis include cancer staging, pain,

palpable masses, and pregnancy complications. MRI of the pelvis can help detect enlarged lymph nodes, fibroids, ovarian masses, prostate cancer staging, metastasis disease, testicular cancer, and structural abnormalities.

Musculoskeletal

MRI is able to evaluate the shoulder, wrist, knee, ankle, and feet with great detail. An MRI of a joint or soft tissue may be ordered when pain, swelling, weakness, palpable mass, or a decrease in range of motion are indicated. MRI of the joints and soft tissue can help detect torn ligaments, torn cartilage, edema, arthritic changes, tumors, lesions, and structural changes.

Magnetic resonance angiography

MRI is now able to use the blood as its own contrast agent to evaluate the blood vessels of the head and neck. This technique is known as magnetic resonance angiography (MRA). MRA can evaluate blood vessels of the head and neck without injecting the patient with a contrast agent. This non-invasive technique requires only one additional set of pictures be taken in addition with a standard MRI exam. MRA of the head and neck can help detect vessel narrowing (stenosis), blood vessel blockage, cerebral aneurysm, and blood vessel dissection.

Contrast-enhanced magnetic resonance angiography

Contrast-enhanced MRA utilizes an injected MRI contrast media into the blood stream while simultaneously acquiring MRA pictures. Contrast-enhanced MRA is now utilized for the evaluation of blood vessels in the thorax, abdomen, pelvis, and legs.

628. Magnetic resonance instrumentation

MRI systems made by different manufacturers offer a variety of features often described in very different terms. It is important to review the manufacturer's information carefully as you learn to work with a specific type of scanner. While there may be many differences, each system has the following basic components: primary and secondary magnets, patient table, shim coils, gradient coils, body RF coil, computer console, pulse sequence controller, image processor, storage unit, and various power units.

The MR scanner is visibly recognizable as a hollow unit and an attached patient table (fig. 3–25). The main or primary magnet is located within the gantry of this unit. While an MRI has a similar appearance to a CT unit, the aperture is generally smaller, which sometimes becomes an issue for claustrophobic patients. Other parts include a computer and operator console.

Primary magnet

MR scanners use one of three types of magnets: permanent, resistive, or superconductive. Nearly all scanners manufactured today use superconducting magnets. MRI magnets in clinical use produce magnetic forces ranging from 0.2–4.0 Tesla (T), or more. Tesla and Gauss (G) are units of magnetic field strength, but Tesla is the unit most commonly used in MRI. One T is equal to 10,000 G (so 0.035 T = 350 G). As a point of reference, the earth's magnetic field is approximately 0.5 G. A 3.0-T system (30,000 G) is, therefore, approximately 60,000 times stronger than the earth's magnetic field.

Permanent magnets

These magnets are made out of materials such as magnetized ceramics, and capable of producing magnetic fields up to about 0.35 T. Disadvantages of permanent magnets include low magnetic field strengths and relatively non-uniform fields, as well as the inability to turn the magnet off. However, they are much less expensive than other magnets and cost virtually nothing to operate.

Resistive magnets

Resistive magnets used in MRI consist of several large coils arranged in a specific configuration that provides the most uniform field possible. Continuous electric power in the 80 kW range is required to energize these resistive coils. The power consumption of such a magnet is equivalent to that of a large office building. When direct current is passed through a wire, a static magnetic field is created around

the wire. If these wires are configured in the shape of a coil, this is known as a solenoid electromagnet. Solenoid electromagnets are lighter than permanent magnets, are low in field strength, and require a power supply. The wire used in such coils is a good, but not perfect conductor. Consequently, these magnets have some small finite resistance, hence the name resistive magnets. Even though the resistance of wire is slight, a significant amount of heat is generated because of the large electric current being passed through the coils. Such a magnet must be water-cooled. The typical field strength of various resistive MR systems ranges from approximately 0.2 T up to approximately 0.7 T.

Figure 3–25. Basic MRI system.

Superconducting magnets

We solve the problem of resistance by using superconducting material to carry the electric current. After being cooled to extremely low temperatures, superconducting materials have virtually no resistance to an electric current. Once the current begins to flow through the coil of a superconducting material, it can continue almost indefinitely without the need for additional power. However, the temperature of the magnet coils must remain near absolute zero (-459.67°F), or they will not maintain their superconducting properties.

Superconducting magnets are cooled by a cryogen, most commonly liquid helium (He). The temperature of liquid He is 4 Kelvin (K), or -452°F. The magnet itself is housed in a large insulated container called a Dewar, which functions like a Thermos bottle. The innermost chamber of the Dewar contains the magnet coils, bathed in liquid He. Surrounding this area is an outer chamber also filled with liquid He recycled from the boil off of the inner chamber, or liquid nitrogen. The two chambers are separated by vacuum chambers, which also isolate them from room temperature.

Despite this insulation, liquid He and nitrogen are subject to evaporation and must be replaced periodically. This represents a significant maintenance cost since cryogens are very expensive and their handling requires special training. Currently, scientists and engineers are working on room temperature superconducting magnets, which would greatly reduce the cost of MRI units; however, this is still not a reality and may be several years away.

The main magnetic field is located within the imager, but also extends out and around the imager. This field (outside the imager) is known as the fringe field or stray field. In order to confine the fringe field, magnetic field shielding is used. Magnetic field shielding is either passive or active. Passive shielding is similar to the shielding used in radiography, whereby the imaging room walls are lined with metal. In x-ray rooms, this metal shielding is lead; in MRI rooms, the passive shielding is iron. (Bear in mind that MR systems are also shielded for RF. Another method for shielding in MRI is known as active shielding. The term active implies activity or current. For MRI, active shielding uses additional electromagnets to confine the fringe field.

The following table demonstrates the various advantages and disadvantages to each type of primary magnet as a summary to our discussion.

Magnet	Pro	Con
Permanent	• Body size can deliver a field in the Gauss range. • Low maintenance cost. • Small fringe field.	• Significant weight (>100 tons). • Low thermal stability. • Foreign objects pulled into the bore are very difficult to remove.
Resistive	• Lowest cost alternative. • Magnetic power to 0.15 Tesla. • 0.12 Tesla achieved with a 20-kW power supply.	• Weak magnetic field. • Instabilities in the power supplies. • High electrical power consumption.
Superconducting	• Higher field available than any other type. • Extremely stable. • Improved signal-to-noise ratio. One of the primary concerns regarding MRI quality. • Several configurations available to adapt to most any site.	• Highest cost alternative. • Cryogen cost.

Secondary magnets

There are several secondary magnets used within the MR scanner: shim coils, gradient coils, RF coils, and others (fig. 3–26). Although the secondary magnet does not need a continuous supply of electric current to function, a power supply is required to ramp up (power up) the magnet and maintain the secondary magnets (shim coils and gradient coils).

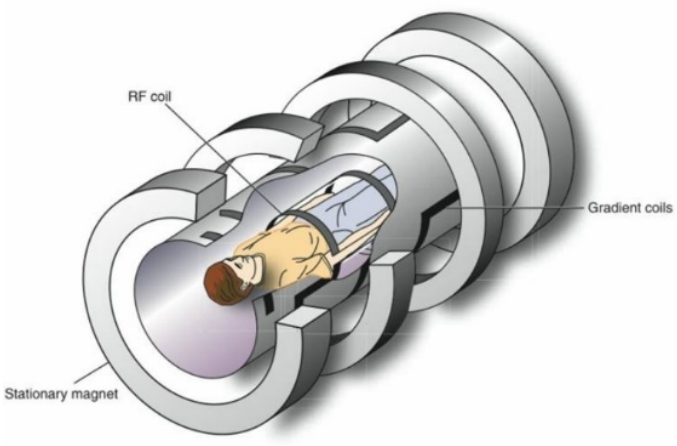

Figure 3–26. The secondary magnet coil system located within the bore of the main magnet. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Shim coils

All primary magnets have irregularities in their fields due to the shape and construction of the magnets themselves, the location of ferromagnetic materials near the MRI unit, and the materials used in the construction of the MRI room. To compensate for these irregularities, a series of corrective magnetic coils, called shim coils, are added.

Shim coils are resistive magnets used to make corrections to the main magnetic field, thus improving the quality of images. Shim coils are located inside the gantry of the MRI scanner along with the magnet and gradient coils. After the main magnet has been brought up to field, the current and polarity of each shim coil is adjusted to produce maximum consistency in the main magnetic field. This process is called shimming the magnet.

Gradient coils

To obtain spatial information about the tissue from which the MRI signal is emitted, it is necessary that the primary magnetic field be slightly varied by using a gradient magnetic field, which is produced by three gradient coils. To obtain projections from a variety of directions, one must be able to orient the gradient magnetic field along the X, Y, or Z axes, or along the oblique plane. Each of the three coils is identified as the X, Y or Z-axis coils, respectively. Normally, the Z-gradient coils are used for selection of a transverse slice, the X-gradient coils for a coronal slice, and the Y-gradient coils for a sagittal slice. Energizing all three coils at once produces an oblique slice. Physically, the gradient coils are embedded in a ring fitting snugly inside the shim coils within the patient aperture.

RF coils

RF energy is transmitted and detected by the MRI scanner. It delivers RF pulses to the patient and records resultant RF signals. Because RF coils must be placed as close as possible to the body part being imaged, they are often sized and shaped to fit specific body parts. These are called surface coils. Some common coils are head, neck, body, extremity, and general-purpose coils (fig. 3–27). In general, larger coils are uniform with regard to transmission, whereas smaller coils are more sensitive with regard to reception.

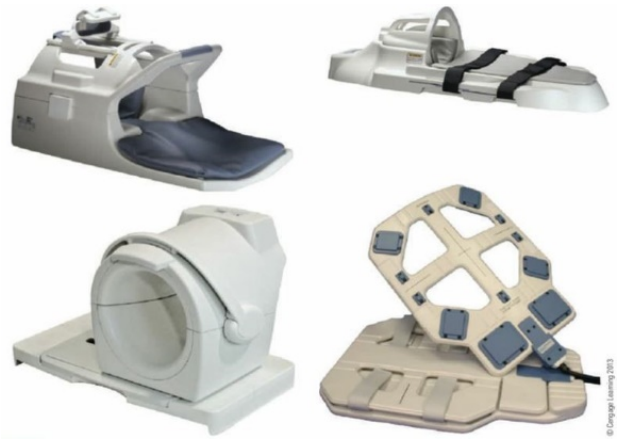

Figure 3–27. Various RF coils. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Computer console

The MRI scanner also includes a host computer that controls the various parts of the system and allows the operator to interact with the system. A computer console provides the operator with a keyboard, and one or more video monitors. Depending upon the type of MRI scanner at your facility, the main console acts as a one- or two-person station that allows you to input data, select commands from the menu, and evaluate images. On command, the video monitor displays stored images or images being generated.

The video monitor also displays menus that allow you to select pre-set commands or input original data for scanning and other purposes. Again, depending on the type of MRI scanner used, you can interact with the computer in a variety of ways. Some consoles have touch-sensitive monitors that allow you to make a selection by touching a box on the screen. Other devices commonly seen on the console include a mouse, trackball, and, of course, keyboard.

It is very common for MR systems to include one or more secondary consoles in addition to the main console. These consoles are used to archive or retrieve images, or as secondary workstations for scanning. This is much like the physician's viewing console used in CT.

Pulse sequence controller, image processor, and storage unit

The pulse sequence controller ensures RF pulses are transmitted accurately, using the time intervals selected by the operator. The image processor performs the calculations and other operations necessary to convert the raw RF signal data into an MRI image.

The image storage unit archives the image electronically, storing it into computer memory. Images may be retrieved and displayed on the video monitor, be printed onto film by selecting the appropriate commands on the computer, or be part of a PACS.

Power units

A number of power units supply electricity to the MRI scanner. They operate the secondary magnet coils, the host computer, and other system components. Initially, power is also used to ramp up the superconducting magnet, but this is turned off and normally not used again.

629. Operating principles

MRI is often compared to CT; however, MRI is very different from CT and provides a number of significant advantages including:

- Superior soft tissue contrast resolution.
- Direct multiplanar imaging.
- Direct flow measurements.
- No ionizing radiation.
- No bone or air artifacts.

These advantages stem from the differences in the physical principles that form the basis for MRI and CT. CT, as you know, utilizes X-ray attenuation for image formation. MRI, on the other hand, depends not on just a single parameter (i.e., X-ray μ), but on three principal independent parameters—T1, T2, and spin density (SD)—plus several secondary parameters. As shown in figure 3–28, these MRI parameters have a considerable range of values from one tissue to another. Look at the minor differences in HU from one tissue to the next. The three parameters used in MRI make each tissue unique and easily distinguishable, hence the superior low contrast resolution.

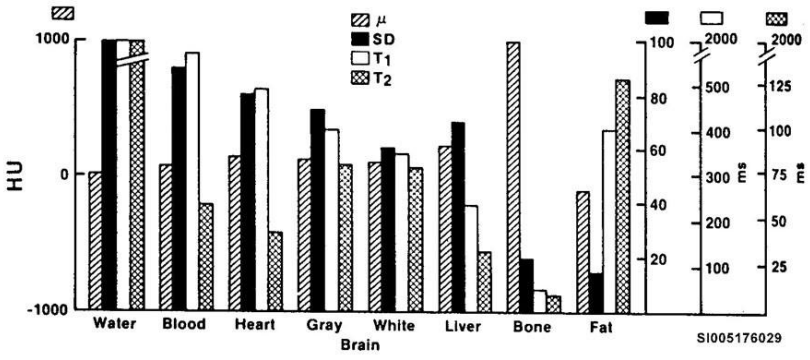

Figure 3–28. Histogram of the three MRI parameters against the CT attenuation coefficient.

MRI images demonstrate subtle differences in contrast between body tissues (especially soft tissue), which respond in varying ways to radio waves due to the difference in their chemical make-up. Bone is not imaged directly and, therefore, provides an absence of signal, allowing an excellent demonstration of brain tissue.

We'll now continue our discussion with some introductory concepts, followed by nuclear magnetism, longitudinal and transverse magnetism, resonance, longitudinal and transverse relaxation, and image weighting.

Introductory concepts

MRI images are created by sending radio waves to a patient lying in an external magnetic field, measuring the resultant signal from body tissues, and assembling this signal into a visual representation of internal body anatomy.

Water

A variety of chemical substances (i.e., sodium, potassium, and fluorine) within the body can be used to generate detectable signals in MRIs. However, for imaging purposes, it is desirable to rely on a substance that is generally available in all body tissues, even if present in different amounts. This substance is water, of course. The human body consists mostly of water. Each water molecule consists of two hydrogen atoms bound to one atom of oxygen (H_2O). Clinical MRI is based on the generation of signals from hydrogen atoms in the body. Most of these atoms may be found in one of three forms: as free (bulk) water in body fluids, such as cerebrospinal fluid; as water bound to large molecules, such as proteins; or as hydrogen atoms within fat. While fat plays a significant role in MRI, for the time being we'll focus on the impact MRI has on hydrogen atoms in water.

Magnetic field

The magnetic field generated by an MRI magnet is much larger and more powerful than a bar magnet. Like the bar magnet however, the MRI scanner generates the magnetic field with a specific strength and direction (fig. 3–29). The vector B_o, a convention used throughout this discussion, commonly represents both strength and direction.

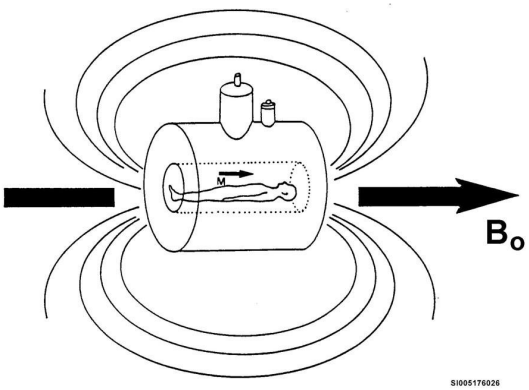

Figure 3–29. Field lines demonstrating magnetic force around the MRI magnet.

Electromagnetic energy

Radio waves are low-energy electromagnetic waves that occupy a substantial portion of the electromagnetic spectrum. They consist of a range of frequencies commonly used for communication broadcasts. Higher frequency radio waves are also used in MRIs. Radio waves are commonly referred to as RF waves, while pulsed radio waves are called RF pulses.

By directing RF pulses at a specific frequency to hydrogen atoms in a magnetic field, it is possible to generate detectable signals. These signals are analyzed and recorded as an image. Hydrogen atoms

respond to RF pulses with coherent signals because they have magnetic properties of their own that allow them to behave in specific and predictable ways when exposed to an external magnetic field.

Nuclear magnetism

Hydrogen has the simplest atomic structure compared to all other elements. Hydrogen atoms usually contain only one proton and one electron. This means the hydrogen nucleus may also be referred to as a "proton." In fact, people often refer to hydrogen nuclei as protons when they discuss MRI.

Nuclear magnetic moment

As fundamental particles, protons are associated with a property called spin. Spin has a different meaning for subatomic particles than it has for visible objects—these particles are so tiny that commonplace ideas of motion don't really apply, and proton spin takes on an abstract, mathematical meaning. However, spin can be imagined as motion, and certainly the best way to visualize this property in protons is to see them as spinning tops. The spin of different protons may be added together, resulting in an overall spin, which is the spin of the nucleus. Therefore, an atomic nucleus can also be imagined as a tiny spinning top.

Scientists have known for a long time that electricity and magnetism are related forces. Faraday's law of induction states that a changing magnetic field produces an electric current in a closed loop of conducting material. Conversely, an electric current flowing in a circuit produces a magnetic field around the wire forming the circuit. In fact, any moving charged particle generates magnetic force. As a spinning charged particle, a proton is also associated with magnetic force. Each proton generates a tiny magnetic field called a magnetic moment. The proton itself can be compared to a tiny bar magnet with a north and south pole. Since it contains two magnetic poles, the proton is called a magnetic dipole.

Since a hydrogen nucleus consists of a single proton, it is also a magnetic dipole characterized by a magnetic moment. This is the property that allows hydrogen atoms to behave in predictable ways within the magnetic field produced by the MRI scanner. The most important concept to remember is the nuclear magnetic moment, because it sets the stage for understanding the alignment of hydrogen in a magnetic field, which in turn sets the stage for the generation of MRI images.

Alignment of hydrogen nuclei

Within body tissues, hydrogen nuclei are normally oriented at random with their magnetic dipoles pointing in all directions (fig. 3–30, C). However, when these nuclei are placed within an external magnetic field (i.e., the field of the MRI scanner, symbolized by B_o[fig. 3–30, A]), they align themselves in one of two directions (fig. 3–30, B).

Low-energy hydrogen nuclei align themselves with the magnetic field (parallel orientation). High-energy nuclei align themselves against the magnetic field (anti-parallel) orientation. Although the number of hydrogen nuclei oriented in either direction is quite large, the relative proportion is more important: slightly more than half are oriented parallel to the magnetic field, while the balance are oriented against the magnetic field. This results in a constant difference between the number of parallel and anti-parallel nuclei. Low-energy nuclei may outnumber high-energy nuclei by only a few protons per million, but this difference is all that counts. The rest of the hydrogen nuclei simply cancel each other out, leaving the remaining low-energy nuclei to determine the overall magnetization of the patient's body. In figure 3–30 (B), hydrogen nuclei are shown oriented parallel to an external magnetic field. The individual magnetic moments of these nuclei can be added together, resulting in the bulk net magnetization symbolized by the vector M.

Since M plays a significant role in the MRI process, it is important to have a clear understanding of what this quantity means. Remember it actually represents the sum of many small magnetic moments, generated by the slight majority of hydrogen nuclei oriented parallel to the external magnetic field. The strength and direction of M are determined by the behavior of these nuclei; any change in M reflects a change in their status.

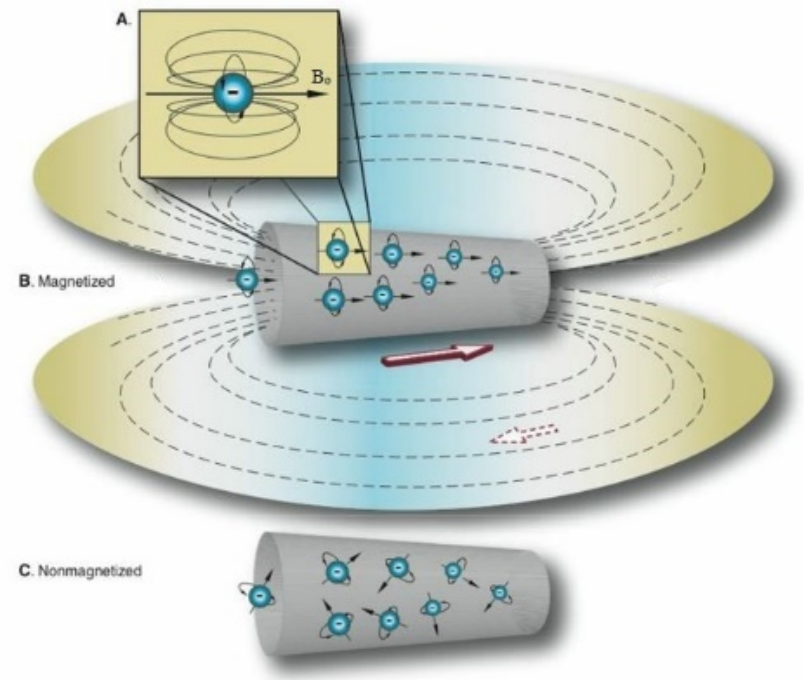

Figure 3–30. Alignment of hydrogen nuclei in a magnetic field. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

Precession

Precession is a type of rotation that can also be described as a wobble. You have seen precession if you have ever noticed a spinning top wobble or if you have seen the motion of a gyroscope. Hydrogen nuclei precess in a magnetic field because they are not aligned exactly parallel or anti-parallel to the magnetic field. Instead, they are oriented at an angle to B_o and precess around it (fig. 3–31, A). As they precess, hydrogen nuclei follow a cone-shaped path (fig. 3–31, B).

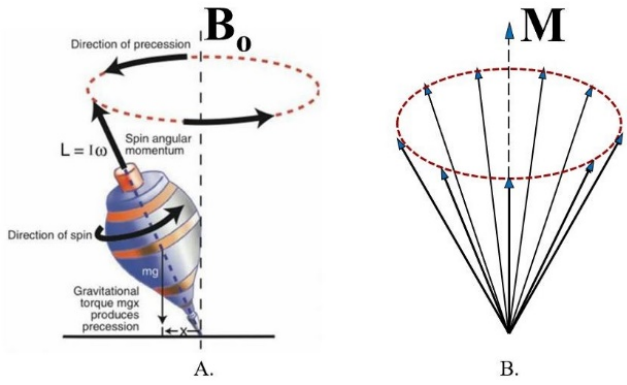

Figure 3–31. Precession of hydrogen nuclei.

Within a magnetic field, all protons precess is at the same rate or frequency. This frequency is determined by the strength of the magnetic field and a constant, γ. The precise relationship between the two is shown by the Lamor equation:

$$F = \gamma B_o .$$

In this equation, F represents precessional frequency, B_o represents the strength of the external magnetic field, and γ is the gyromagnetic ratio. The gyromagnetic ratio is a constant for each type of magnetic nucleus. Hydrogen has one value for γ and carbon –13 another. From looking at this

equation, you can see the precessional frequency of protons depends clearly on the strength of the magnetic field. This means the stronger the magnetic field, the higher the precessional frequency. The precessional frequency is also known as the Lamor frequency, and is obviously an important quantity in MRI.

Longitudinal and transverse magnetization

After a patient is positioned within the magnetic field of an MRI scanner, the second step of MRI begins. A burst of radio waves (RF pulse) is transmitted to the patient's body. If the frequency of the RF pulse matches the Lamor frequency, it has two effects:

1. It alters the precession of hydrogen nuclei.
2. It flips vector M away from its equilibrium position.

Changes in precession are talked about in the next section. To understand changes in M, it is necessary to divide the vector conceptually into two parts: longitudinal and transverse magnetization. These two components represent the strength of M in two different directions. Longitudinal and transverse magnetization are often represented as vector components of M (fig. 3–32).

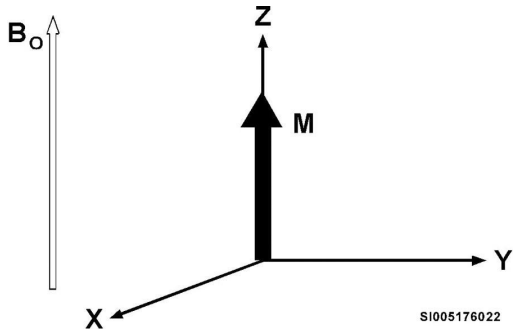

Figure 3–32. Longitudinal and transverse magnetizations represent the value M in two directions.

On a 3D coordinate system, the vertical axis Z designates the longitudinal direction in MRI, while the horizontal axis Y designates the transverse direction. Suppose vector M points in the longitudinal direction. Its longitudinal component (Z) is equal to the full magnitude of M, and its transverse component (Y) is zero. Conversely, if M points in the transverse direction, its transverse component is equal to the full magnitude of M, and its longitudinal component is zero. If M lies between the longitudinal and transverse axes, its longitudinal and transverse components have magnitudes larger than zero. See figure 3–33 for the three described examples.

As part of the initial conditions of an MRI scan, M is oriented in the longitudinal direction, parallel to the magnetic field. Thus, the transverse magnetization of M is zero and its longitudinal magnetization is equal to the full magnitude of M. Sending an RF pulse to the patient changes M by moving it away from the longitudinal axis. This changes the values of the longitudinal and transverse magnetization, and forms the basis for MRI.

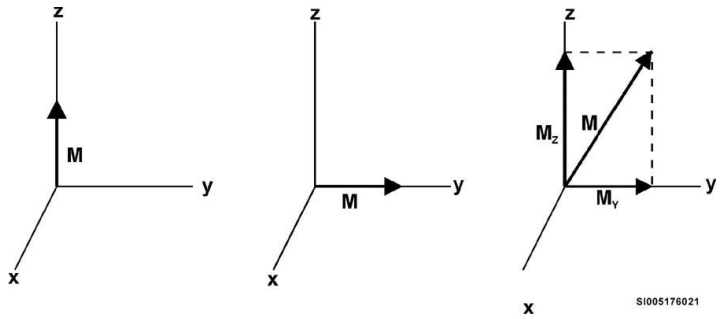

Figure 3–33. Three positions of M.

Resonance

We have now reached the third step of MRI, resonance. First, a patient lies within a uniform magnetic field; second, an RF pulse is transmitted to the patient; and third, hydrogen atoms within the patient's body interact with the RF pulse.

Resonate frequency

The RF pulse causes the hydrogen nuclei to absorb energy and resonate. For resonance to occur, however, the RF pulse must have the same preferred frequency (resonate frequency) as the nuclei.

Remember, this is the Lamor frequency. As these nuclei absorb energy, they become excited, undergoing a transition from a low-energy to a high-energy state. Excited hydrogen nuclei then align against the magnetic field in the anti-parallel orientation.

The purpose of an RF excitation pulse is to excite a certain portion of low-energy hydrogen nuclei. The behavior of these excited nuclei alters the M of the entire sample. Unlike the nuclei, M does not adopt an anti-parallel orientation; instead, it rotates away from its initial vertical position. Longer and stronger RF pulses cause more hydrogen nuclei to absorb more energy, resulting in larger rotations of M.

Phase coherence

Besides inducing hydrogen nuclei to absorb energy, the RF excitation pulse forces hydrogen nuclei to precess in phase or in unison. Nuclei precessing in phase are said to be phase coherent. Phase coherence is maintained as long as the RF excitation pulse is left on. After the RF pulse is shut off, hydrogen nuclei rapidly lose phase coherence and, once again, begin to precess out of phase and revert back to different directions. However, while it lasts, phase coherence is an important effect, with its own influence on the behavior of M.

Net magnetization

As individual hydrogen nuclei respond to the RF excitation pulse, the vector M rotates away from its vertical (longitudinal) position. M rotates by a specific distance or angle, known as the flip angle. RF excitation pulses are named after the flip angles they induce; thus, we have 90° RF pulses leading to 90° flip angles, and 180° RF pulses leading to 180° flip angles.

Figure 3–34 represents M as a vector "flipped" into the transverse plane after a 90° excitation pulse. This is a convenient way to think of what happens to M, but it only describes part of the motion that actually occurs.

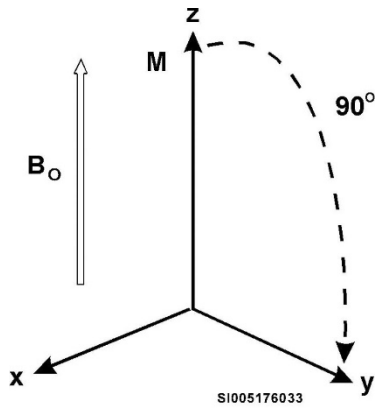

Figure 3–34. Demonstrates the 90° RF pulse and resultant rotation of vector M.

After a 90° RF pulse is applied, M begins to rotate or precess. At the same time, it moves from its longitudinal orientation to the transverse plane. This results in a spiral motion, like a path winding down a mountain. Figure 3–35 shows M spiraling down into the transverse plane after an RF pulse has been applied.

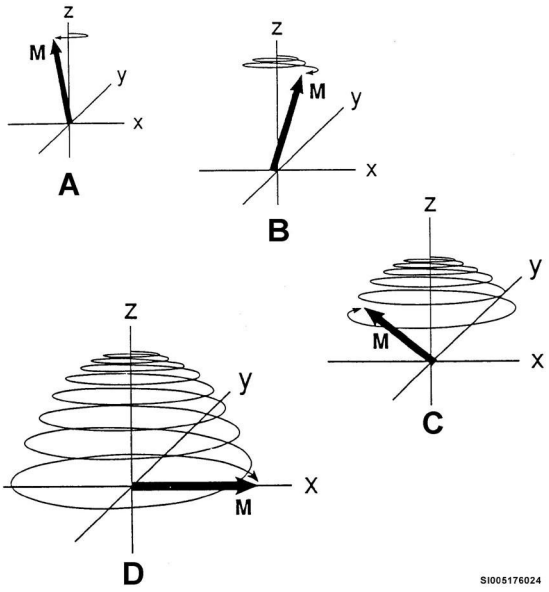

Figure 3–35. Instead of "flipping" into the transverse plane, M spirals down as a rotating vector.

At this point, it is worth noting the changes that occur in longitudinal and transverse magnetization as M spirals down toward the transverse plane. Before the RF excitation pulse, longitudinal magnetization is equal to the full magnitude of M and transverse magnetization is zero. But after the pulse is transmitted and M rotates 90°, transverse magnetization equals the full magnitude of M and longitudinal magnetization is zero. Almost immediately after the RF pulse is applied, it is shut off. M then begins to return to its equilibrium position. Longitudinal magnetization increases as the transverse magnetization progressively becomes smaller and returns to zero. This happens through a process called relaxation.

Free induction decay

After M undergoes its 90° rotation, it spins or precesses within the transverse plane like a spinner in a board game. This precession is useful, because it induces an electric current within a coil of wire used as a receiver. This coil of wire is also known as a RF receiver, because by detecting the motion of M, it is detecting the presence of RF waves—the MRI signal (fig. 3–36).

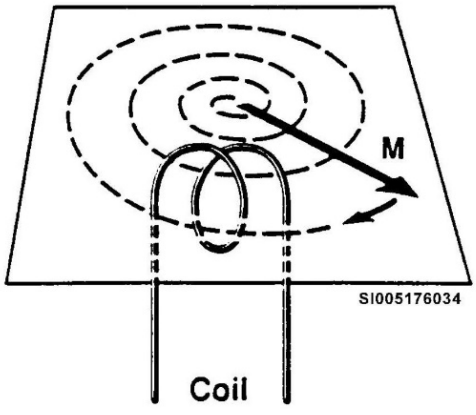

Figure 3–36. Precession of vector M within the transverse plane.

The current detected in the RF receiver coil detects and records RF signals, or the precession of M, in the transverse plane. As M returns to equilibrium, the signal grows weaker or decays. This signal

decay is called free induction decay (FID). It may now seem we have reached the end of our discussion by identifying FID as the signal produced by excited hydrogen atoms. Yet, FID is not used by itself for clinical imaging and its generation is not the final step in the MRI process. One reason for this is the lack of information provided by a simple FID signal: it does not convey spatial information (the location of signal in body tissues, expressed as light and dark areas on an MR image). A variety of techniques have been developed to modify the MRI signal, enabling it to carry enough information for image generation. The MRI scanner performs some of these techniques automatically, while others are under operator control.

Longitudinal and transverse relaxation

The return of resonating hydrogen nuclei to equilibrium is a matter of considerable interest in MRI, since this process influences the type of signal obtained for imaging. As excited hydrogen nuclei return to their original states, two events occur: energy is released and the system becomes more random. These two changes—the loss of energy and the increase in randomness—are called relaxation. To examine relaxation in-depth, we need to break down the vector M into its component parts—longitudinal and transverse magnetization.

Longitudinal relaxation

Longitudinal relaxation is the return of the longitudinal magnetization to equilibrium and transverse magnetization to zero. It is also known as spin-lattice relaxation, because it is the process by which hydrogen nuclei give up energy to their environment. After a 90° excitation pulse, some hydrogen nuclei absorb energy and orient themselves anti-parallel to the magnetic field. During longitudinal relaxation, these nuclei lose their energy and drop back down to their lower energy state. This restores the original equilibrium conditions, where a slight majority of low energy nuclei align themselves parallel to B_o. The vector M reflects these changes by first rotating down to the transverse plane after the 90° RF pulse and then undergoing changes that re-establishes its original position. Immediately after the RF pulse, longitudinal magnetization falls to zero, and then steadily increases until it reaches its original magnitude.

Transverse relaxation

Transverse relaxation is the return of transverse magnetization to equilibrium. It is also known as spin-spin relaxation. This name refers to the interactions between the individual hydrogen nuclei that are partially responsible for transverse relaxation. Unlike longitudinal relaxation, transverse relaxation does not reflect the absorption or dissipation of energy. Instead, it is related to the amount of randomness in the movement of hydrogen nuclei. After an RF excitation pulse, excited hydrogen nuclei precess in phase. As transverse relaxation takes place, the nuclei diphase, which means they lose their phase coherence and begin precessing more randomly. Dephasing occurs because hydrogen nuclei interact, exerting magnetic force through their individual magnetic moments. It also occurs due to local magnetic field inconsistencies and other factors. These influences lead to differences in the magnetic forces experienced by individual nuclei, which then begin to precess slightly faster or slower. The nuclei move further out of phase with one another until they are once again precessing at random.

When you look at longitudinal and transverse magnetization as component vectors of the vector M, one caution to remember is it is only an analogy—they aren't really vectors; they are forces represented by vectors. During relaxation, longitudinal magnetization is simply a force that increases in magnitude, while transverse magnetization decreases in magnitude. These processes may occur at the same rate, but if they don't, transverse magnetization always decays more rapidly than longitudinal magnetization completes its regrowth. Longitudinal and transverse relaxations occur at exponential rates that will not be fully defined here. However, it is important to note the rate for longitudinal relaxation is characterized by a time constant, T1. Similarly, the rate for transverse relaxation is characterized by another time constant, T2.

Spin echo signal

Generally, the FID signal decays too quickly for convenience and it is preferable to work with RF signals that last a little longer. A number of techniques have been developed to delay the disappearance of transverse magnetization. One well known technique is to transmit a 180° RF pulse at a specified time after the 90° excitation pulse. After the 90° excitation pulse, hydrogen nuclei begin to dephase. The 180° pulse "flips" M by 180° and rephases its components. This results in the formation of a spin echo signal, which then begins to decay as the first signal did—by dephasing. To obtain another spin echo signal, a second 180° pulse is transmitted. Although the amplitude of the RF signal decreases with every echo, the use of repeated 180° RF pulses after a 90° excitation pulse allows the spin echo to last much longer than an FID signal. Eventually, signal decay progresses to the point where no further echoes can be obtained. Overall, the rate of decay for a spin echo signal is characterized by T2.

T1 and T2 relaxation

At this time, you need to start thinking of longitudinal relaxation as T1 relaxation and transverse relaxation as T2 relaxation. These terms emphasize the fact that longitudinal and transverse relaxations occur at specific rates characterized by T1 and T2, respectively.

Within a single population of hydrogen nuclei, T1 and T2 are constant, but they are different for different populations. These populations of hydrogen nuclei are contained within the body tissues that are characterized by unique conditions, including the amount of water and particular chemical constituents present. We now have the basis for distinguishing among the different body tissues, which, after all, is the whole point of diagnostic imaging. Different body tissues have different values for T1 and T2. These values influence the type of signal obtained during MRI. The differences in signal are reproduced as images, creating high-quality diagnostic images that show sensitive distinctions between different types of body tissue. The generation of this image does require complex and sophisticated technology.

Image weighting

The parameters selected by the operator at the MRI console (to control the contrast of the MR image) are referred to as extrinsic parameters. These include things like time of repetition (TR), and echo delay time or echo time (TE). TR and TE are two MRI operator parameters that must be understood before image weighting can be defined.

TR is the amount of time allowed to elapse between successive 90° RF pulses. It is usually measured in ms. TE is the time interval that elapses between a 90° RF pulse and measurement of the first spin echo signal—if the 90° RF pulse is followed by one or more 180° RF pulses. It is also measured in ms.

The MRI operator can manipulate the TR and TE by altering the time intervals between RF pulses to produce varying types of MR images. A series of RF pulses transmitted at preset values for TR and TE is known as a pulse sequence.

T1 image weighting

Two of the most common types of MR images are T1 and T2 weighted images. T1 weighting is performed to produce images in which the contrast between tissues reflects differences in T1. Similarly, T2 weighting produces images that reflect differences in T2.

T1 weighting is performed by manipulating TR. To understand how this works, we'll look at the relationship between T1 and TR. In figure 3–37, two tissues (A and B) are shown after a 90° RF excitation pulse. Note that for tissue A, T1 relaxation happens more quickly than for tissue B. Regrowth of the longitudinal magnetization associated with A is completed first, before regrowth of longitudinal magnetization associated with B. In other words, A has a shorter T1 than B.

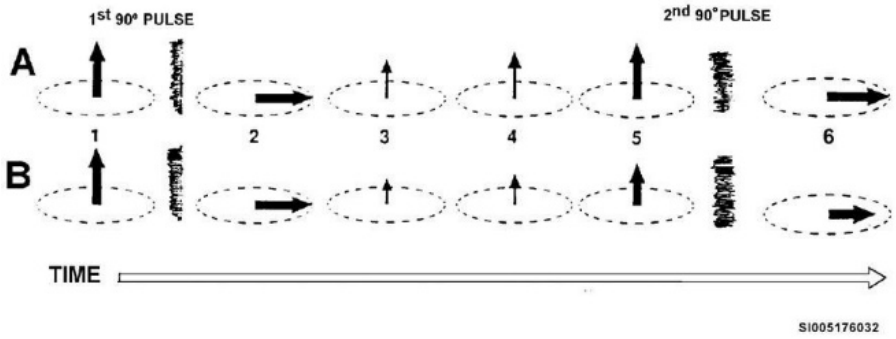

Figure 3–37. Comparison of two tissues and their T1 relaxation time.

If we wait long enough, A and B complete their relaxation, and longitudinal magnetization for both tissues returns to equilibrium. However, if we do not allow this process to come to completion and transmit another 90° excitation pulse instead, each longitudinal vector will rotate directly into the transverse plane without any further increase in magnitude. After rotation, the transverse vector associated with A is larger than the vector associated with B (fig. 3–37, #6). Since transverse magnetization is directly related to RF signal, this means the signal intensity emitted by A is higher than that emitted by B. This is how differences in T1 influence RF signal intensity as measured from different tissues. Light and dark areas on MR images represent variations in signal intensity.

To produce images based on T1 differences, we use a short time interval between 90° excitation pulses. If we want to minimize differences in signal intensity related to T1, we use a long time interval, allowing longitudinal magnetization for all tissues to reach equilibrium. As you should recall, we defined the time intervals between 90° RF pulses as TR. This means a short TR must be used to generate T1 weighted MR images.

T2 image weighting

Since T2 relaxation is based on the decay of transverse magnetization rather than the increase of longitudinal magnetization, T2 weighting is based on somewhat different principles. After a 90° excitation pulse is transmitted to body tissues, it is followed by one or more 180° RF pulses. T2 relaxation occurs and transverse magnetization decays at a rate characterized by T2. Tissues with short T2 are characterized by rapid decay of transverse magnetization, while tissues with long T2 maintain transverse magnetization for longer periods of time. The effect this has on RF signal intensity depends on when you choose to measure the signal. If you allow enough time to elapse before taking your measurement, the amount of transverse magnetization associated with tissue A will be significantly greater than the amount of transverse magnetization associated with tissue B. As before, this means the RF signal obtained from A is higher than the signal intensity obtained from B.

On the other hand, if you make your measurement without allowing very much time to pass, the transverse magnetization vectors for A and B will not have decayed significantly. The differences in their decay rates may not be apparent in your measurement. Each vector has the same magnitude and the RF signal intensity obtained from both tissues is about the same. The time interval between a 90° excitation pulse and measurement of spin echo signal is TE. Thus, to maximize differences in RF signal based on T2, it is important to use a long TE. To minimize these differences, a short TE is used. Optimal settings for TR and TE vary for institutions using MRI scanners with different magnetic field strengths.

Intermediate SD MR images

Although T1 and T2 have very important effects on RF signal intensity, there is another parameter that influences the amount of signal obtained from body tissues. This parameter is the number of hydrogen nuclei within a specified volume of tissue—also called proton density or SD. Tissue

containing large numbers of hydrogen nuclei emits a higher intensity signal, while tissue with fewer hydrogen nuclei emits fewer signals. One example of the influence of proton density is bone, which contains little water and emits very little RF signal. This is why bone is generally not visible on MR images, although tissues associated with bone such as cartilage and bone marrow may be visualized. Pathological processes that increase the amount of water in bone may also be visualized on MR images.

Images that demonstrate differences in signal intensity based solely on proton density may be obtained by minimizing T1 and T2 effects. Long TR minimizes T1 effects and short TE minimizes T2 effects. A strict definition of proton density for MR images specifies the value for TR must be at least four times as long as T1 (of the tissue of interest), while TE must be as close to zero as possible. In practice, many images with long TR and short TE are actually generated as intermediate weighted MR images, due to the stringent requirements for generating a true proton density image.

Clinical MRI is performed by combining effects from all three parameters—T1, T2, and SD—in the most constructive way possible to highlight differences between body tissues. Figure 3–38 is a diagram of TR and TE in a spin echo pulse sequence. Hopefully, this will help you visualize how TR and TE fit in with T1 and T2 weighted images.

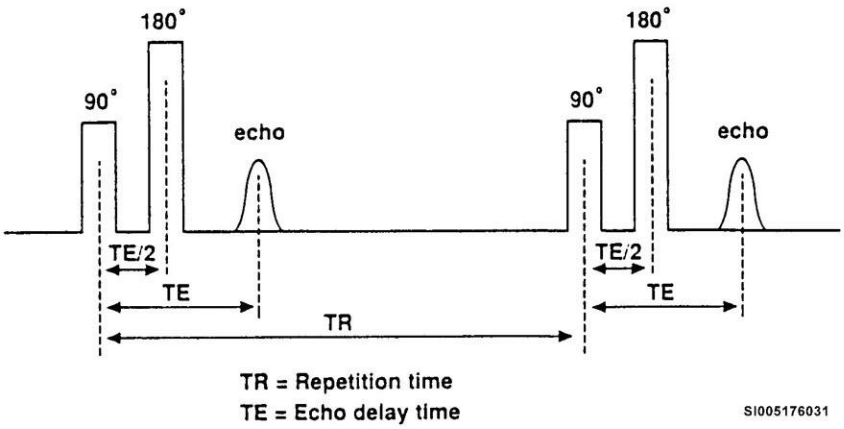

Figure 3–38. TR and TE in a spin echo pulse sequence.

Image acquisition and reconstruction

The process begins as a slice selection of the magnetic field gradient (gradient coils) turns on just prior to RF excitation, causing hydrogen nuclei in a single body slice to resonate. The unit then applies a phase-encoding gradient, followed by a frequency-encoding or read-out gradient. These two gradients code the voxels of the slice along two coordinate axes, allowing the signal from these voxels to map to the pixels of the MRI.

Although the RF signal is received from multiple locations within the tissue slice at the same time, the phase and frequency information it contains is readily obtained through a mathematical function called Fourier transformation—the black box of the computer system. As the signal maps to the image matrix, bright pixels represent a high-intensity signal and dark pixels represent a low-intensity signal. This is known as 2D MR imaging.

In 3D imaging, the entire tissue volume is excited by an RF pulse. A variable-phase-encoding gradient is then used for slice selection. At the same time, another phase-encoding gradient is applied to the phase axis. Finally, a frequency-encoding gradient is used along the remaining axis and a signal is detected. This method provides the advantage of thinner, contiguous slices. Multi-slice imaging is performed by exciting and detecting signals from different slices at the same time. It improves the efficiency of data acquisition, reducing scan time.

By utilizing mathematical calculations, the computer does a complete reconstruction of the image. We are not going to go into this reconstruction phase as we have discussed slice image reconstruction at length in previous lessons. The main focus of the MRI section is image production, acquisition, and theory, as these concepts are significantly different from other imaging modalities.

Self-Test Questions

After you complete these questions, you may check your answers at the end of the unit.

627. Clinical applications

1. Identify the appropriate symptoms that may indicate the need for an MRI examination and some possible conditions the exam may detect for the brain, neck, spine, abdomen, pelvis, and musculoskeletal parts of the body.

2. What is used as a contrast agent during MRA?

628. Magnetic resonance instrumentation

1. Name the three types of primary magnets used in MR systems.

2. How many times stronger is a 3.0-T system than Earth's magnetic field?

3. What is the temperature requirement for the magnet coils of a superconducting magnet in order to maintain its superconducting properties?

4. What kind of cryogen is used to cool a superconducting magnet?

5. What is a shim coil used for?

6. What are some common types of surface (RF) coils?

7. How are secondary consoles used?

629. Operating principles

1. What substance within the body is relied upon for production of an MR image?

2. What is the symbol for strength and direction of the magnetic field?

3. What is the magnetic field produced by the proton called?

4. How do hydrogen nuclei align themselves within body tissues? How do they align themselves within an external magnetic field?

5. What does M symbolize?

6. Using the Lamor equation, explain the relationship between precessional frequency and field strength.

7. What two things occur when an RF pulse in the Lamor frequency is transmitted to a patient's body within a magnetic field?

8. How do the hydrogen atoms within a patient's body interact with an RF pulse?

9. How long does phase coherence occur?

10. Name the two common flip angles used in MRI.

11. The RF receiver coil detects and records signals in what plane?

12. What is longitudinal relaxation?

13. What occurs with the nuclei during transverse relaxation?

14. What do T1 and T2 symbolize?

15. What is TR and TE?

16. What length of TR should we use for T1 weighted images?

17. What are T2 images based on?

18. What is done to TR and TE when a signal intensity based solely on proton density is needed?

19. Name the three parameters used for constructing an MR image.

20. What is the name of the mathematical function used to reconstruct an image?

3–4. Diagnostic Ultrasound

Ultrasound is a diagnostic imaging modality that has become an important part of radiology. Although it has been used in other fields for quite some time, its application to the medical field is still relatively new. You should recall the basic ultrasound principles from the basic BMET course, as well as the lessons on ultrasound therapy from 4A251B, volume 2, so we won't go in depth on that information. However, let's see how ultrasound fits into the radiology department as a diagnostic imaging modality rather than a therapeutic concept. In the following sections, we'll cover the clinical applications of diagnostic ultrasound, principles involved, effects of ultrasound to tissue, transducer characteristics, and modalities of displaying and recording the image.

630. Clinical applications and principles

Diagnostic ultrasound has cemented its place in the MTF by becoming a well-rounded and convenient method of imaging internal anatomy and physiology. Diagnostic ultrasound, also known as sonography, is used in much the same way as the other imaging modalities previously discussed. Ultrasound procedures are relatively inexpensive, have no known side effects (as it is non-ionizing radiation), and can examine almost every area of the body. The benefits of diagnostic ultrasound have led it to be a highly popular imaging modality, second only to standard radiography. It is particularly well suited for the examination of soft tissues and for looking at blood flow in arteries and veins.

Clinical applications

For ease of understanding, we'll break down the types of ultrasound testing areas into three categories: general tests, echocardiography, and vascular tests.

General tests

Some of the specific procedures for diagnostic ultrasound are:

- Head, neck, and chest areas.
- Breast—used when a mass has been identified by mammogram or physical exam, and to determine if the mass is solid or cystic (fluid-filled).
- Ultrasound guided biopsy (performed by the radiologist using a thin needle to obtain a small tissue sample).
- Abdomen, including liver, gall bladder, spleen, and pancreas.
- Kidneys, including renal and abdominal vessels (especially abdominal aorta), and retroperitoneal lymph nodes.
- Obstetrics—used to view fetal anatomy, including spine, cranial structures, heart, liver, stomach, urinary bladder, kidneys, and long bones.
- Female pelvis, including uterus, ovaries, vagina, cul-de-sac, and urinary bladder in various planes with measurements.
- Male pelvis, including urinary bladder, prostate, and seminal vesicles.
- The scrotum and testicles.

Echocardiography

Echocardiography (often simply called echo) tests are specifically designed to image the heart and surrounding structures. It can be quite helpful in establishing a specific diagnosis and estimating the severity of various cardiac diseases. Results of an echocardiogram are combined with information about a patient's history and results from a physical exam, cardiac and pulmonary auscultation, thoracic radiographs, and other ancillary tests to determine a patient's complete diagnosis.

In general, echocardiography evaluates cardiac chamber size, wall thickness, wall motion, valve configuration and motion, and the proximal great vessels. Using ultrasound, anatomic relationships can be determined and information regarding cardiac function can be derived. This technique provides a sensitive method for detecting pericardial and pleural fluid accumulation, and can allow identification of mass lesions within and adjacent to the heart. There are three types of echocardiography used to evaluate the heart—M-mode, 2D, and Doppler (we will discuss modes of operation in the next lesson).

Vascular tests

Vascular ultrasound is a technique for visualizing arteries and veins in the body using ultrasonic principles. There are several different ultrasound modes or techniques used in examining blood vessels. 2D ultrasound images directly visualize the vessel walls and lumen, and can reveal plaques causing narrowing in arteries, dilatations of arteries known as aneurysms, clots within the blood vessels. Duplex scans use the Doppler principle to study the velocity, direction, and character of flowing blood through the vessels. Echoes returning from a moving target (in this case blood) experience a shift in frequency proportional to the velocity of blood flow. The frequency shift can be analyzed by the system's computer to estimate the velocity of blood flow. Narrowing in arteries is associated with increased velocities; therefore, an estimate of the severity of a narrowing can be obtained from the velocity of blood flow recorded at the site of narrowing. Color Doppler mode superimposes Doppler information on the 2D images by color-coding flowing blood according to the direction and velocity of flow. This provides an instantaneous graphic display of the character of blood flowing through a vessel. Using a combination of the above techniques, it is possible to document the presence and severity of narrowing and complete occlusions; the presence of blood clots in the vessels; the extent and size of aneurysms or varicosities; the presence of abnormal connections between arteries and veins; and any anatomic variations in vascular anatomy. Using ultrasound for vascular testing usually involves several different tests. Some of them focus on arteries and others on veins. Physicians choose these tests first because they are noninvasive, have no

radiation exposure, and are less expensive than other invasive diagnostic procedures. Let's briefly discuss some of the more common vascular tests.

Carotid artery imaging

This is an ultrasound of the neck arteries to evaluate atherosclerotic plaque formation and blood flow characteristics. These arteries are the primary blood supply to the brain and, if occluding, could lead to stroke symptoms (i.e., dizziness, fainting, visual/speech problems, and confusion or memory loss).

Arterial Doppler

This is a test used to grade the quality of blood flow in the vessels of the upper and lower extremities. This test helps to locate segments of limbs that have significant plaque buildup by using blood pressure cuff measurements along the length of the limb.

Arterial duplex imaging

This test is usually done to follow up an arterial Doppler—usually to further investigate diseased vessels identified on the arterial Doppler. By using an ultrasound probe, this test measures the blood as it flows through the narrowed area.

Renal artery Doppler

This is an ultrasound of the aorta and most of its major branches, including the renal arteries in their entirety. Plaques are identified and blood flow characteristics are measured to determine if there is evidence of renal artery narrowing or blockage.

Venous duplex imaging

This test is performed to find blood clots in the deep veins of the upper and lower extremities. A series of external compressions and blood flow imaging is done to rule out a blood clot, which can be clinically characterized by calf pain, redness, hot sensation, and swelling.

Principles of ultrasound

Ultrasound applications are based on the pulse-echo principle. We can easily explain this principle by comparing it to a person who stands on one side of a canyon and shouts "Hello!" toward the other side. The shout would be the "pulse." That pulse travels through the air at about 331 meters per second until it hits the opposite wall of the canyon. It then bounces back (reflected) toward the person (the source of the pulse). The pulse becomes an "echo" because it has been reflected. The echo travels at the same speed on its return trip. This, basically, is the pulse-echo principle. Therefore, we can say ultrasonography works on the principle of transmitting sound through a medium and detecting any echoes.

One remarkable use of the pulse-echo principle is the ability to measure distances between the source of the sound pulse and the object that causes the reflection. We can describe this advantage by relating to the person at the canyon. We can judge the distance the sound traveled by multiplying the number of seconds it took for the shout to be echoed by the speed of sound in the air. If the echo is heard in 4 seconds, then the sound traveled 1,324 meters (4 seconds multiplied by 331 meters per second equals 1,324 meters). Because 1,324 meters is our total distance the sound traveled (two trips across the canyon), simply divide by 2 to get the distance of one trip across the canyon. Measuring distances by sound is not new. Some animals, such as bats, have this ability. This principle has been used for years by naval forces with sound navigation and ranging (SONAR) to detect submerged submarines. Only recently has it been used for medical diagnosis.

Ultrasound terms

Comparing our description of the pulse-echo principle to ultrasonography can help clarify some ultrasonography terms. Ultrasound equipment needs something to create a sound pulse and to "hear" the echo; this device is known as a transducer. Just as the object in our example reflected a person's voice, an object also must reflect a pulse of ultrasound. We would call this striking an interface, which is a surface that forms the common boundary between two parts of matter or space. An

interface occurs whenever two tissues that have different acoustic impedance are in contact with each other. Acoustic impedance of a tissue is the product of the density of the tissue and the speed of sound in the tissue. Since the speed of sound in tissue is constant, we can say the only thing that affects the acoustic impedance is the density of tissue. For our purposes, we can presume acoustic impedance is the same thing as the density of tissue.

Interaction of ultrasound and tissue

Sound traveling through air, such as the "hello" of our friend at the canyon, has a pretty uncomplicated life. It zips across the canyon, hits the far wall, and returns. Probably the only noticeable change is in the strength of the echo; it is not nearly as loud as the shout. An ultrasonic pulse traveling through soft tissues also undergoes modifications.

The most significant change of the ultrasound beam is attenuation. For our purposes, attenuation is the progressive weakening of the sound beam as it travels through tissue. As with our echo at the canyon, the farther the sound travels through tissue, the weaker it gets. The sound beam attenuates primarily through three processes: absorption, reflection, and scattering.

Absorption

Absorption of the sound beam refers to an all-or-nothing phenomenon, as in the photoelectric absorption of an X-ray. It occurs when the tissue captures (absorbs) the sound energy. Most of this captured sound energy changes to heat within the tissue. The absorption process is the basis for ultrasound diathermy, a common therapeutic use of ultrasound. At the lower energy levels used in diagnostic ultrasound, this biological effect of absorption is minimal. Absorption of the sound beam increases as the frequency increases. This results in less penetration of the sound beam.

Reflection

Reflection is the redirection of some of the ultrasound beam toward its source, the transducer. Reflection produces the echoes that form the basis of diagnostic ultrasound scanning, the pulse-echo principle. Whenever a sound beam passes from a tissue of one acoustic impedance to a tissue of different acoustic impedance, a small portion is reflected and the rest of the beam continues on (fig. 3–39). The transmitted beam passing through the interface leaves that interface at a slightly different angle from that of the incident beam. This deviation, called refraction, is sometimes confused with reflection. Your principal interest lies in the reflected beam or, actually, its intensity relative to the incident beam.

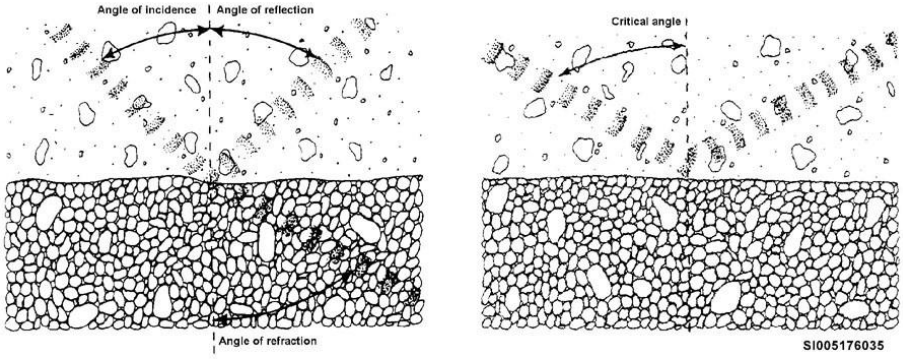

Figure 3–39. Reflection, refraction, and critical angle.

The refracted beam continues on to strike another interface and produces another echo. This repeats until its energy is depleted. In soft tissues, refraction is so small it is of no concern to you. But when you scan across a soft tissue-bone interface, refraction greatly increases.

Scattering

As the name suggests, a portion of the beam is "scattered" in all directions (fig. 3–40). This occurs when the beam encounters an interface that is irregular and smaller than the sound beam. Since the interfaces that produce scattering are small, only a small percentage of the beam is affected. Therefore, you need not be concerned too much with scattering, other than to know that it does occur.

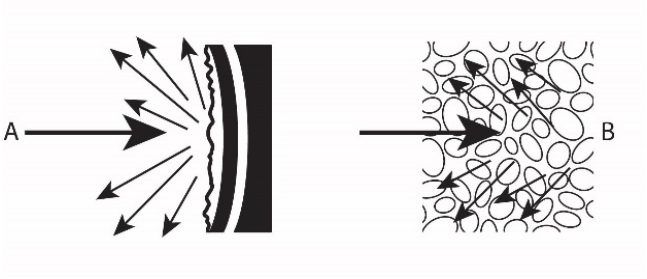

Figure 3–40. Scattering in ultrasound.

Biological effects of ultrasound

Unlike radiography (which uses ionizing radiation), ultrasound uses non-ionizing energy to demonstrate and record data. For this reason, ultrasound is considered to be a "safe" diagnostic tool. However, this consideration must be taken with caution because little is known about the biological action of ultrasound.

The law of conservation of energy states that matter and energy can neither be created nor destroyed; only changed from one form to another. Since ultrasound delivers energy to tissues, there must be some sort of biological action. The manner in which a biological effect is produced is called the mechanism of action. For ionizing radiation, the mechanism of action is ionization and excitation. With ultrasound, the mechanism of action is in three forms: thermal effects (temperature elevation), cavitation, and various viscous stresses.

Thermal effects

Ultrasound can raise the temperature of tissue through the molecular agitation and relaxation produced by sound waves. An excessive power level may raise tissue temperature and damage the molecular structure and membranes.

Cavitation

Cavitation is the formation of tiny gas bubbles or cavities in tissues as a result of violent relaxation forced upon the molecules.

Viscous stresses

The viscosity of tissue on each side of an interface is probably not equal. As ultrasound interacts along the interface, a "viscous stress" exerts on the boundary. Within cellular layers near the boundary, small-scale fluid motions called microstreamings are produced. These stresses can disrupt membranes and cells near the interface.

Dose-response relationships

The power output or intensity of the ultrasound beam is measured in watts per square centimeter (W/cm^2) when biological effects are considered. The power level of diagnostic ultrasound scanners falls in the range of 1 – 20 milliwatts per square centimeter (mW/cm^2; a milliwatt is one-thousandth of a watt). The absolute minimum dose level reported for observable effects in experimental tissues is 100 mW/cm^2, and then only after many hours of continuous ultrasound exposure. Of course, if the intensity is increased, the exposure time factor is decreased. Fortunately, our diagnostic power levels are far below any damaging level of dosage.

631. Ultrasound instrumentation and operation

New technologies are continuing to develop in the field of ultrasound, however, the general concepts remain the same. No matter how complex a machine gets, we can break a unit into various parts for ease of understanding. The general components of an ultrasound machine are: transducer, CPU, display, keyboard, storage, and image capturing devices. In this lesson, we will discuss the ultrasound components, as well as the many modes of operation and the ultrasound imaging phantom. The heart of any ultrasound unit is the transducer, so it makes sense to begin there.

Ultrasound transducer

We have said ultrasound equipment needs something to create a sound and "hear" the echo. The device that does this is a transducer. Specifically, an ultrasound transducer converts electrical energy into sound energy and sound energy into electrical energy. Its operation is based on the piezoelectric effect. When a suitable crystalline material is electrically stimulated, it expands. When the polarity of the electrical signal reverses, the crystal contracts. This repetitive motion causes the crystal to oscillate.

Ultrasound equipment operates at a high electrical frequency that causes the crystal to oscillate at a high frequency. It is this very rapid motion of the crystal that produces the ultrasound wave. More precisely, an ultrasound transducer converts electrical signals into mechanical motion and the mechanical motion into sound waves. The reverse is also possible. Indeed, we must have a means to "hear" our echo sounds to have pulse-echo ultrasound. The transducer, with its crystal and because of the piezoelectric effect, also serves this purpose. When an echo reaches the crystal, its energy causes a slight compression and expansion of the crystal. This oscillation produces a weak electrical signal that is processed and becomes an image on the display.

Transducer components

An ultrasound transducer (fig. 3–41) is made up of several components. The case provides structural support for the internal filling and allows the ultrasonographer to manipulate the transducer without damaging it. It also insulates the patient from electrical shock. An electrical cable enters the case through a connector, with two electrode leads that deliver the electrical charge to the piezoelectric crystal. It also receives the electrical charge of the converted mechanical echo and conveys this charge back to the machine to form the image. The crystals are made of ceramic materials. Since the crystal vibrates when activated by an electrical stimulus or sound pressure, it must not be allowed to reverberate or ring too long. Since the transducer must also receive the returning echo sound, the crystal must stop its motion (to listen). Most crystals used today are man-made and use lead, zirconate and titanate (commonly called PZT). A backing material in the case dampens the movement of the crystal when the electrical stimulus is removed to reduce reverberation. In pulse-echo ultrasound, transducers emit sound pulses only about 1 percent of the time; 99 percent of the time, the transducer waits for echoes.

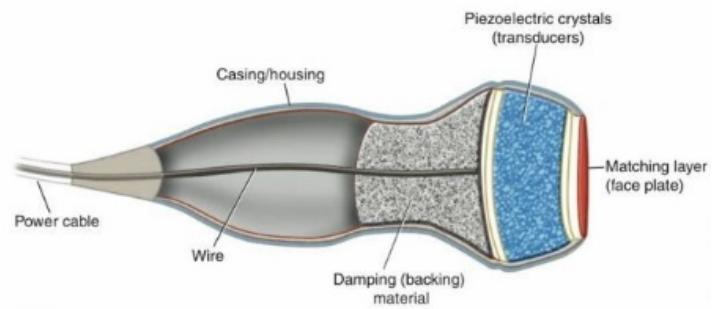

Figure 3–41. Ultrasound transducer. (Reproduced by permission, Carlton/Adler. Principles of Radiographic Imaging, 5E. © 2013 Delmar Learning, a part of Cengage, Inc.)

The size of the crystal is also a prime factor in the operation of a transducer. The thickness of the crystal affects the efficiency of the transmission and reception of ultrasound. Manufacturers slice the crystal so its thickness is a half wavelength. For example, for a 2.5 megahertz (MHz) transducer, the crystal thickness is 0.31 mm. The diameter of the crystal controls the width of the beam, which is very important to resolution and focusing. The faceplate of the crystal provides a protective acoustic window for transmitting the ultrasound to the patient. The faceplate, along with the crystal, can be shaped to focus the beam.

Types of transducers

There are a variety of transducers, each being unique with specific diagnostic values (fig. 3–42). Some are shaped for the organ systems they examine, and each has a range of frequencies for the best possible image. The exam type and patient will determine which transducer is used for an exam.

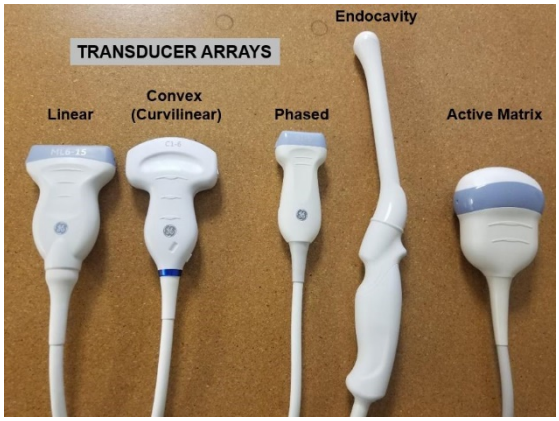

Figure 3–42. Common ultrasound transducer arrays.

Most transducers in use today are arrays. An array contains a collection of elements (crystals) in a single transducer. Each element is connected to its own electronic circuitry called channels. Scanning of the beam is done electronically. As the scanner sweeps across the part, frames of images are acquired. Common types of arrays include linear arrays, phased arrays, convex arrays, annular phased arrays, vector arrays, and matrix arrays.

Linear array

The elements in a linear array are in a line and fire sequentially. There is no steering involved. The elements line up in a straight line that gives a rectangular-shaped image. The width of the image is approximately equal to the length of the array. It also has fixed focusing.

Phased array

Phased array transducers can be focused and steered. The crystals fire almost simultaneously. They can fire in different patterns that focus and steer the beam. Electronic curvature focuses the beam, while electronic slope steers the beam. They produce a fan-shaped or sector-shaped image.

Convexed array (curvilinear)

The crystals in this array are arranged in a curve. They can be a sequential or phased array. The transducer is usually large with a large footprint. Convexed arrays produce a sector-type image.

Annular phased array

Annular phased arrays contain concentric rings cut from the same circular slab of piezoelectric material. The elements are arranged in an arc. Small-diameter rings have a shorter focus but will diverge quickly. Large rings have a deep focal length. They have electronic focusing in all planes at all depths with mechanical steering. An annular phased array produces a fan-or sector-shaped image.

Vector arrays

Vector arrays have groups of elements that fire together to steer the pulses in various directions. The image format can change from rectangular to sector. The footprint is smaller than the convex, and the top of the display is flat. They can be used for linear applications where the elements fire in different directions and not straight down. The rectangular display can change to a parallelogram or a trapezoidal shape.

Matrix arrays

Matrix-array transducers are composed of about 3000 independent piezoelectric elements. These piezoelectric elements are arranged in a matrix configuration within the transducer in order to steer the ultrasound beam electronically. The electronically controlled phasic firing of the elements in that matrix generates a scan line that propagates radially and can be steered both laterally and in the elevation in order to acquire a volumetric pyramid of data.

Mechanical Transducer

Mechanical transducers can be a curve or linear shape. They have a single circular disc-shaped active element. The curvature of the crystal or adding a lens is what focuses the beam at a specific depth. In order to change a focus, the sonographer must change the scanhead. These scanners give a fan or sector-shaped image and a defective crystal will destroy the entire image.

Beam focusing

The primary benefit of beam focusing is the resolution (the ability to identify closely spaced interfaces), which can greatly improve with a focused beam. Ultrasound beams have two zones: Fresnel and Fraunhofer zones. In figure 3–43, notice the Fresnel zone is the near field of the ultrasound beam, while the Fraunhofer zone is the far field of the beam. The focused beam has a constant width, whereas the unfocused beam diverges. The best image resolution is obtained in the near field. On the other hand, the beam's intensity is more uniform within the far field. The divergence of the beam causes a loss of resolution and increases attenuation.

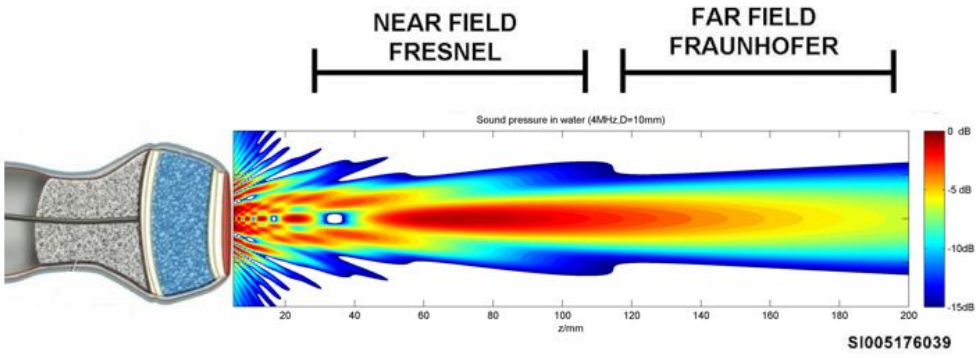

Figure 3–43. Transducer near and far field.

Most transducers are internally focused. This means the crystal itself focuses the beam. Figure 3–44 (A) shows how a beam's shape changes because of the concave shape of the crystal. A beam can also be externally focused (fig. 3–44, B) by using acoustic lenses to change the shape of the beam. These lenses are usually built into the transducer in the path of the beam.

Radiology departments have more than one focused transducer for each system. For example, suppose you are trying to visualize an object that is 9 cm into the abdomen. One type of focused transducer may enable good visualization only within 5 cm, while another may give optimum resolution as deep as 10 cm. For the best results, you would obviously use a transducer with the ability to clearly resolve an object within the desired depth.

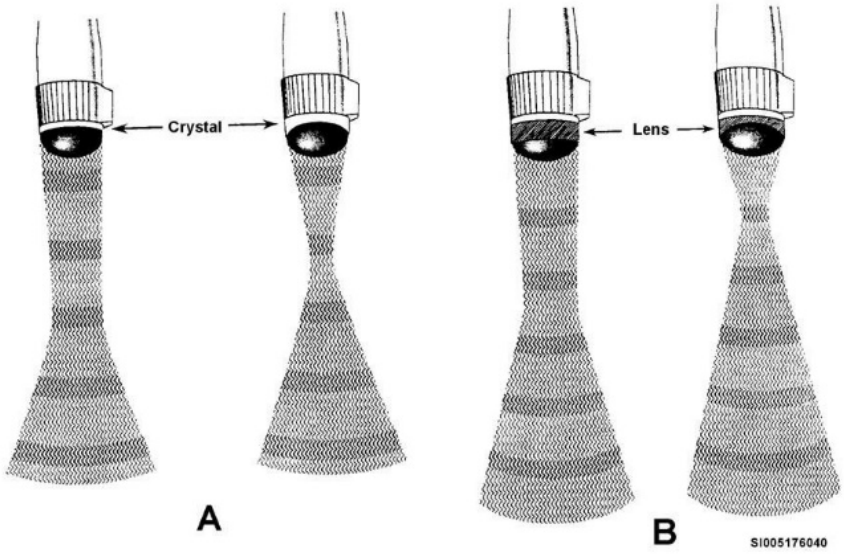

Figure 3–44. Beam focusing.

Central Processor Unit

The CPU is a vital part of any ultrasound equipment, second only to the transducer in importance. As you probably already realize, the CPU is the brain of the ultrasound machine. The CPU is basically a computer that contains the microprocessor, memory, and power supplies, as well as the pulser and receiver circuits. The CPU produces the electrical frequency used by the transducer and also receives the echoes, processes the data, and converts the echoes into usable information that can be displayed on the monitor. Also, in units with advanced features, the CPU allows the sonographer to manipulate the image (much like digital X-ray) to offer a better view of an area of interest. The CPU can also store the processed data or image on disk. The CPU also allows the unit to connect to the PACS network.

Pulser

The pulser produces the electrical signal that excites the crystal into producing the sound wave. It will determine the pulse repetition frequency, pulse repetition period, and the pulse amplitude. It also determines the firing sequence for a phased array. The frequency of the voltage applied to the crystal will determine the resonant frequency of the transducer. Each time voltage is applied to the crystal, it produces a pulse. The greater the voltage amplitude is, the greater the intensity and power in the beam. The sonographer can control the pulser by changing the depth and increasing/decreasing power. Components found in the pulser circuit include a clock generator, a high-voltage pulse generator, and the transducer. The clock generator controls the number of voltage pulses, which activates the crystals. It starts the pulse cycle and sends the timing signals to the pulse generator, time gain compensation circuitry, and memory. The high-voltage pulse generator delivers short, high-amplitude signals to the transducer, which receives the voltage generated by the pulser.

Receiver

The returning echoes convert back into an electrical form by the transducer, and the signal then goes to the receiver. The receiver turns these signals into a form that the display unit can use. Functions of the receiver include amplification, compensation, compression, demodulation, and rejection.

Display

The display is basically a computer monitor that shows the processed data from the CPU. Displays can be black-and-white (monochrome) or color, depending upon the model of the equipment. Of course, high-end equipment—especially that used for echocardiography—will have a color monitor. The display will have control knobs to determine the brightness of the display and the contrast.

Keyboard

Ultrasound machines have a keyboard, and often a trackball, built in to allow the operator to add notes to and take measurements from the data. If the unit has advanced features that allow for manipulation of an image, these controls are used to perform the desired operations.

Storage

The processed data and images can be stored on disk for future reference. The most common types of storage medium are hard disk, compact disc-recordable (CD-R), or digital versatile disc (DVD). Typically, a patient's ultrasound scans are stored on a disk and archived with the patient's medical records. If the unit is part of a PACS, the images may be sent via a network to an archival system.

Image capture system

Many ultrasound systems have some type of image capture device to produce a hard copy or "motion picture" of the scan. Some equipment has thermal printers, used to capture a hard copy of the image from the display. These images may be printed out to give to the attending physician or, in the case of many obstetrical applications, given to the patient for memorabilia. Other systems, especially those used in echocardiography, have a CD or DVD burner, which records multiple patient scans on one disk. Newer models are incorporating digital recorders to capture images or scans, allowing for easy handling and manipulation.

Operational modes

There are several different modes of operation used in diagnostic ultrasound. We will explore five of those modes in this lesson. We'll start with the oldest, the amplitude mode (A-mode).

Amplitude mode

The oldest and simplest imaging mode is the amplitude mode (A-mode). In this imaging mode, the amplitudes of the pressure waves generated by the transducer are displayed as a series of blips on a monitor with regard to the position of the boundaries in the examined tissue. The distance between the blips is proportional to the distance between interfaces; the height of each blip is proportional to the intensity of the reflected beam. Therefore, distal reflections produce smaller blips than do proximal reflections. Today, the A-mode, which only conveys 1D information, is used infrequently, but is still applied in ophthalmology and otolaryngology. Therapeutic ultrasound aimed at a specific tumor or calculus also utilizes A-mode, to allow for a pinpoint accurate focus of the destructive wave energy.

Brightness mode

In brightness mode (B-mode) ultrasound, a linear array of transducers simultaneously scans a plane through the body that can be viewed as a 2D image on screen. This is more commonly known as 2D mode now. 2D echocardiography allows imaging of a plane of tissue (depth and width). In contrast to the A-mode, where the amplitudes are represented in the form of peaks, the amplitudes in B-mode are represented as pixels. The brightness of the pixels corresponds to the intensity of the electrical signals (i.e., the strength of the echoes). Stronger echoes are represented in higher brightness, which explains why this mode is known as brightness mode. Modern diagnostic ultrasound systems normally offer 256 different brightness levels, the so-called grey scales (the human eye is able to distinguish up to 20 different grey levels on the same image). On the screen, the separate pixels are displayed in the form of a line. Before each transmit pulse, the beam is shifted accordingly. It achieves the 2D image by arranging the new lines according to the position of the structures from where the echoes reflected. B-mode can visualize flowing blood and surrounding stationary tissues simultaneously. It is thus an alternative or complement to Doppler ultrasonography in visualizing blood flow.

Motion mode

In motion mode (M-mode), the beam is not shifted as in B-mode, but held in a fixed position over the examined area. Now, the single lines generated on the basis of the echoes returned are arranged on the time-axis to show how the echo data changes over time. With a penetration depth of 15 cm and a scan

time of 0.2 microseconds (μs), up to 5,000 images can be constructed per second, allowing the documentation of very quick motions (e.g., moving heart valves). With the known scale of the graph and the time-axis, the precise documentation of motion, velocity, and acceleration is possible in M-mode, which explains the high importance this imaging mode has gained for echocardiography applications. An advanced feature of M-mode display allows it to be synchronized with an ECG tracing for even better evaluation of cardiac function. The M-mode echocardiograph uses a high sampling rate and can provide cleaner images of cardiac borders. Measurements of cardiac dimensions and motion throughout the cardiac cycle are often more accurately obtained from M-mode tracings, especially when coupled with a simultaneously recorded ECG or phonocardiogram.

Doppler mode

Another method of ultrasonically monitoring the movement of tissue interfaces is based on the Doppler Effect. A familiar example of the Doppler Effect is that of a passing train (fig. 3–45). The sound from the whistle of the train appears very high pitched (high-frequency, short wave length) as it approaches an observer. When the train passes, however, the pitch makes an abrupt change to a lower note or lower frequency.

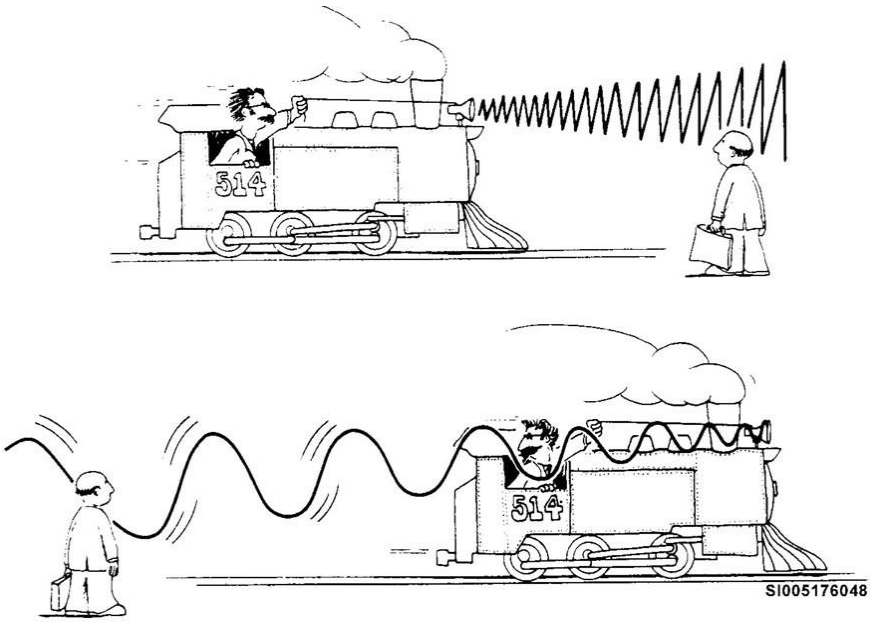

Figure 3–45. Doppler Effect.

Doppler imaging allows evaluation of blood flow patterns, direction, and velocity in vessels; thus, it permits documentation and quantification of valvular insufficiency or stenosis, and cardiac shunts. You can also make estimations of blood flow and cardiac output. The transmitted ultrasound is reflected by the red cells in the blood stream. Doppler is based on detection of frequency changes (the Doppler shift) occurring as ultrasound waves reflect off individual blood cells moving away from or toward the transducer. The difference between the incident frequency and the reflected frequency is the Doppler shift. Calculation of blood flow velocity is possible when the flow is parallel to the angle of the ultrasound beam. Since calculations become increasingly inaccurate as the angle of incidence of the ultrasound beam and the path of blood flow diverges from 0°, measurement of maximal blood flow velocity requires the ultrasound beam be as close to parallel with the path of blood flow as possible. There are two types of Doppler: pulsed wave (PW) and continuous wave (CW). We'll also describe flow mapping, which combines these two types of echocardiography with blood-flow imaging.

Pulsed wave

PW Doppler uses short bursts of ultrasound transmitted to a point at some given distance from the transducer. The advantage of this type of Doppler includes the ability to calculate blood flow velocity, direction, and spectral characteristics from a specified point in the heart or blood vessel. The main disadvantage is the limited ability to measure the maximum velocity due to the limited pulse repetition frequency.

Continuous wave

CW Doppler uses dual crystals, so that ultrasound waves can be simultaneously and continuously sent and received. There is no maximum measurable velocity with CW, so it can measure high velocity flows. The main CW Doppler disadvantage stems from the fact that the sampling of blood flow velocity and direction occurs all along the ultrasound beam, not in a specified area.

Flow mapping

Another type of Doppler is called flow mapping, a form of PW Doppler, which combines the M-mode and 2D modalities with blood flow imaging. However, instead of one sample volume, it analyzes many along multiple scan lines. The mean frequency shift, obtained from these many sample volumes, is color-coded for direction and velocity. Several types of mapping are usually available. Most systems code blood flow toward the transducer as red and flow away as blue. In figure 3–46, the red vessel is showing a positive Doppler shift and is flowing toward the transducer while the blue vessel is showing a negative Doppler shift and is flowing away from the transducer. Differences in the relative velocity of flow can be accentuated, and the presence of multiple velocities and directions of flow (turbulence) can be indicated by different maps, which utilize variations in brightness and color.

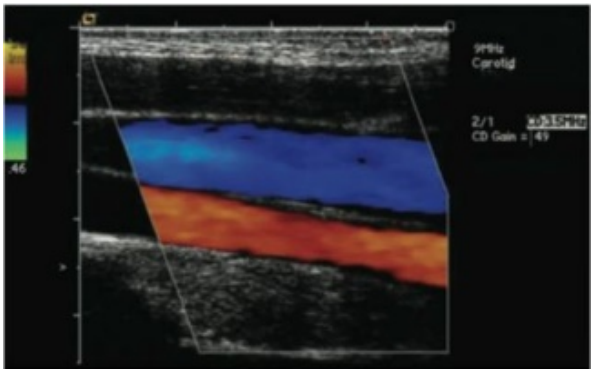

Figure 3–46. Doppler shift – color flow mapping.

Turbulent blood flow is typically seen as a mixture of colors or mosaic pattern that clearly stands out against the homogeneous red or blue flow around it. Flow imaging allows a cardiologist to evaluate specific cardiac chambers and valves for normal and abnormal blood flow. It also helps locate problem areas where Doppler can be used for quantitative measurements.

3D imaging

This type of imaging offers a better view of an organ than the traditional 2D image, which includes all the previously discussed modes. In units with 3D capability, several 2D images are acquired by moving the probe across the body surface or rotating inserted probes. The 2D scans are then combined by specialized computer software to form 3D images.

4D imaging

3D imaging allows us to visualize structures and internal anatomy as static 3D images. However, 4D ultrasound allows us to add live streaming video of the images, showing the motion of the heart wall or valves, or blood flow in various vessels. It is thus 3D ultrasound in live motion. It uses either a 2D transducer, which rapidly acquires 20–30 volumes, or a matrix array 3D transducer.

Ultrasound phantom

The purpose of the ultrasound phantom is to measure the image quality of high resolution ultrasound systems. There are different types of ultrasound phantoms, each designed for specific testing. For the purpose of this lesson, we will cover a typical precision multi-purpose grey scale phantom to illustrate some common phantom functions. Grey scale targets provide for monitoring contrast and temporal resolution, as well as distinguishing the different intensities of brightness and border delineation capabilities of the ultrasound system (fig. 3–47). The ultrasound carefully places targets to measure resolution, depth of penetration, and electronic caliper distance accuracy. Axial resolution pin spacing patterns are small, offering better axial resolution tests. In the multi-purpose phantom, grey scale targets are set at -6 decibel (dB), +6 dB and high scatter, relative to the background material, and with equivalent attenuation properties. There is also a 10 mm anechoic cyst provided to evaluate system noise and geometric distortion.

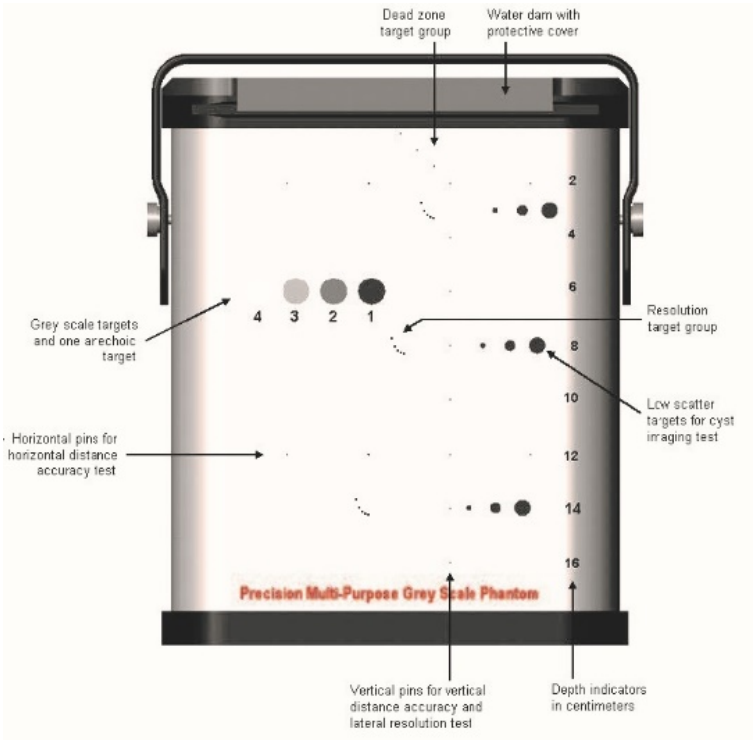

Figure 3–47. Precision multi-purpose grey scale phantom. (Credit to: Gammex Inc.)

The phantom incorporates a tissue mimicking gel, ultrasonically similar to human tissue, which provides the background texture and spacing between targets. This allows the use of normal scanner control settings and ensures that the measured performance closely approximates the scanner's performance in a clinical examination. The water dam on top of the phantom allows you to fill the top with water or ultrasound gel, providing a medium to couple the phantom with the transducer. If using water as the medium, make sure to use distilled water as it introduces less artifacts than tap water. For the best image quality, however, use gel. Now that you know the purpose and construction of the phantom, let's go over some of the tests in which you would use it.

Depth of penetration

The sensitivity of an ultrasound instrument determines the weakest echo signal level that can be detected and clearly displayed. In practical terms, this translates into how far one can "see" into the patient (i.e., the depth of penetration). The frequency of the transducer, and the output power and electrical noise of the system electronics limit the maximum depth of penetration. The maximum depth of penetration should remain constant over time; variations indicate performance degradation.

Changes in the depth of penetration are caused by damage to the transducer or cable, as well as possible malfunctions in the system's transmit and receive circuits. The maximum sensitivity or depth of penetration is determined by measuring the depth (in a tissue mimicking phantom) at which usable echo information disappears. In figure 3–48, you might notice that the tissue echoes near the bottom of the phantom fade into noise. The depth markers on the phantom label will help you determine the depths of the targets.

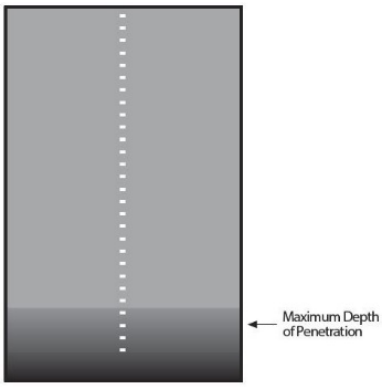

Figure 3–48. Depth of penetration. (Credit to: Gammex Inc.)

Distance accuracy

Vertical and horizontal distance measurement errors are not always obvious and can easily go unnoticed. The vertical distance test determines the accuracy of distances measured along the beam axis. Drift or failure in the system's internal timing circuits can cause vertical distance errors. The horizontal distance test assesses the accuracy of distances measured perpendicular to the beam axis. Horizontal distance errors can be the result of flaws in the transducer geometry, either in its design or through damage. You can assess distance accuracy by comparing the measured distance between selected pin targets in the phantom with the known distance (fig. 3–49). The test distance used should correspond with the distances normally measured in clinical studies.

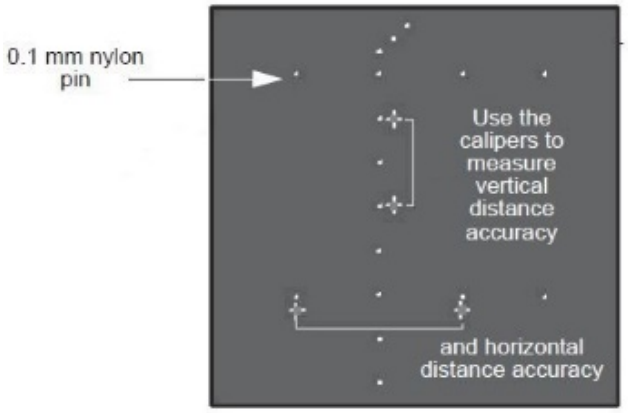

Figure 3–49. Distance accuracy. (Credit to: Gammex Inc.)

Lateral resolution

Lateral resolution describes the instrument's ability to distinguish small, adjacent structures perpendicular to the beam's major axis. Lateral resolution is approximately equal to beam width and varies with depth, the transducer focusing characteristics, and the system's gain and sensitivity settings. Objects smaller than the ultrasound beam are displayed with a width equal to the width of the ultrasound beam at that depth. The lateral resolution of transducers with a fixed focus will vary

noticeably with depth. Systems with multiple focal zones or "dynamic focus" may produce more uniform lateral resolution over a wider range of depths. Loss of transducer elements or problems in the system's beam-forming circuits typically affect lateral resolution. To determine lateral resolution, measure the width of pin targets at depths corresponding to the transducer's near, mid, and far field zones (fig. 3–50). Notice how the pin targets are narrowest in the focal zone. The pin width demonstrates the width of the ultrasound beam at that depth and approximates the lateral resolution of the scanner.

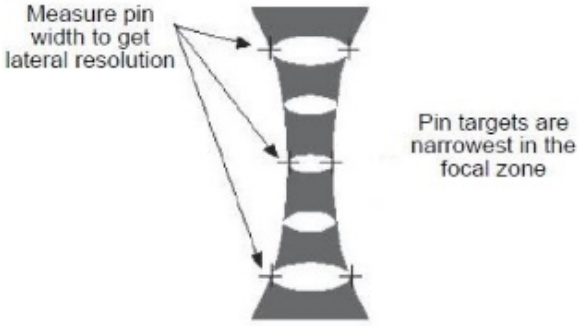

Figure 3–50. Lateral resolution.
(Credit to: Gammex Inc.)

Axial Resolution

Axial resolution describes the scanner's ability to detect and clearly display closely spaced objects that lie on the beam's axis. Axial resolution depends on the transducer's spatial pulse length or pulse duration, which depend on the center frequency and damping factor. The phantom's axial resolution targets contain pin targets with decreasing vertical spacing. Each pin target is offset horizontally by a small distance to avoid shadowing. The system's axial resolution is determined by locating the two pin targets with the smallest vertical separation. Figure 3–51 shows an axial resolution of 1.0 mm, because the 0.5mm to 1.0 mm spacing is the 1st set of pin distances that we can clearly distinguish (or resolve) the targets as two separate objects.

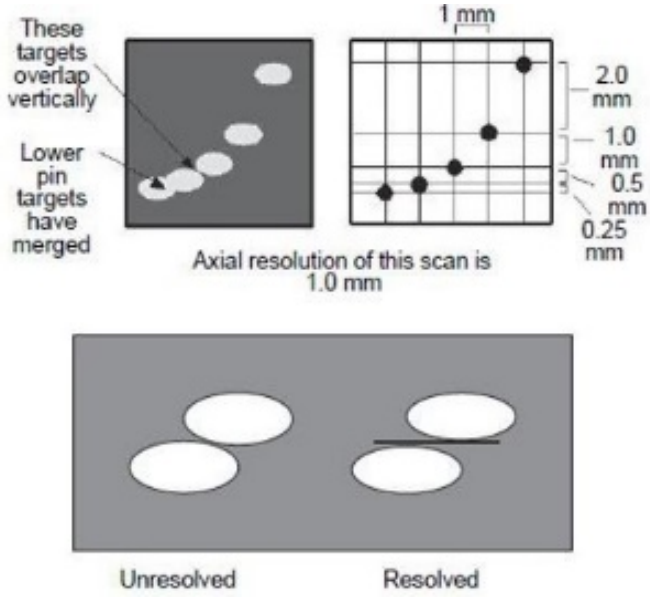

Figure 3–51. Axial resolution. (Credit to: Gammex Inc.)

Image uniformity

Ultrasound systems can experience various image artifacts and non-uniformities. Image non-uniformities are a serious problem because they can mask subtle variations in tissue texture and increase the risk of false negatives. Major non-uniformities should be corrected immediately. Even though one can often "work around" minor non-uniformities, these defects should be seen as potentially large problems and should also be corrected if consistently present. Non-uniformities may be caused by hardware malfunctions such as bad transducer elements or poor electrical contacts in cables or circuit boards. Failures in the image processing circuitry and/or software bugs can also introduce non-uniformities. Poor acoustic coupling between the patient and transducer may also introduce reverberations and other artifacts. Image uniformity is assessed by scanning a uniform region of the tissue mimicking phantom that is relatively free of targets and identifying any deviations from the expected smooth tissue texture (fig. 3–52).

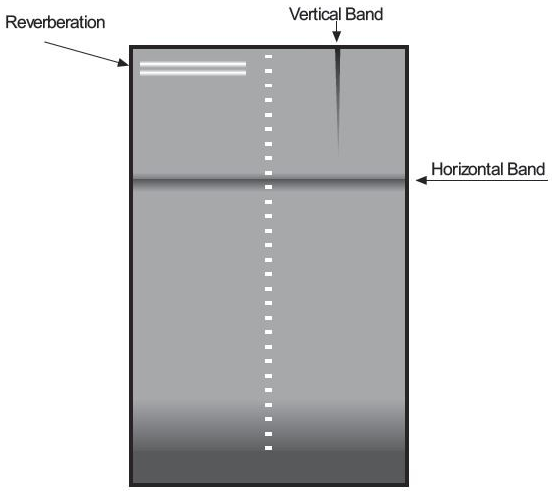

Figure 3–52. Image uniformity. (Credit to: Gammex Inc.)

Dead zone

The dead or "ring down" zone is the portion of the image directly under the transducer where image detail is missing or distorted. The dead zone is the result of reverberations in the transducer and adjacent tissue and the scanner's attempts to compensate for these problems. Although many of today's instruments are normally free from noticeable dead zones, damage to the transducer or poor acoustic coupling may produce this defect. To identify the dead zone, measure the depth of the pin target nearest the transducer. As you can see in figure 3–53, the 1 mm pin target is somewhat cut off, and the transducer cannot clearly visualize targets between the transducer face plate to a 4 mm depth.

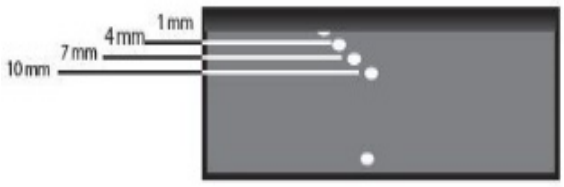

Figure 3–53. Dead zone. (Credit to: Gammex Inc.)

Cyst imaging

The cyst imaging test looks at the system's ability to accurately display a round, negative contrast object. This test combines aspects of contrast resolution and image uniformity into a single test. Cyst image quality can be affected by electrical noise, side lobes in the transducer beam, and problems in

the image processing hardware. Evaluate the smallest cyst in each cyst group that you can easily see in the image and grade the image, with the below criteria in figure 3–54. Because this test is subjective, you should use images from previous tests for comparison.

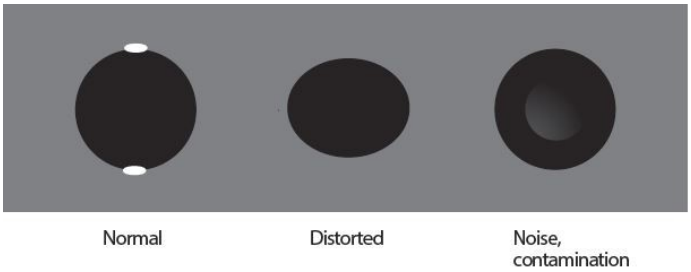

Figure 3–54. Cyst imaging. (Credit to: Gammex Inc.)

Self-Test Questions

After you complete these questions, you may check your answers at the end of the unit.

630. Clinical applications and principles

1. List some benefits discussed for using diagnostic ultrasound in imaging.

2. List at least five general body areas where diagnostic ultrasound is used.

3. In general, what is echocardiography used to evaluate?

4. List the three types of echocardiology used to evaluate the heart.

5. What do 2D vascular ultrasound images visualize?

6. What are some of the more common vascular tests performed with diagnostic ultrasound?

7. Define transducer, interface, and acoustic impedance

8. What are the three processes that attenuate ultrasound?

9. Concerning absorption, what happens to the sound beam as frequency increases?

10. What two things happen when an ultrasound beam passes through tissue of one acoustic impedance into tissue of different acoustic impedance?

11. When does scattering occur in ultrasound?

12. In ultrasound, the mechanism of action is in what three forms?

13. How is the intensity of an ultrasound beam measured and in what range does diagnostic ultrasound fall?

631. Ultrasound instrumentation and operation

1. What are the general components of any diagnostic ultrasound unit?

2. What effect is the transducer based on?

3. What is used within the transducer to prevent the transducer from reverberating?

4. How much time does an ultrasound transducer spend sending a sound pulse versus the time it spends listening for the echo?

5. The thickness of a crystal equates to what factor on a transducer?

6. What does the diameter of the crystal control?

7. What is the faceplate used for?

8. List three common types of transducer arrays.

9. What is resolution in ultrasound?

10. What are the two ways an ultrasound beam can be focused?

11. What is the function of the pulser and what does it determine?

12. What are the most common types of storage medium for storing ultrasound images?

13. What is the oldest and simplest ultrasound operational mode and what is it used for?

14. Briefly describe the B-mode operational mode.

15. What can the M-mode of operation be used to document?

16. What is Doppler mode normally used for in diagnostic ultrasound?

17. What are the two types of Doppler echocardiography?

18. How do most Doppler echocardiography systems code blood flow?

19. Why are ultrasound phantoms filled with a tissue-mimicking gel?

20. What factors limit the maximum depth of penetration of an ultrasound?

21. Briefly describe lateral resolution.

22. How would you use a phantom to determine the axial resolution of an ultrasound unit?

Answers to Self-Test Questions

620

1. (1) b.
   (2) g.
   (3) a.
   (4) e.
   (5) h.
   (6) c.
   (7) d.
   (8) f.
2. Rectilinear.
3. The exposure angle used during the tomogram.
4. (1) FFD.
   (2) Amplitude.
   (3) FOD.
   (4) Distance of the object from the image receptor.
5. The computed radiographic images are manipulated by a post-acquisition algorithm to simulate tomographic exposures instead of actually producing additional images. It combines digital image capture and processing with simple tube/detector motion as used in conventional radiographic tomography. Tomosynthetic processing requires about 10 exposures to establish an adequate image base from which reprocessing can be done.

621

1. A CT image is sharper, more focused, and 3D, which allows better differentiation between tissues than in standard X-rays.
2. Any four of the following:
   (1) Sinus studies—the CT scan can show details of sinusitis and bone fractures. Physicians may order CT of the sinuses to provide an accurate map for surgery.
   (2) Brain studies—brain scans can detect hematomas, tumors, and strokes. The introduction of CT scanning, especially spiral CT, has helped reduce the need for more invasive procedures (i.e., cerebral angiography).
   (3) Body scans—CT scans of the body will often be used to observe abdominal organs, such as the liver, kidneys, adrenal glands, spleen, and lymph nodes, and extremities.
   (4) Aorta scans—CT scans can focus on the thoracic or abdominal aorta to locate aneurysms and other possible aortic diseases.
   (5) Chest scans—CT scans of the chest are useful in distinguishing tumors and in detailing accumulation of fluid in chest infections.
   (6) In obese patients, CT scanning may be more useful than ultrasound, since large amounts of body fat can interfere with ultrasound waves.
3. To scan for abnormal masses, which might be malignant tumors. A CT scan can show the size and shape of a tumor, its precise location in the body, and whether it's solid or hollow.
4. Iodine.
5. It is the process of creating a cross-sectional tomographic plane (slice) of any part of the body. By using X-ray absorption measurements collected at multiple points about the body part's periphery, a computer creates the reconstructed image.
6. Conventional radiography needs a 2 – 5 percent difference; CT needs only a 0.5 percent difference.

7. (1) a.
   (2) e.
   (3) c.
   (4) d.
8. When multiple slices are required to cover a large part of the body using previous technology, the patient table is moved in discrete steps through the plane of the X-ray beam and detector. A single slice is acquired at each position with a time delay between to obtain the image. That process is inefficient and can result in spatial misrepresentations between slices if the patient moves. The spiral technique compensates for this.
9. Speed; it is 5–8 times faster than single-slice helical CT units.
10. Voxels that are a perfect cube; equal in length, width, and height (e.g., 1 x 1 x 1 mm).
11. The ratio of the distance the table moves (feed) during one 360° tube rotation to the total beam collimation.
12. CT fluoroscopy.
13. It uses two X-ray sources (tubes) and two X-ray detectors that expose tissues simultaneously to determine how tissue behaves at different radiation energy levels.
14. It is a composition of small blocks, or cells, arranged in rows and columns.
15. The individual cells become smaller on the screen and the image detail (resolution) increases.
16. A two-dimensional representation of the average density of a volume of tissue.
17. 0.046875 cm or 0.46875 mm.
18. The sum of all μ of the voxels between the X-ray tube and detector.
19. Water = 0; air= –1,000; and dense bone= +1,000.
20. The window width represents the range of scale numbers used for the gray scale and is equivalent to adjusting the contrast of the image.

622

1. The X-ray tube, detector assembly, slip rings, collimators, DAS and also contains the mechanics that provide motion used in the CT unit.
2. To support and move the patient through the central opening of the gantry.
3. The same tissue can be scanned twice or missed altogether.
4. X-ray tube failure.
5. (1) Solid-state scintillation.
   (2) Xenon-gas ionization.
6. 1–8 detectors per centimeter; 1–5 detectors per degree.
7. Approximately 45 percent; detector interspace.
8. Approximately 50 percent; low absorption efficiency.
9. To receive and analyze information from the DAS and convert it into a video form so that an image can be displayed on a monitor.
10. Around 65–68° Fahrenheit.
11. Operator.
12. Scan mode.

623

1. CBCT uses cone-shaped beam, while helical CT uses a fan-shaped beam.
2. One rotation or less.
3. (1) Pulsed.
   (2) Continuous exposure.
4. It produces a higher mAs for the actual CBCT scan.
5. (1) Filtered back projection (FBP) such as the Feldkamp algorithm.
   (2) Algebraic reconstruction techniques (ART) such as iterative reconstruction.
6. Image quality and time (for set up, image acquisition, and image reconstruction).

7. 96% lower.
8. The size of the exit window (collimation), the range of exposure factors, and the amount of beam filtration.

624

1. A γ camera.
2. During an X-ray, radiation comes from the X-ray tube, passes through the body, and is then detected and recorded on film or by a computer. In nuclear medicine procedures, a radioactive substance is introduced into the patient and the nuclear medicine equipment detects the radiation emitted by the patient.
3. It assists in the diagnosis and treatment of heart disease by showing the function of the myocardium.
4. (1) The abdomen; for GI bleeding and other such disturbances.
   (2) The brain; for tumors, aneurysms, or evaluate stroke.
   (3) The breast; to image breast cancers.
   (4) The heart; for coronary artery disease, valve disease, or heart attack.
   (5) The kidneys; for renal function or detect renal tumors.
   (6) The liver/spleen; for cirrhosis or cancer.
   (7) The lungs; for pulmonary embolism (blood clot) or test for smoke inhalation injury in burn patients.
   (8) The skeletal system; for cancer or to test for hidden bone trauma in sports injuries.
   (9) The stomach; for stomach function and to confirm ulcers or cancer.

625

1. (1) h.
   (2) c.
   (3) e.
   (4) i.
   (5) f.
   (6) g.
   (7) b.
   (8) a.
2. The radioactive decay of the radionuclide.
3. Because they can only penetrate a few millimeters of tissue and cannot be detected.
4. Because it tells the physician when the material has localized, when it will be emitting the strongest, and when it will leave the body.
5. (1) Neutron capture.
   (2) Nuclear fission.
   (3) Charged-particle bombardment.
   (4) Radionuclide generators.
6. Any three of the following:
   (1) 99m-Tc.
   (2) 123-I and 131-I.
   (3) Xe.
   (4) Gallium.
   (5) Indium.
   (6) Thallium.
   (7) Fluorine.

626

1. The collimator, scintillation crystal, array of PMTs, PHA, imaging monitor, and control console.
2. To reduce scatter and allow the γ camera to localize the radionuclide in the patient by absorbing and stopping most radiation, except that arriving almost perpendicular to the detector face.

3. Parallel-hole.
4. The lead walls between the holes of a collimator; they absorb γ rays that do not emanate from the direction of interest.
5. NaI(Tl); it is used because of the efficiency with which it converts an ionization event into visible light, its long decay time, and high light output.
6. Converts a light pulse into an electrical signal of measurable magnitude.
7. To discard signals from background and scattered radiation, or radiation from interfering isotopes so only photons known to come from the photopeak of the isotope being imaged are recorded.
8. Energy resolution is the ability to discriminate between light pulses caused by γ rays of differing energies; spatial resolution refers to the ability to display discrete, but contiguous, sources of radioactivity.
9. The rotating head system; it can be used for traditional nuclear imaging studies, as well as SPECT studies.
10. A CT scan only shows anatomy, while a PET scan shows how a body part is functioning.
11. In a PET/CT scanner the patient stays on the same imaging table and undergoes PET and CT in the same session, within minutes of each other. This results in considerably more successful image fusion and improves patient outcomes, particularly in the field of oncology.
12. (1) Geiger counter.
    (2) Sodium iodide well counter.
    (3) Dose calibrator.
13. A radioactive isotope must be calibrated prior to injection into a patient.

627

1.

Body Part	Symptoms	Possible Conditions
Brain	Headaches, dizziness, visual changes, hearing loss, seizures, nausea, history of cancer, autoimmune disease, or tingling in the extremities.	Tumors or other lesions, metabolic disorders, and multiple sclerosis or other debilitating conditions.
Neck	Enlarged lymph nodes and blood vessels, or a palpable mass.	Tumors or other lesions, or vascular or structural abnormalities.
Spine	Back pain, numbness or tingling in the extremities, a history of cancer, or loss of bladder or bowel control.	Herniated or bulging disks, arthritic changes, tumors or other lesions, differences between a post-operative scar or recurrent disk, or structural abnormalities.
Abdomen	History of cancer, pain, loss of organ function, bleeding, cirrhosis of the liver, or hepatitis.	Enlarged lymph nodes, metastasis disease, tumors and other lesions, aneurysms, or structural abnormalities.
Pelvis	Cancer staging, pain, palpable masses, or pregnancy complications.	Enlarged lymph nodes, fibroids, ovarian masses, prostate cancer staging, metastasis disease, testicular cancer, or structural abnormalities.
Musculoskeletal	Pain, swelling, weakness, palpable mass, or a decrease in range of motion in joints or soft tissue.	Torn ligaments or cartilage, edema, arthritic changes, tumors, lesions, or structural abnormalities.

2. The blood.

628

1. (1) Permanent.
   (2) Resistive.
   (3) Superconductive.
2. Approximately 60,000 times stronger.
3. Near absolute zero (-459.67° F).
4. Liquid He.

5. To compensate for irregularities in the magnetic field of the primary magnet.
6. Head, neck, body, extremity, and general-purpose.
7. To archive or retrieve images, or as secondary workstations for scanning.

629

1. Water.
2. $B_{o.}$
3. Magnetic moment.
4. Random; parallel or anti-parallel.
5. Bulk net magnetization, which is all individual magnetic moments added together.
6. The precessional frequency of protons depends clearly on the strength of the magnetic field.
7. It alters the precession of hydrogen nuclei and flips vector M away from its equilibrium position.
8. They become excited and transition from a low-energy state to a high-energy state, and then align against the magnetic field in an anti-parallel orientation.
9. As long as the RF excitation pulse is left on.
10. (1) 90°.
    (2) 180°.
11. Transverse.
12. The return of longitudinal magnetization to equilibrium and transverse magnetization to zero.
13. The nuclei dephase, lose their phase coherence, and precess more randomly.
14. T1 symbolizes longitudinal relaxation; T2 symbolizes transverse relaxation.
15. TR is the amount of time allowed to elapse between successive 90° RF pulses; TE is the time interval that elapses between a 90° RF pulse and the measurement of the first spin echo signal.
16. Short.
17. Long TE.
18. A long TR minimizes T1 and short TE minimizes T2.
19. (1) T1.
    (2) T2.
    (3) SD.
20. Fourier transformation.

630

1. Are relatively inexpensive, has no known side effects, can be used to examine almost every area of the body, and is particularly well suited for examination of soft tissues and for looking at blood flow in arteries and veins.
2. Any five of the following:
    (1) Head, neck, and chest.
    (2) Breast.
    (3) Abdomen, including liver, gall bladder, spleen, and pancreas.
    (4) Kidneys, including renal and abdominal vessels (especially abdominal aorta), and retroperitoneal lymph nodes.
    (5) Female pelvis, including uterus, ovaries, vagina, cul-de-sac, and urinary bladder.
    (6) Male pelvis, including urinary bladder, prostate, and seminal testicles.
    (7) Scrotum and testicles.
3. Cardiac chamber size, wall thickness, wall motion, valve configuration and motion, and proximal great vessels.
4. (1) M-mode.
    (2) 2D.
    (3) Doppler.

5. The vessel walls and lumen and can demonstrate plaques causing narrowing in arteries, dilatations of arteries known as aneurysms, and clots within the blood vessels.
6. Carotid artery imaging, arterial Doppler, arterial duplex imaging, renal artery Doppler, and Venous duplex imaging.
7. A transducer creates a sound pulse then listens for its echo; an interface is a surface that forms the common boundary between two parts of matter or space; and acoustic impedance is the product of tissue density and the speed of sound in the tissue.
8. (1) Absorption.
   (2) Reflection.
   (3) Scattering.
9. Absorption of the sound beam increases.
10. (1) Reflection.
    (2) Refraction.
11. When the beam encounters an interface that is irregular and smaller than the sound beam.
12. (1) Thermal effects.
    (2) Cavitation.
    (3) Viscous stresses.
13. W/cm^2; 1–20 mW/cm^2.

631

1. Transducer, CPU, display, keyboard, storage, and image capturing devices.
2. Piezoelectric.
3. A backing (damping) material.
4. 1 percent versus 99 percent.
5. Half a wavelength.
6. The width of the beam.
7. A protective acoustic window for transmitting the ultrasound to the patient.
8. Any three of the following:
   (1) linear arrays.
   (2) Phased arrays.
   (3) Convex arrays.
   (4) Annular phased arrays.
   (5) Vector arrays.
   (6) Matrix arrays.
9. The ability to separate two closely spaced interfaces.
10. (1) Internally by the shape of the crystal.
    (2) Externally with acoustic lenses.
11. It produces the electrical signal that excites the crystal into producing the sound wave. It will determine the pulse repetition frequency, pulse repetition period, and the pulse amplitude. It also determines the firing sequence for a phased array.
12. Hard disks, CD-Rs, or DVDs.
13. A-mode; used in ophthalmology and otolaryngology.
14. Amplitudes are represented as pixels and the brightness of the pixels corresponds to the intensity of the echoes.
15. Very quick motions, such as moving heart valves.
16. Measuring blood flow patterns, direction, and velocity in vessels.
17. (1) PW.
    (2) CW.

18. Blood flow towards the transducer is coded red; flow away from the transducer is coded blue.
19. Because it is ultrasonically similar to human tissue and provides the background texture and spacing between targets. This allows the use of normal scanner control settings and ensures that the measured performance closely approximates the scanner's performance in a clinical examination.
20. The frequency of the transducer and the output power and electrical noise of the system electronics.
21. The instrument's ability to distinguish small, adjacent structures perpendicular to the beam's major axis. Lateral resolution is approximately equal to beam width and varies with depth, the transducer focusing characteristics, and the system's gain and sensitivity settings.
22. By locating the two pin targets with the smallest vertical separation.

Unit Review Exercises

Note to Student: Consider all choices carefully, select the *best* answer to each question, and *circle* the corresponding letter. When you have completed all unit review exercises, transfer your answers to the Field-Scoring Answer Sheet.

Do not return your answer sheet to AFCDA.

69. (620) What term describes the body layer that appears *sharpest* on a tomogram?
    a. Lever.
    b. Fulcrum.
    c. Focal plane.
    d. Exposure angle.

70. (620) What is the *simplest* tomographic tube movement?
    a. Hypocycloidal.
    b. Rectilinear.
    c. Elliptical.
    d. Circular.

71. (620) Section thickness in a tomogram depends upon
    a. amplitude.
    b. lever length.
    c. rate of travel.
    d. exposure angle.

72. (621) What is *not* a use of computed tomography scans in cancer detection?
    a. To make the final diagnosis on whether an abnormal mass is cancerous or noncancerous.
    b. To show the size, shape, and precise location of an abnormal mass.
    c. To determine whether an abnormal mass is solid or hollow.
    d. To guide a needle biopsy of an abnormal mass.

73. (621) When using a computed tomography scan for gastroenterology procedures, what is the *most commonly* used contrast agent?
    a. Barium sulfate.
    b. Iodine.
    c. Water.
    d. Air.

74. (621) Which generation of computed tomography scanner included a fan shaped beam and a linear detector array?
    a. First.
    b. Second.
    c. Third.
    d. Fourth.

75. (621) Which generation of computed tomography scanner has a rotate-stationary action of the tube and detectors?
    a. First.
    b. Second.
    c. Third.
    d. Fourth.

76. (621) In a computed tomography scanner, what area on a patient corresponds to the pixel on the monitor?
    a. Acquired projection.
    b. Pixel point.
    c. Voxel.
    d. Slice.

77. (621) What does the computed tomography number of zero represent?
    a. Air.
    b. Fat.
    c. Bone.
    d. Water.

78. (621) When you adjust the window *width* for a computed tomography scan, you are actually adjusting the
    a. depth of slice.
    b. resolution.
    c. contrast.
    d. density.

79. (622) What are the three main groups of a computed tomography system?
    a. Computer, control, and imaging.
    b. Generator, tube, and detector.
    c. Tube, computer, and imaging.
    d. Gantry, console, and table.

80. (622) What is the principal cause of a computed tomography scanner malfunction?
    a. Extreme temperatures.
    b. Computer overload.
    c. Excessive humidity.
    d. X-ray tube failure.

81. (622) What are the two types of radiation detectors used in computed tomography scanners?
    a. Solid-state scintillation and photomultiplier tubes.
    b. Xenon-gas ionization and solid-state scintillation.
    c. Photomultiplier tubes and optic crystal.
    d. Optic crystal and xenon-gas ionization.

82. (622) Which mode of operation has precedence on a computed tomography system?
    a. Image reconstruction.
    b. Calculate.
    c. Diagnose.
    d. Scan.

83. (623) What are the two categories of image reconstruction used in cone beam computed tomography (CBCT)?
    a. Filtered back projection (FBP) and the Feldkamp algorithm.
    b. Filtered front projection (FFP) and the Feldkamp algorithm.
    c. FBP and algebraic reconstruction techniques (ART).
    d. FFP and ART.

84. (624) In a gamma camera, the scintillation crystal converts
    a. radiation signals to light.
    b. light signals to radiation.
    c. light signals into a reconstructed image.
    d. radiation signals into a reconstructed image.

85. (624) In a nuclear medicine image, an abnormality in the body may show up as an area of *reduced* radioactivity, which is known as a
    a. hot spot.
    b. cold spot.
    c. gamma spot.
    d. radioactive spot.

86. (624) A stress thallium study is used to observe the
    a. heart.
    b. brain.
    c. abdomen.
    d. skeletal system.

87. (625) If the biologic half-life is 3 hours and the physical half-life is 6 hours, what is the *effective* half-life?
    a. 18 hours.
    b. 12 hours.
    c. 3 hours.
    d. 2 hours.

88. (625) Which method of radionuclide production involves using a cyclotron?
    a. Nuclear fission.
    b. Neutron capture.
    c. Radionuclide generation.
    d. Charged-particle bombardment.

89. (625) Which radionuclide is used for *over* 90 percent of all nuclear imaging procedures in the United States?
    a. Technetium 99m.
    b. Thallium 201.
    c. Gallium 67.
    d. Iodine 123.

90. (626) In a gamma camera system, the collimator is located between the
    a. scintillation crystal and the photomultiplier tube.
    b. patient and the photomultiplier tube.
    c. patient and the scintillation crystal.
    d. the patient and the film.

91. (626) Which collimator is *most widely* used in nuclear medicine laboratories?
    a. Aperture-diaphragm.
    b. Parallel-hole.
    c. Pinhole.
    d. Cone.

92. (626) In general, as the septa in a collimator increase in length, the
    a. resolution increases and the sensitivity decreases.
    b. resolution decreases and the sensitivity increases.
    c. resolution increases and the sensitivity increases.
    d. resolution decreases and the sensitivity decreases.

93. (626) In a gamma camera system, the photomultiplier tube converts
    a. incidental light into brighter light.
    b. light into an electrical signal.
    c. light into photoelectrons.
    d. X-rays into light.

94. (626) Which gamma system has *greater* resolution?
    a. One with 19 hexagonal photomultiplier tubes (PMT).
    b. One with 37 round PMTs.
    c. One with 62 round PMTs.
    d. One with 93 hexagonal PMTs.

95. (626) The nuclear imaging technique that involves the use of radiopharmaceuticals that emit two gamma rays 180° apart is known as
    a. single photon emission computed tomography.
    b. double gamma ray emission tomography.
    c. enhanced nuclear imaging tomography.
    d. positron emission tomography.

96. (626) In a well counter, what reacts to radioactivity?
    a. Crystal.
    b. Inert gas.
    c. Ion chamber.
    d. Photomultiplier tube.

97. (626) What piece of equipment in the nuclear medicine laboratory calibrates an isotope *prior* to injection?
    a. Cold kit.
    b. Well counter.
    c. Dose calibrator.
    d. Radioactivity counter.

98. (627) When performing a magnetic resonance imaging scan on the heart, cardiac leads are used to help eliminate
    a. blood flow.
    b. heart motion.
    c. intercostal space.
    d. excess oxygen in the airway.

99. (627) Concerning breast examinations, a magnetic resonance imaging scan is the procedure of choice for
    a. cancer detection.
    b. cancer screening.
    c. guiding needle biopsies.
    d. evaluating silicon implants.

100. (627) In the technique known as magnetic resonance angiography, what is used as a contrast agent?
    a. Air.
    b. Iodine.
    c. Blood.
    d. Barium.

101. (628) What is *not* used as a *primary* magnet on a magnetic resonance imaging system?
    a. Iron core.
    b. Resistive.
    c. Permanent.
    d. Superconductive.

102. (628) Which type of magnetic resonance imaging magnet has the *disadvantage* of high electrical power consumption?
    a. Iron core.
    b. Resistive.
    c. Permanent.
    d. Superconductive.

103. (628) Which type of magnetic resonance imaging magnet requires a cryogen?
    a. Iron core.
    b. Resistive.
    c. Permanent.
    d. Superconductive.

104. (628) What secondary magnet compensates for irregularities in the main magnetic field?
    a. Shim coil.
    b. Field coil.
    c. Gradient coil.
    d. Radio-frequency coil.

105. (628) What type of coil is used to slightly vary the magnetic field during a magnetic resonance imaging scan, allowing spatial information to be obtained?
    a. Radio-frequency coil.
    b. Gradient coil.
    c. Surface coil.
    d. Ring coil.

106. (629) In what direction do low-energy hydrogen nuclei align themselves when placed in a magnetic field?
    a. With the magnetic field.
    b. Randomly within the field.
    c. Against the magnetic field.
    d. 90° perpendicular to the magnetic field.

107. (629) What vector symbolizes the bulk net magnetization within the magnetic field?
    a. B.
    b. M.
    c. X.
    d. Y.

108. (629) The precessional frequency of protons is determined by the
    a. alignment of the hydrogen nuclei.
    b. strength of the magnetic field.
    c. resonant frequency.
    d. phase coherence.

109. (629) The phase coherence of the hydrogen nuclei occurs as long as the
    a. radio-frequency pulse is on.
    b. gradient coil is on.
    c. body coil is on.
    d. shim coil is on.

110. (629) Spin-spin relaxation is also known as
    a. field relaxation.
    b. transverse relaxation.
    c. longitudinal relaxation.
    d. radio-frequency relaxation.

111. (629) The symbol *T*1 refers to
    a. longitudinal relaxation.
    b. transverse relaxation.
    c. spin density.
    d. spin echo.

112. (630) Diagnostic ultrasound is second in popularity to what imaging modality?
    a. Magnetic resonance imaging.
    b. Computed tomography scan.
    c. Standard radiography.
    d. Nuclear medicine.

113. (630) Which ultrasound test is used to grade the quality of blood flow in the vessels of the upper and lower extremities?
    a. Arterial Doppler imaging.
    b. Venous duplex imaging.
    c. Carotid artery imaging.
    d. Renal artery Doppler.

114. (630) What ultrasound term is used for the product of the density of the tissue and the speed of sound in the tissue?
    a. Speed of transmission.
    b. Acoustic impedance.
    c. Acoustic interface.
    d. Interface.

115. (630) What are the three processes that attenuate an ultrasound beam?
    a. Absorption, reflection, and scattering.
    b. Acoustic impedance, reflection, and scattering.
    c. Scattering, reflection, and speed of transmission.
    d. Absorption, acoustic impedance, and speed of transmission.

116. (630) What are the three forms of mechanism of action in ultrasound?
    a. Ionization, excitation, and thermal effects.
    b. Excitation, thermal effects, and cavitation.
    c. Cavitation, viscous stresses, and ionization.
    d. Thermal effects, cavitation, and viscous stresses.

117. (630) What is the power range for diagnostic ultrasound scanners?
    a. 1 – 20 milliwatts per square centimeter (mW/cm^2).
    b. 1 – 50 mW/cm^2.
    c. 20 – 50 mW/cm^2.
    d. 50 – 100 mW/cm^2.

118. (631) What is considered the *heart* of a diagnostic ultrasound unit?
    a. Image display.
    b. Transducer.
    c. Crystal.
    d. Lens.

119. (631) The *far* field of an ultrasound beam is called the
    a. focal zone.
    b. Freznel zone.
    c. diverging zone.
    d. Fraunhofer zone.

120. (631) What is the *oldest and least frequently* used ultrasound operational mode?
    a. Brightness.
    b. Amplitude.
    c. Doppler.
    d. Motion.

121. (631) What echocardiographic mode allows a plane of tissue (depth and width) to be imaged?
    a. 2D.
    b. Doppler.
    c. Pulsed wave.
    d. Motion mode.

122. (631) In which ultrasound operational mode is the beam held in a fixed position over the area to be examined?
    a. Brightness.
    b. Amplifier.
    c. Doppler.
    d. Motion.

123. (631) What echocardiographic mode is often coupled with an electrocardiogram recording to provide an accurate measurement of cardiac dimension and motion throughout the cardiac cycle?
    a. 2D.
    b. Doppler.
    c. Pulsed wave.
    d. Motion mode.

124. (631) In echocardiography, blood flow toward and away from the transducer is *normally* represented by the colors
    a. red and blue, respectively.
    b. blue and red, respectively.
    c. green and yellow, respectively.
    d. yellow and green, respectively.

Glossary of Symbols, Acronyms, and Abbreviations

Symbols

α	Alpha
β	Beta
γ	Gamma
π	Pi (equal to 3.14)
μCi	microcurie
μs	microsecond
μ	attenuation coefficient
1ϕ	single-phase
123-I	Iodine 123
131-I	Iodine 131
2ϕ	two-phase
3ϕ	three-phase
99m-Tc	Technetium 99m
Ω	ohms

Abbreviations and Acronyms

2D	two-dimensional
3D	three-dimensional
A	ampere
AB	accreditation body
ABC	automatic brightness control
ABS	automatic brightness stabilization
AC	alternating current
ACR	American College of Radiology
ADC	automatic dose control/analog-to-digital converter
AE	application entity
AEC	automatic exposure control
AF	Air Force
AFMOA	Air Force Medical Operations Agency
AGC	automatic gain control
Al	aluminum
A-mode	amplitude mode
ART	algebraic reconstruction technique
a-Si:H	hydrogenated amorphous silicon
Bq	becquerel

BMET	biomedical equipment technician
B-mode	brightness mode
B_O	strength and direction
C	Celsius
C/kg	coulomb per kilogram
CAT	computerized axial tomography
CBCT	cone beam computed tomography
cc	cubic centimeter
CC	craniocardal projection
cc/sec	cubic centimeters per second
CCD	charge-coupled device
CD	compact disc
cd/m2	candela per square meter
CDC	career development course
CD-R	compact disc-recordable
CFR	Code of Federal Regulations
CHCS	Composite Healthcare Computer System
Ci	curie
cm	centimeter
CMOS	complementary metal oxide semiconductor
CMX	complete mouth survey
CPU	central processor unit
CR	computed radiography
CRT	cathode ray tube
CsI	cesium-iodide
CT	computerized tomography
CTAT	computerized transverse axial tomography
CTDI	computed tomography dose index
CW	continuous wave (doppler)
DAS	Data Acquisition System
dB	decibel
DC	direct current
DE	dose equivalent
DEL	detector element
DICOM	Digital Imaging and Communications in Medicine
DLP	dose length product

DOD	Department of Defense
dps	disintegrate per second
DR	digital radiography
DVD	digital versatile disc
ECG	electrocardiogram
ECN	equipment control number
EDF	equipment data file
ENT	ear, nose, and throat
Ep	primary voltage (of a transformer)
ER	emergency room
Es	secondary voltage (of a transformer)
eV	electron volts
F	Fahrenheit
FBP	filtered back projection
FC	footcandle
FDA	Food and Drug Administration
FDG	fluorodeoxyglucose
FFD	focal film distance/focus to film distance
FID	free induction decay
fL	foot-lambert
FOD	focus to object distance
FOV	field of view
FPD	flat panel detector
G	Gauss
GB	gigabyte
GI	gastrointestinal
Gy	gray
H_2O	two hydrogen atoms bound to one atom of oxygen
He	helium
HIS	hospital information system
HL7	Health Level 7
HT	high tension
HTTP	Hypertext Transfer Protocol
HTTPS	Hypertext Transfer Protocol Secure
HU	heat unit
HVL	half value layer

Hz	hertz
ICRU	International Commission on Radiation Units and Measurements
ICU	intensive care unit
IP	Internet Protocol
IV	intravenous
J	Joule
K	Kelvin
keV	kiloelectronvolts or thousands of electron volts
kHz	kilohertz
kV	kilovolt
kVp	kilovolt peak
kW	kilowatt
LAN	local area network
LCD	liquid crystal display
LOR	line of response
lx	lux
M	bulk net magnetization
mA	milliampere
mA meter	milliammeter
mAs	mA x time
MB	megabyte
mCi	millicurie
MDCT	multidetector helical computed tomography
MERC	medical equipment repair center
MeV	megaelectronvolts or millions of electron volts
MHz	megahertz
ml	milliliter
mm	millimeter
mm Al	millimeters of aluminum
M-mode	motion mode
Mo	molybdenum
MQSA	Mammography Quality Standards Act
mR	milliroentgens
MR	magnetic resonance
MRA	magnetic resonance angiography
mrem	bulk net magnetization radiation equivalent man
MRI	magnetic resonance imaging
ms	millisecond
MTF	medical treatment facility

MUGA	multiple gated acquisition
mW/cm^2	milliwatts per square centimeter
NaI(Tl)	Thallium-activated sodium iodide
NIC	network interface card
NiCad	nickel-cadmium
NO	normally open (relay contacts)
OID	object to image distance
OSL	optically stimulated luminescence
p/n	positive/negative
PACS	picture archiving and communication system
PBL	positive beam-limiting
PC	personal computer
PCRI	post calibration radiation inspection
PET	positron emission tomography
PHA	pulse height analyzer
PHT	primary of the high-tension
PMT	photomultiplier tube
POC	point of contact
psi	pounds per square inch
PW	pulsed wave (doppler)
Q	energy
QA	quality assurance
QC	quality control
R	roentgen
R/F	radiography/fluoroscopy
rad	radiation absorbed dose
RAID	redundant arrays of independent disks
RC	radius of the area covered
REM	radiation equivalent man
RF	radio-frequency
RFID	radio-frequency identification
RGB	red, green, and blue
RIA	radioimmunoassay
RIS	radiology information system
ROI	region of interest
rpm	revolutions per minute
RT	reconstructive tomography
SCR	silicon-controlled rectifier
SD	spin density
SHT	secondary of the high-tension

SI	International System of Units
SID	source-to-image distance
SMPTE	Society of Motion Picture and Television Engineers
SOD	source-to-object distance
SONAR	sound navigation and ranging
SPECT	single photon emission computed tomography
SQL	Structured Query Language
Sv	sievert
T	Tesla
T1	longitudinal relaxation time constant
T2	transverse relaxation time constant
tan	tangent
TB	terabyte
TE	echo delay time or echo time
TFT	thin-film transistor
TJC	The Joint Commission
TLD	thermoluminescent dosimeter
TR	time of repetition
Tr	turns ratio (of a transformer)
V	volt
VA	Department of Veteran Affairs
VAC	volts alternating current
W	watt
W/cm^2	watts per square centimeter
WAN	wide area network
Xc	capacitive reactance
Xe	xenon
X$_L$	inductive reactance
X-ray	X-radiation
Z	atomic number of a material

Student Notes

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